Acute effects of suspension training and other perturbative sources on lower limb strength tasks

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UNIVERSITAT RAMON LLULL

DOCTORAL THESIS

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Department	Sport Sciences
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Some insights

"In dialogue, the one who acknowledges the truth is not defeated, but learns"

Ramon Llull (1296)

"Achieving what you have dreamed of makes you happy, but above all, it makes you happy to remember the effort you put in to achieve it."

Rafael Nadal (2018)

"Talent depends on inspiration, but the effort depends on the individual"

Pep Guardiola (2015)

"*If you don't work for your dreams, someone else will hire you to build theirs"*

Steve Jobs (2005)

ABSTRACT

Abstract Catalan version

Actualment, els dispositius de suspensió són un dels materials més utilitzats per produir pertorbació i enfortir de forma global la majoria de grups musculars. Encara que, manquen evidències dels seus efectes sobre l'extremitat inferior. Així, l'objectiu principal d'aquesta tesi doctoral va ser quantificar la producció de força, l'activitat muscular i la magnitud de la pertorbació a l'esquat búlgar i altres exercicis de l'extremitat inferior en condicions d'inestabilitat. Es van analitzar 18 estudis per dur a terme una revisió sistemàtica (estudi 1) i 75 participants físicament actius van ser reclutats per realitzar els diferents estudis transversals sobre els efectes dels dispositius de suspensió, les superfícies inestables i les vibracions mecàniques (plataforma vibratòria i vibració superposada) en exercicis de l'extremitat inferior (estudis 2-6). Es va confirmar que l'activació a la part inferior del cos només va ser investigada en el concentrat d'isquiosurals en suspensió (estudi 1). La posició i el ritme d'execució (70 bpm) van ser determinants per la producció de força exercida sobre el tirant de suspensió a l'esquat búlgar (estudi 2). El dispositiu de suspensió a l'esquat búlgar va augmentar les forces verticals contra el terra (estudi 3). Sobre el dispositiu la producció de força va ser major quan el nivell d'inestabilitat era baix (estudi 3 i 4), però a nivell muscular el dispositiu va ser igual de demandant que l'exercici tradicional (estudi 3). Un augment de la pertorbació, va incrementar l'activació muscular (estudis 3, 4, 5) i la magnitud de la inestabilitat per l'esquat búlgar i el mig squat amb barra (estudis 4 i 5). Així, la vibració superposada en un dispositiu de suspensió esdevé un repte per incrementar el nivell de pertorbació i millorar la força, la resistència muscular i l'estabilització (estudi 6). A més, els sensors de força són una eina adequada i usable per valorar les forces exercides sobre els dispositius de suspensió, i l'ús de l'acceleròmetre permet determinar la magnitud de la pertorbació que ofereixen els diferents materials desestabilitzadors mesurant l'acceleració del centre de masses corporal.

Abstract Spanish version

Actualmente, los dispositivos de suspensión son uno de los materiales más utilizados para producir perturbación y fortalecer globalmente la mayoría de los músculos. Aunque, faltan evidencias de sus efectos sobre la extremidad inferior. Así, el objetivo principal de esta tesis doctoral fue cuantificar la producción de fuerza, la actividad muscular y la magnitud de la perturbación en la sentadilla búlgara y otros ejercicios de la extremidad inferior en condiciones de inestabilidad. Se analizaron 18 estudios para llevar a cabo una revisión sistemática (estudio 1) y 75 participantes físicamente activos fueron reclutados para realizar los diferentes estudios transversales sobre los efectos de los dispositivos de suspensión, las superficies inestables y las vibraciones mecánicas (plataforma vibratoria y vibración superpuesta) en ejercicios de la extremidad inferior (estudios 2-6). Se confirmó que la activación en la parte inferior del cuerpo sólo fue investigada en el concentrado de isquiosurales en suspensión (estudio 1). La posición y el ritmo de ejecución (70 bpm) fueron determinantes para la producción de fuerza ejercida sobre el tirante de suspensión en la sentadilla búlgara (estudio 2). El dispositivo de suspensión en la sentadilla búlgara aumentó las fuerzas verticales contra el suelo (estudio 3). Sobre el dispositivo la producción de fuerza fue mayor cuando el nivel de inestabilidad era bajo (estudio 3 y 4), pero a nivel muscular el dispositivo fue igual de demandante que el ejercicio tradicional (estudio 3). Un aumento de la perturbación incrementó la activación muscular (estudios 3, 4, 5) y la magnitud de la inestabilidad en la sentadilla búlgara y la media sentadilla con barra (estudios 4 y 5). Así, la vibración superpuesta en un dispositivo de suspensión se convierte en un reto para incrementar el nivel de perturbación y mejorar la fuerza, la resistencia muscular y la estabilización (estudio 6). Además, los sensores de fuerza son una herramienta adecuada y usable para valorar las fuerzas ejercidas sobre los dispositivos de suspensión, y el uso del acelerómetro permite determinar la magnitud de la perturbación que ofrecen los diferentes materiales desestabilizadores midiendo la aceleración del centro de masas corporal.

Abstract

Nowadays, suspension devices are one of the most widely used pieces of equipment to produce perturbation and strengthen most muscle groups globally. However, there is a lack of evidence of their effects on the lower limb. Thus, the main objective of this doctoral thesis was to quantify force production, muscle activity and the magnitude of perturbation in the Bulgarian squat and other lower extremity exercises under unstable conditions. Eighteen studies were analysed for a systematic review (study 1) and 75 physically active participants were recruited to perform the different cross-sectional studies on the effects of suspension devices, unstable surfaces, and mechanical vibrations (vibration platform and superimposed vibration) on lower limb exercises (studies 2-6). It was confirmed that lower body activation had only been previously investigated in the suspended hamstring curl (study 1). Position and pace (70 bpm) were determinants for the force exerted on the suspension strap in the Bulgarian squat (study 2). The suspension device in the Bulgarian squat increased the vertical ground reaction forces (study 3). The force production was higher on the device when the level of instability was low (study 3 and 4), but for muscle activity the device was just as demanding as a traditional exercise (study 3). Increased perturbation enhanced muscle activation (studies 3, 4, 5) and the magnitude of instability in the Bulgarian squat and barbell half-squat (studies 4 and 5). Thus, superimposed vibration on a suspension device becomes a challenge to increase the level of perturbation and improve strength, muscular endurance, and stabilisation (study 6). In addition, load cells are a suitable and practical tool to assess the forces exerted on suspension devices, and the use of an accelerometer makes it possible to determine the magnitude of the perturbation offered by different equipment providing instability by measuring the acceleration of the body's centre of mass.

List of publications

The studies included in the PhD thesis are shown below. The appendix section shows all publications. Open access publications are shown in the original format, and limited access publications are shown in the format of the accepted version. Besides, this PhD thesis also includes a patent registration, and the most relevant information is shown below. The appendix section shows a detailed description of the device and the patent application.

STUDY 1 – limited access

Title: Muscle activation in suspension training: a systematic review

Citation: Aguilera-Castells, J., Buscà, B., Fort-Vanmeerhaeghe, A., Montalvo, A. M., & Peña, J. (2020). Muscle activation in suspension training: a systematic review. *Sports Biomechanics*, *19*(1), 55–75.

Doi: 10.1080/14763141.2018.1472293

STUDY 2 – open access

Title: Suspended lunge exercise: assessment of forces in different positions and paces

Citation: Aguilera-Castells, J., Buscà, B., Peña, J., Fort-Vanmeerhaeghe, A., Solana-Tramunt, M., & Morales, J. (2019). Suspended lunge exercise: assessment of forces in different positions and paces. *Aloma: Revista de Psicologia, Ciències de l'educació i de l'esport Blanquerna*, *37*(1), 57–64.

Doi: 10.51698/aloma.2019.37.1.57-64

STUDY 3 – open access

Title: Muscle activity of Bulgarian squat. Effects of additional vibration, suspension and unstable surface

Citation: Aguilera-Castells, J., Buscà, B., Morales, J., Solana-Tramunt, M., Fort-Vanmeerhaeghe, A., Rey-Abella, F., Bantulà, J., & Peña, J. (2019). Muscle activity of Bulgarian squat. Effects of additional vibration, suspension and unstable surface. *PLOS ONE*, *14*(8), e0221710.

Doi: 10.1371/journal.pone.0221710

STUDY 4 – open access

Title: Correlational data concerning body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions in high-standard track and field athletes

Citation: Aguilera-Castells, J., Buscà, B., Arboix-Alió, J., McEwan, G., Calleja-González, J., & Peña, J. (2020). Correlational data concerning body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions in high-standard track and field athletes. *Data in Brief*, *28*, 104912.

Doi: 10.1016/j.dib.2019.104912

STUDY 5 – open acces

Title: Influence of the amount of instability on the leg muscle activity during a loaded free barbell half-squat

Citation: Buscà, B., Aguilera-Castells, J., Arboix-Alió, J., Miró, A., Fort-Vanmeerhaeghe, A., & Peña, J. (2020). Influence of the amount of instability on the leg muscle activity during a loaded free barbell half-squat. *International Journal of Environmental Research and Public Health*, *17*(21), 8046.

Doi: 10.3390/ijerph17218046

STUDY 6 – open access

Title: sEMG activity in superimposed vibration on suspended supine bridge and hamstring curl

Citation: Aguilera-Castells, J., Buscà, B., Arboix-Alió, J., Miró, A., Fort-Vanmeerhaeghe, A., & Peña, J. (2021). sEMG activity in superimposed vibration on suspended supine bridge and hamstring curl. *Frontiers in Physiology*, *12*, 712471.

Doi: 10.3389/fphys.2021.712471

PATENT

Title: Vibratory system for suspension training

Inventors: Buscà, B., Aguilera-Castells, J., & Peña, J.

Agency: Spanish Patent and Trademark Office (OEPM)

Application number: 202030652

List of abbreviations

%MVIC	Percentage of maximum voluntary isometric contraction
3D	Three-dimensional
4D	Four-dimensional
BCMA	Body centre of mass acceleration
BOSU	Both sides up
BOSU-down	Dome side down
BOSU-up	Dome side up
Bpm	Beats per minute
CMJ	Countermovement jump
COD	Changes of direction
COP	Centers of pressure
EMG	Electromyography
JCR	Journal Citation Reports
mV	Millivolts
MVIC	Maximum voluntary isometric contraction
OEPM	Spanish patent and trademark office
OMNI-Res	OMNI- Perceived Exertion Scale for Resistance Exercise
Q:H	Quadriceps:Hamstrings ratio
RFD	Rate of force development
RPA	Repeated power ability
SCI	SCImago Journal Rank
SD	Standard deviation
SE	Standard error of the mean
sEMG	Surface electromyography
SENIAM	Surface electromyography for the non-invasive assessment of muscles
SJ	Squat jump
TRX	TRX suspension training
VGRF	Vertical ground reaction forces
Vibro30	Vibration at 30 Hz
Vibro40	Vibration at 40 Hz
WBV	Whole-body vibration

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INTRODUCTION

In strength and conditioning, recent trends support functional exercises to improve the efficacy of multidirectional sports skills, enhancing the quality of resistance training. These skills include locomotor, manipulative, and stability actions while maintaining the kinetic chain under control (G. Cook et al., 2006). Traditionally, the exercises included in strength and conditioning programs tended to progress in load in order to improve sports skills because it has been the ideal strategy for increasing muscular demands but, in recent years, different unstable environments have been used with similar purposes (McBride et al., 2006, 2010; Wahl & Behm, 2008). The use of different sources of instability has commonly required different devices to enrich the effects of several exercises on muscle activation, force production, motor control, and consequently, sports performance (V. Andersen et al., 2014; K. Anderson & Behm, 2005b; Behm & Anderson, 2006). The design of these devices is intended to alter the relationship between the base of support, the body's spatial position, and the intention of the athlete to maintain balance during the execution of a task. Therefore, the amount of instability depends on the nature of the task, characteristics of the subject (weight, height, muscle abilities and motor control) and the different features of the device (shape, material, friction, size and display) (Behm & Anderson, 2006). Thus, performing conditioning exercises in an unstable environment, such as BOSU (both sides up), stability balls, rubber discs, freeman plates, or hanging loose objects on the barbell, creates perturbations in whole-body stability, adding a new challenge for somatosensory, vestibular, and visual systems (Taylor, 2011). Perturbed tasks increase the co-contractile activity, and the role of antagonists is enhanced to mitigate the uncertainty produced by the chosen source of instability (Behm et al., 2002).

In this vein, one of the most widely used and popular training methods that contribute to the improvement of sports skills by offering a progression of both the magnitude of the instability and the level of exercise difficulty is suspension training. This destabilizing device has become popular in professional (sports clubs and elite athletes) and recreational fitness practices. It is a highly versatile and portable gadget allowing to perform multiple exercises by modifying the athlete's position and thus the demands of the exercise. These features have turned the suspension devices into one of the most functional and widely used destabilizing pieces of equipment, besides being essential in strength and conditioning, return-to-play, and rehabilitation programs. However, the impact of the instability provoked for the suspension devices on performance parameters such as muscular activation and force has not been widely studied concerning the number of exercises that can be done for both the upper and lower body, and their different variants, such as dynamic, isometric, unilateral, bilateral exercises. Additionally, a very important issue for strength and conditioning coaches and practitioners is the fact that it is not easy to quantify the load in this type of exercise to progress through demands or exercise variations, to determine the magnitude of the instability applied or to get to know the amount of muscle activation or the forces that will be generated in this unstable environment.

In this regard, the thematic unit presented is the unstable environment, mainly with a suspension device and exercises to strengthen the lower body. The studies presented in this PhD thesis have focused on the Bulgarian squat, lunge, and half-squat as knee extensors exercises and the supine bridge and hamstring curl as knee flexors because these exercises are the most widely used in strength and conditioning programs to strengthen the lower limb. The Bulgarian squat and lunge are unilateral exercises connected with different sport skills, such as horizontal and lateral jumps and change of direction. Although the half-squat has been more studied in unstable environments, the Bulgarian squat, lunge, supine bridge, and hamstring curl have been less studied, and the scientific literature does not show a consensus on the effects of unstable surfaces on the activation of the hip and the thigh muscles. Furthermore, the study of the effects of suspension devices have been traditionally focused on the upper body, thus causing a gap with the lower body. Besides, exercises such as the suspended lunge are widely used by strength and conditioning coaches in different places (gym, field, or court) with different purposes, for example, warming up, due to its functionality and similarity with sport actions, although ignoring the precise magnitude of the instability offered by the exercise and its demands.

Therefore, study 1 is a systematic review of muscle activation in suspension training showing the most investigated exercises in suspended conditions, mainly of the upper body, and compared with its counterparts performed under stable conditions (traditional exercise on the floor). Study 2 focuses on assessing of the forces exerted on the suspension strap when a suspended lunge is performed with variations in position, pace and type of contraction. Study 3 compares the Bulgarian squat with the suspended lunge (single instability), the combination of the suspended exercise with an unstable surface (BOSU, dual instability) and the mechanical vibration (dual condition) to evaluate, on the one hand, the muscular activation of the hip and thigh muscles, and on the other hand, the production of force on the suspension strap and the vertical ground reaction forces (VGRF) of the front leg. Study 4 associates the amount of instability (measured by accelerometer) with muscle activation and force production under different conditions of the suspended lunge exercise (single and dual instability). Study 5 provides a protocol to establish the amount of instability under different half-squat conditions (floor, foam, BOSU-up and -down) and measures the muscle activity, the OMNI-Perceived Exertion Scale for Resistance Exercise (OMNI-Res) and correlates the amount of instability with the muscle activity. Finally, study 6 shows the effects of a vibratory system for suspension training, using a patent-pending device (number 202030652, OEPM), on the lower-body muscle activity and OMNI-Res in the suspended supine bridge and suspended hamstring curl exercises under non-vibration, 25 Hz and 40 Hz vibration condition. The previous studies have been published in the Journal Citation Reports (JCR; study 1,3,5 and 6) and SCImago Journal Rank (SJR; study 2 and 4) indexes. The specific contribution of the PhD candidate has been that of the lead author (studies 1-4 and 6) and second author for study 5 with very significant participation in the phases of conceptualization, formal analysis, investigation, methodology, writing-original draft and review-writing.

BACKGROUND

CONCEPTUAL APPROACH TO TRAINING WITH PERTURBATIONS

Traditional strength training in team sports has been based on the periodisation proposed by Bompa & Buzzichelli (2015) and its different phases. Currently, because of the competitive demands determined by a pressured calendar for team sport, athletes have to achieve a high level of performance and play 2 or 3 matches per week (Mujika et al., 2018). For this reason, it is not feasible to devote four weeks to the anatomical adaptation phase or six to eight weeks to the hypertrophy phase (Bompa & Buzzichelli, 2015). Likewise, traditional strength training focuses on improving athletic performance by training muscle groups with very systematic exercises, such as bench presses or knee extensions (Evangelista et al., 2019). In addition, traditional strength training has also included Olympic weightlifting exercises, such as the snatch, clean, and jerk, to improve the vertical vector (Z-axis) in vertical jump performance (Arabatzi et al., 2010; Hackett et al., 2016). Several studies have also used Olympic weightlifting exercises to enhance sprint and agility test performance (Chiu & Schilling, 2005; Hedrick & Wada, 2008; Hoffman et al., 2004; Hori et al., 2005, 2008). However, after eight weeks of Olympic weightlifting training, Tricoli et al. (2005) did not find improvements in the 10-m sprint, 30-m sprint, and agility test performance.

Muscle group-based training for team sports lacks transfer between the trained muscle groups and the demands of the sport concerning the improvement of intramuscular and intermuscular coordination (Brearley & Bishop, 2019; Young, 2006). Different team sports (football, basketball, handball, futsal, and field hockey) require sprinting, high-intensity running, lateral shuffling, or cutting. For instance, football players sprint a distance between 117 m and 831 m, with an average duration of 2 s, and run between 3,000 m and 9,000 m, generating a total number of accelerations >2.5 ms^{-2}, which, per match, correspond to decelerations (between 16 and 32) and accelerations (between 4.8 and 8.0) of high intensity. Cutting or change of direction

(COD) produce these accelerations and decelerations more than 300 times per match, being the 90º COD the most reproduced during a match with 45 and 49 cuttings. Basketball players, on average, sprint between 18 and 105 times per match with an average duration from 0.5 to 2.4 s, covering a distance of 70 to 90 metres. For handball and field hockey players, the sprinting distance is slightly higher between 57 and 168 metres and between 114 and 124 metres, respectively. The sprint duration for basketball players ranges from 0.9 to 3.0 s, and for field hockey players, it is 1.8 s with an average of 7 to 30 sprints per match. Moreover, field hockey players cover between 1652 and 2554 metres running at high intensity. In futsal, the distance covered in sprinting ranges from 308 metres to 422 metres, with a duration between 1.6 and 1.9 s being more prolonged than in the previous sports. The average number of COD per match is higher in basketball (from 997 to 2733) than in handball (from 18 to 37.9). The number of COD performed in handball depends on the player's position. Lateral displacement in futsal is much lower (between 9.6 and 11.0 metres) than in handball and basketball (between 270 and 666 metres). Lastly, the jump action is performed by basketball players between 41 and 56 times per match, and it is much higher than for handball, with an average between 8.2 and 19.1 jumps per match (Taylor et al., 2017).

Demands of sport have led to the evolution of strength training beyond the traditional approach based on muscle groups, and there is a greater focus on the movements and most representative actions of each sport (Boyle, 2017; Tous-Fajardo, 2017). One of the most researched sporting actions is the COD, where athletes must generate large eccentric force productions to decelerate and then accelerate (Chaabene et al., 2018). The inertial flywheel (accommodated resistance) is the most suitable method to improve COD because it allows the reproduction of specific sports actions (Vicens-Bordas et al., 2018). Several studies have shown that inertial flywheel training also increases COD performance compared to traditional training (de Hoyo et al., 2014, 2016; Núñez et al., 2018; Tous-Fajardo et al., 2016). In contrast, to improve straight sprinting performance, horizontal force-velocity profiling needs to be considered, where the dominance of the exercise will vary according to the velocity of the movement and the load: 1) velocity dominant exercise (maximal velocity sprinting), 2) power dominant velocity (squat jump), and 3) force dominant velocity (resisted sprinting with a prowler sledge) (Hicks et al., 2020).

Moreover, for most sporting actions (COD, sprinting, lateral actions), except for vertical jumping, the horizontal force vector is a significant determinant of performance (Randell et al., 2010). Thus, exercises such as the pivot press, jammer sprint or bunding, and horizontal jump, based on force vector theory, improve horizontal force production and are widely used in movement-based strength training (Moran et al., 2021; Zweifel, 2017). As a horizontally-oriented exercise, the hip thrust, has also been extensively studied in a systematic review conducted by Neto et al. (2019). It has been found that the performance on the hip thrust is associated with an increased 10 m (Loturco et al., 2018) and 20 m (Abade et al., 2019; Contreras et al., 2017) sprint performance.

In addition, Gonzalo-Skok et al. (2017) found that performing unilateral movements, such as shuffling steps, side steps or inertial flywheel crossover cutting, compared to the traditional squat, resulted in a better performance in the 10-m COD and in horizontal jump. These researchers further (Gonzalo-Skok, Tous-Fajardo, Suarez-Arrones, et al., 2017) compared bilateral (traditional squat) and unilateral (single-leg squat) training and concluded that unilateral training was more effective in improving a 180º COD and the maximum power of both legs. In addition, a lateral lunge at an inertial flywheel, compared to a half-squat, was more effective in improving performance in the 90° COD (Núñez et al., 2018). The inclusion of unilateral movements performed with accommodated resistance or resistance training, such as the rear foot elevated split squat, single-leg squat or split squat (Mausehund et al., 2019), and plyometric exercises [broad jumps, box jumps, and forward or lateral countermovement jump (CMJ)] (Bogdanis et al., 2019; Fisher & Wallin, 2014) are critical for specificity and improving performance in sports where unilateral actions predominate (Stern et al., 2020). However, longitudinal research showed no significant neuromuscular differences in the sprint (10 m, 20 m, and 40 m) and COD performance between unilateral (front step on inertial flywheel, rear foot elevated split squat or step-ups) and bilateral (half-squat on Smith machine and back squat) movements (Appleby et al., 2019; de Hoyo et al., 2015; Speirs et al., 2016).

Another aspect that has changed in the conditioning training of team sports is the intensity of the exercises (expressed as a percentage of the maximum repetition, %1RM) and the volume of the different tasks in sets and repetitions, focusing on repeated power ability (RPA) to monitor the athlete's performance (Tous-Fajardo, 2017). RPA consists of training with a load that maximises power production, i.e., the optimal load (Cormie et al., 2007). These researchers found that the optimal load for the squat corresponded to 56% (1RM) with a peak power value of slightly above 3,000 W or the power clean with 80% (1RM) and a peak power close to 5,000 W. Similarly, Loturco et al. (2013) set the optimal load in the half-squat at 65% of the 1RM load. Several studies suggest that improvements in peak power output and training at the optimal load increase maximal strength, mean power, and mean propulsive power in the back squat. Improvements in the 20-m sprint (Loturco et al., 2013), jumping performance, sprinting and agility tests (Cormie et al., 2007), repeated-sprint ability (Suarez-Arrones et al., 2014) and a reduction in fatigue resistance and intra-set power fluctuations (Gonzalo-Skok et al., 2019) have been reported. In addition, the meta-analysis conducted by Freitas et al. (2017) indicated that the combination of heavy load resistance training with plyometric or power exercises performed consecutively (complex training) causes an improvement in the ability to generate an optimal load with medium effect on sprint performance.

The main difference between traditional and movement-based training is the presence of greater specificity, individualization, and variation in the design of different tasks in terms of strength and conditioning, to meet better the demands required by different sporting actions (McGuigan et al., 2012). In this context, the theory of three-dimensional (3D) strength, Moras (2017) state that different sporting actions are generated in three axes of movement. For this reason, introducing variations in the way of performing the exercises while maintaining the basic structure causes the optimization of the performance of the sporting movement. Thus, 3D strength training produces alterations in the pattern of muscle recruitment, from different positions, during the execution of exercises such as squats and leg presses. Moreover, these alterations demand different force production and muscle activation (Da Silva et al., 2008; Escamilla et al., 2001). However, in team sports, the amount of variability of the different actions (Stergiou et al., 2006) and the uncertainty caused by a set of new or changing situations of the sport (Hossner et al., 2016) make the 3D dimension of strength insufficient. Therefore, Moras (2017) proposed including the perturbation and the four-dimensional (4D) aspect of strength.

The concept of perturbation comes from training or differential learning. Two different successive stimuli (tasks) generate information so that response, in the form of a fluctuation, is provoked, and adaptations are developed to better respond to a new situation (Schoellhorn, 2000). In team sports training, the concept of perturbation is being used to directly unbalance players because, in these sports, there is a lot of fighting and collisions between players, such as when counter-attacking or recovering the ball. Moreover, these actions occur on an irregular pitch that can be slippery if it rains or is over-watered, as in football or field hockey. Similarly, these fights between players also occur in jumping actions, where landing in struggling conditions is very common (Tous-Fajardo, 2017). Several studies have examined the use of perturbative sources on player performance. Thus, a balance master motorised force platform, a tilt board and a roller board were used to obtain better performances in the HOP test (Fitzgerald et al., 2000); different sources of perturbation, such as foam, Dyna disc or wobble board, were used to improve centres of pressure (COP) and muscle activation (Oliveira et al., 2013); a custom-built, motor-driven landing platform with different conditions (sliding, counteracting and stable) were used to reduce the risk of injury in landings (Weltin et al., 2017). The mentioned research introduced the concept of perturbation on the lower extremities, in bilateral or unilateral movements, to increase variability and to increase the task demands. Similarly, Okai & Fujiwara (2013) reported that using perturbation on upper-body postural control in a forward bilateral pushing movement under different conditions (known, unknown and unpredictable) favours inter-repetition and intra-exercise variability. The importance of the perturbation in strength training is not only the variability presented by the task but the fact that it elevates strength training to a fourth dimension (4D). As Moras (2017) theorised, this alters the pattern of muscle recruitment and the deformation of the pattern of the time series of force. The 4D force effect can be achieved by including perturbation, with devices such as fit balls, BOSU®, Wobble board, suspension traps, Pielaster® or mechanical vibration. The vibratory stimulus is perturbing and potentiating, thus enhancing the muscular activation. The 4D effect provoked by destabilizing materials has been studied in neuromuscular performance (Behm et al., 2015; Marquina et al., 2021), balance (DiStefano et al., 2009; Silva et al., 2018) and injury rehabilitation (Behm & Colado, 2012).

Likewise, the 4D effect generated by the vibratory stimulus, using vibration platform, has also been studied on neuromuscular performance (Alam et al., 2018; Osawa et al., 2013; Rehn et al., 2006; Rittweger, 2010), flexibility (Fowler et al., 2019), balance control (Ritzmann et al., 2014; Sierra-Guzmán et al., 2018) and muscle activation (Cardinale & Lim, 2003; Di Giminiani et al., 2013; Hazell et al., 2007, 2010; Marín & Cochrane, 2021; Ritzmann et al., 2013).

The 4D effect, through the inclusion of the perturbation, challenges athletes using different tasks that are proposed to them. The task's difficulty level will depend on the magnitude of instability offered by the destabilizing material and will make the task more or less challenging for the athlete (**Figure 1**). Combining the 3D and 4D effects provides strength and conditioning coaches with a wide range of functional and structural exercises with unlimited applications.

Figure 1. *Practical Application of the 4D Effect: Proposal to Increase the Difficulty of the Exercise.*

Note. (A) Bulgarian squat, (B) Front leg on BOSU, (C) Front leg on BOSU and rear leg on suspension strap cradles.

MUSCLE FUNCTION ASSESSMENT METHODS

Muscle activity

Muscle contraction, a potential action (electric current) conducted in a motor neuron, manages to innervate the muscle fibre and stimulate a neurotransmitter (acetylcholine) at the intracellular level that will cause the excitation of the sarcolemma. Consequently, the different muscle fibres part of a motor unit will contract (Hunter & Harris, 2011). During muscle contraction, the different motor units produce action potentials that can be measured using electromyography (EMG) and thus quantify the muscle activity.

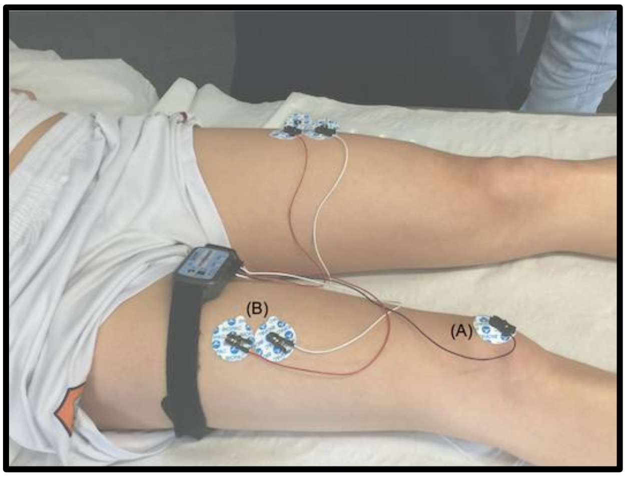
Two types of EMG can be used to record action potentials, intramuscular and surface. Intramuscular electromyography allows the recording of muscle activity from inside the muscle and thus provides local information about the muscle, whereas surface electromyography (sEMG) globally measures the activity during the contraction of the analysed muscle (Kamavuako et al., 2013). Both methods measure muscle activity by applying conductive elements. Intramuscular EMG places the electrode inside the muscle, being an invasive technique, and sEMG applies the electrode on the skin's surface in a non-invasive way (Merletti & Farina, 2009; Zajac, 1989). Comparing the two types of EMG, the intramuscular could avoid cross talk from surrounding muscles. Semciw et al. (2013) recorded gluteus medius activity in three regions (anterior, middle and posterior). These authors justified the use of intramuscular EMG to measure the activity of the anterior and middle segment of the gluteus medius without contaminating the activity record with values from the gluteus maximus and the tensor of the fascia lata, as well as measuring deeper muscles such as the posterior segment of the gluteus medius. However, Kamavuako et al. (2013) compared the activity of the forearm muscles by performing dynamic flexion/extension and pronation/supination exercises measured with intramuscular and surface electrodes, without obtaining significant differences in the recordings of both signals. Therefore, since intramuscular EMG is a more local recording of muscle activity and supposedly without cross-talk, it is difficult to agree on whether the reliability of the signal is better in the intramuscular or sEMG. However, the fact that sEMG is a non-invasive method that can be performed by non-medical personnel with minimal risk to the individual constitute the most commonly used method for recording muscle activity (Day, 2002).

Surface electromyography

Overall, sEMG has different applications in the sport context, such as the study of muscle activity in dynamic actions (provided that this dynamic action is performed in a controlled movement pattern and avoiding relative displacement between the muscle and the electrode), in the biomechanics of a gesture, in gait analysis, in studies of muscle fatigue and sports performance (Botter et al., 2009; Gabriel et al., 2006; Howard et al., 2018; Papagiannis et al., 2019). In the scientific literature, different investigations analyse different muscle groups in a given exercise or action, such as squat, deadlift, lunge or hip Thrust, to determine and compare the degree of activity of the muscles analysed, in which moments a muscle is active and to establish the level of inter-muscular coordination (Andersen et al., 2018; Contreras et al., 2016a, 2016b; Ebben et al., 2009). To apply sEMG in sports science and obtain reliable data, a methodology must be followed both in the phase before recording the muscle activity and when recording the electromyographic signal. As a concerted action funded by the European Commission (BIOMED II-Program), one of the most used protocols during the phase previous to the recording of the electromyographic signal is the Surface EMG for the Non-Invasive Assessment of Muscles Project (SENIAM Project) (Hermens et al., 2000; Merletti & Hermens, 2000). SENIAM Project determines that the distance between electrodes should be 2 cm; that the electrodes should be made of silver/silver chloride (Ag/AgCl); shave the electrode location area if it is covered with hair, clean with alcohol and let it evaporate, and check that it is dry before placing the electrode; orient the electrodes in the direction of the muscle fibres and fix the electrodes with elastic bands; place a reference electrode on the bone surface and lastly, check the electromyographic signal connection.

In addition, the SENIAM Project recommends the sensor locations for different muscle groups in the neck and shoulder, trunk and lumbar area, arm and hand, hip and upper leg, lower leg and foot. For example, SENIAM indicates that the rectus femoris electrode should be placed halfway between the anterior superior iliac spine and the top of the patella (**Figure 2**). However, the protocol established by Hermens et al. (2000) does not include the sensor location of all muscle groups, and for this reason, there are other protocols such as those of Criswell & Cram (2011). The different sensor location protocols have in common the anatomical description and explanation of the sensor placement procedure and recommendations linked to the body areas where to locate the electrode, commonly recommended on the dominant side of the body.

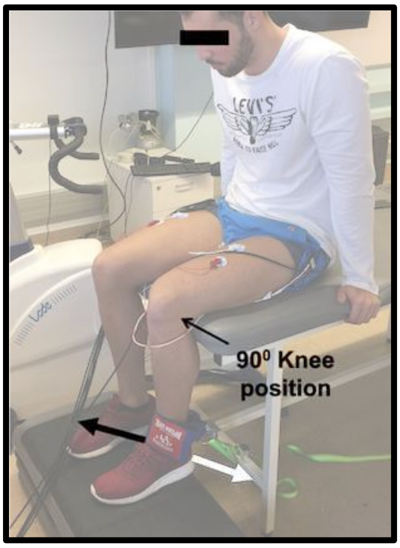
Figure 2. *Sensor Location for Rectus Femoris under the SENIAM Project Recommendations.*



Note. (A) = Reference electrode; (B) = Rectus femoris electrodes

Before starting the recording phase of the electromyographic signal, it is recommended to perform maximum voluntary isometric contraction (MVIC) tests on each muscle to be analysed to normalise the electromyographic signal. These tests imply MVIC against an invincible resistance following different protocols according to the musculature involved. In the scientific literature, there are numerous recommendations of MVIC protocols such as those of Rutherford et al. (2011) for the lower limb, Vera-Garcia et al. (2010) for the abdominal muscles or Konrad (2006) for the upper and lower limbs. For example Konrad (2006) indicates in his MVIC protocol that: for the rectus femoris, the athlete must make a unilateral extension of the knee, maintaining the knee flexion at 90º and in a sitting position (**Figure 3**). Also, athletes should become familiar with the MVIC position and be instructed in muscle recruitment to achieve correct MVIC. This technique will determine the normalization of the electromyographic signal, comparing fast movements with isometric and slow dynamic movements to achieve MVIC; isometric and slow dynamic movements are more reliable and easier to compare the electromyographic signal (Alizadehkhaiyat & Frostick, 2015). In this vein, progressive protocols to obtain MVIC are the most recommended (Jakobsen et al., 2013), such as the one increasing the muscle contraction during 2 seconds, to maintain the MVIC during 3 seconds and relax progressively, in 3 MVIC repetitions with 2 minutes rest between attempts to reduce the fatigue effect, the maximum value obtained during the 3 MVIC repetitions will be used as a reference value to normalize the electromyographic signal.

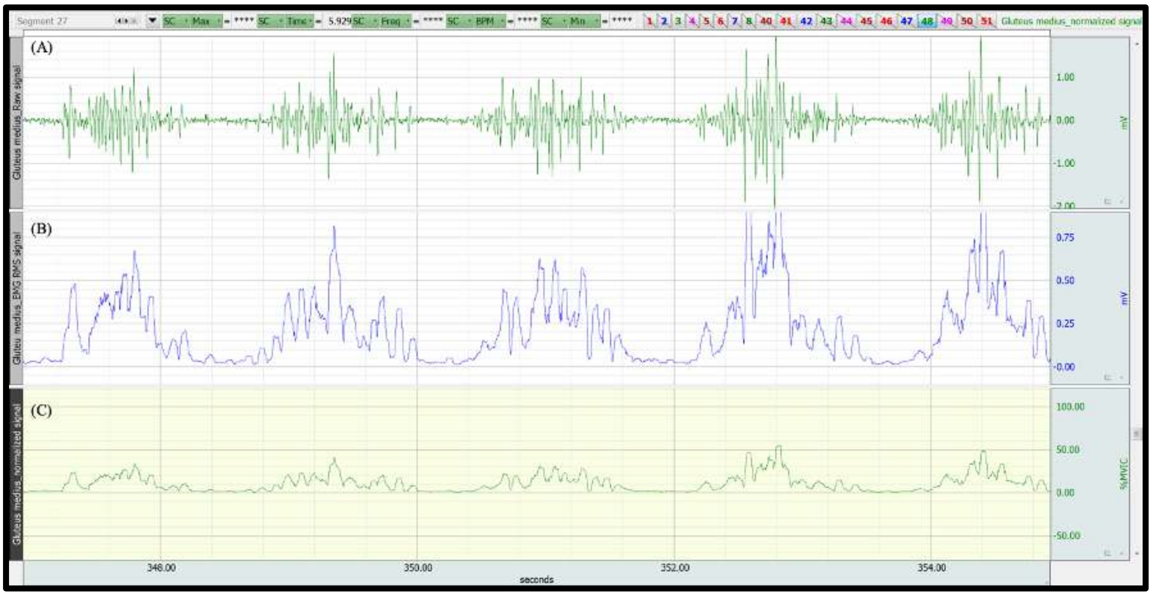
Figure 3. *Standardized Position for Achieving Rectus Femoris MVIC According to Konrad's Protocol.*



Note. The black arrow indicates movement direction, the white arrow the resistance direction.

In the recording phase, the acquisition of the electromyographic signal corresponding to the sporting action or gesture is carried out. The acquisition process is determined by the amplitude and frequency of the signal (Day, 2002). In dynamic actions, it is appropriate to analyse the amplitude of the electromyographic signal, for which different phases process the raw signal: (1) using high-pass filtering, (2) by rectifying and smoothing, or (3) by calculating the root mean square of the signal (**Figure 4**). Finally, the normalization process is applied, defined as converting the signal to scale relative to a known and repeatable value, and allows comparison between subjects and between muscles (Halaki & Ginn, 2012). Thus, the MVIC value is used to normalize the electromyographic signal by dividing the numerical values of the amplitudes resulting from the smoothing algorithm by the MVIC value, producing percentage values relative to the MVIC, expressed in percentage of MVIC (%MVIC) (Halaki & Ginn, 2012) (**Figure 4**).

Figure 4. *Gluteus Medius Electromyography Signal Acquisition during a Dynamic Bulgarian squat.*



Note. (A) = raw signal filtered using high-pass filter; (B) = signal processed by rectifying, smoothing and calculating the root mean square; (C) = signal normalized and expressed in % MVIC; EMG = electromyography; RMS = root mean square

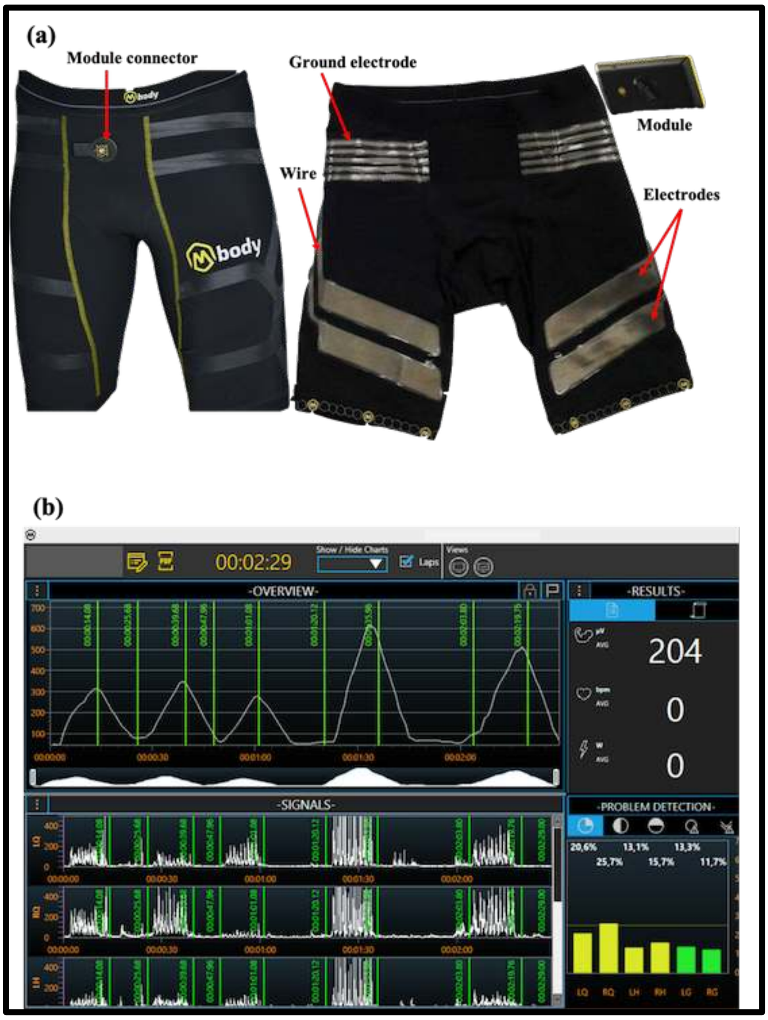
The analysis of the amplitude may vary according to the signal processing techniques and methods used to obtain standardised reference values, e.g.: (1) Maximum (peak) activation levels during maximum contractions, (2) peak mean activation levels obtained during the task under investigation, (3) activations levels during submaximal isometric contractions or (4) peak to peak amplitude of the maximum M-wave (M-max) (Halaki & Ginn, 2012). The method of the Maximum (peak) activation levels during maximum contractions to normalize is the most used, although as indicated by Burden (2010), there is no consensus as to a single "best" method for normalization of EMG data. On the other hand, for isometric actions, a frequency analysis is applied, which consists of processing the electromyographic signal, as mentioned above, and then carrying out a Fast Fourier Transform, to determine the electromyographic frequency spectrum since this does not vary over time (Kilby & Gholam Hosseini, 2004).

Wearable EMG

The main disadvantage of sEMG is the data acquisition system it uses, which is characterized by poor portability, making it difficult to transport, and by using a cable transmission system such as the Biopac MP-150 system supplied with the EMG100C electromyography module (BIOPAC System, Inc., Goleta, CA) (Sarker et al., 2017). Although wireless electromyography signal recording systems, such as BIONOMADIX wireless physiology monitoring devices (BIOPAC System, Inc., Goleta, CA), Trigno Wireless EMG system (Delsys, Natick, MA) or LE230 Wireless EMG sensor (Biometrics Ltd, Newport, UK) are commonly used in studies, their application is restricted to laboratory settings. For strength and conditioning coaches, being able to measure, control and evaluate their athletes during training or competition is more significant because the data they obtain is more in line with the demands of the sport. Therefore, recently clothing equipped with textile electrodes is being used to extract data on muscle activity.

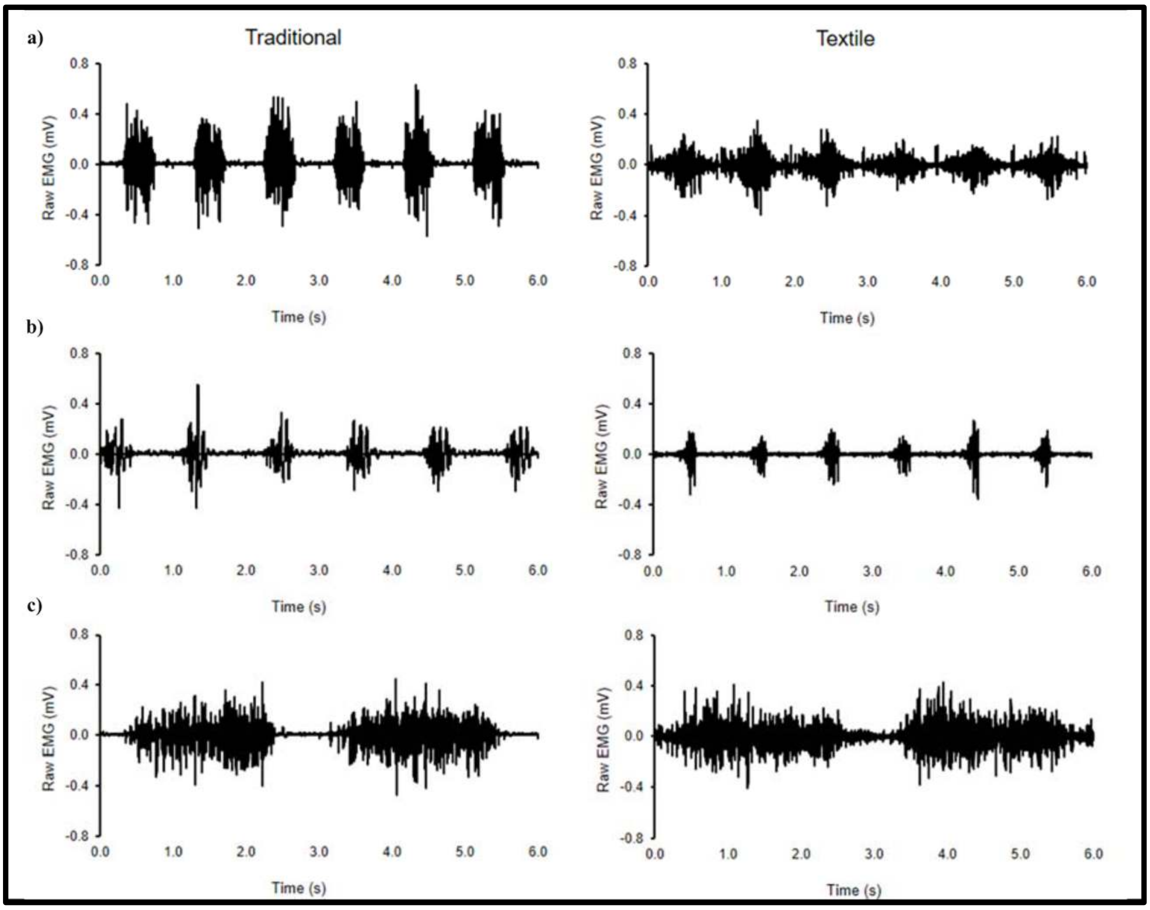
Shorts with embedded textile electrodes (Myontec Ltd., Kuopio, Finland) are the most standard garments. These shorts are manufactured using a knitted fabric similar to elastic sports clothes or functional underwear and are characterized by being equipped with conductive electrodes and wires integrated into the fabric (**Figure 5a**). The electrodes are heat-sealed onto the internal surface of the shorts and cover the gluteal, quadriceps and hamstring muscle groups, with a reference electrode located longitudinally along the side (Finni et al., 2007). The electromyographic signal is transferred from the electrodes to the electronics module; this module acquires the raw data, processes it and allows it to be downloaded to a P.C. using custom software (**Figure 5b**), where it is displayed as averaged rectified electromyography (Tikkanen et al., 2014).

Figure 5. *Shorts with Embedded Textile Electrodes (a) and Data Recorded during a Repeated Sprint Ability Protocol Expressed as Averaged Rectified Electromyography (b).*



The embedded textile electrodes have mainly been used to record muscle activity during normal daily life activities, strength training or endurance training sessions and determine the spectrum of physical activity (Finni et al., 2016; Gao et al., 2018; Marshall et al., 2020). Several studies (Tikkanen et al., 2013, 2016) have shown a close agreement between traditional sEMG and the recording of textile electrodes embedded in shorts and similar day-to-day reproducibility in different static tasks such as lying, standing or half-squatting. However, previous studies have not quantified muscle activity in more functional situations or cyclic and dynamic exercises. Research conducted by Colyer and McGuigan (2018) records the electromyographic signal of the quadriceps, hamstrings and gluteus using embedded textile shorts (Myontec Ltd, Kuopio, Finland) during functional exercises such as running, cycling and squatting in order to determine the validity and reliability of the instrument, concluding that globally embedded textile electrodes shorts can provide comparable and reproducible records of muscle activity with sEMG (**Figure 6**) in alternative environments to traditional laboratory-based methods. To the best of our knowledge, despite the functionality offered by the embedded textile shorts, in the scientific literature, there is no evidence of the use of this technology to measure muscle activity in performance parameters such as jumping ability, agility, ability to repeat sprints or to quantify the training load.

Figure 6. *Comparison between Raw Electromyography Data from Quadriceps Recorded with Traditional and Textile Electrodes during the Cycling (a), Running (b), and Squatting (c) Exercises.*

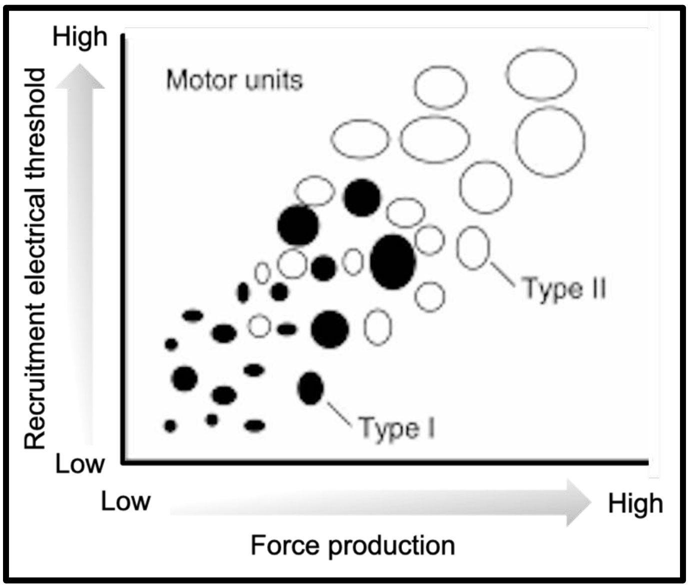


Note. From “Textile Electrodes Embedded In Clothing: A Practical Alternative To Traditional Surface Electromyography When Assessing Muscle Excitation During functional Movements,” by S. Colyer, and P. McGuigan, 2018, *Journal of Sports Science and Medicine,* 17(1), p.103

Force production

A widely known definition of force is that which comes from physics where force is the product of mass and acceleration (F = Mass x Acceleration), and consequently, a force can be considered to be the muscular effort given by a person to push or pull on an external load or resistance (Henriques, 2015). However, Bompa and Haff (2009) clarify the concept of force production and strength, establishing that strength is the capacity of the neuromuscular system to produce force against an external load. Depending on the load or resistance to be overcome, the neuromuscular system will vary the amount of force production through the frequency of activation and the number of motor units activated (recruitment) (Hunter & Harris, 2011). The motor units are made up of muscle fibres that can be slow or fast-twitch. These fibres and the level of force production will determine the recruitment of the motor units based on the size principle (Henneman et al., 1965). The size principle (**Figure 7**) states that motor units are recruited according to the magnitude of force production, i.e. small motor units composed primarily of slow fibres (Type I) will be recruited when reduced force levels are required, while larger motor units composed of fast fibres (Type IIa/Iix) may only be recruited if high force levels are required (Henneman et al., 1965). In addition, the targeting of training and the type of activity will determine the recruitment of motor units; a long-distance athlete recruits small motor units with a frequency of low threshold (Type I fibres) because the activity requires moderate, but prolonged force production and the use of motor units with Type II fibres is unlikely because the training is not in demand. On the other hand, a weightlifter who performs ballistic exercises, although the primary demand is for Type II motor units, the particularity of ballistic exercises means that the two types of fibres are combined in training following the size principle and thus recruit one type of fibre or both depending on the force production demanded by the exercise (Suchomel et al., 2018). Currently, several studies have emerged that question the size principle, particularly when comparing resistance training adaptations using high load (>65% 1 RM) and low load (<50% 1RM) (Schoenfeld et al., 2016). In this vein, Schoenfeld et al. (2017) suggest that training with high loads, with hypertrophic orientation, does not have to be the only one to maximize muscle adaptations and that low loads (<50% 1 RM) in repetitions to momentary muscle failure produce similar adaptations in muscle hypertrophy, isometric strength and muscle composition (muscle thickness). Although, training with high loads significantly increases strength gains (1RM) in both the upper and lower bodies compared to low loads (Schoenfeld et al., 2015, 2017). Besides, performing leg press with high loads achieve the high-threshold motor unit pool but not with low loads following the size principle (Schoenfeld et al., 2014). Hence, in terms of muscle recruitment, the confrontation between low and high loads indicates that resistance training intensity (% 1RM) follows the size principle.

Figure 7. *Size Principle Based on Henneman Model: Relationship between Force Magnitude Required and the Recruitment Electrical Threshold of the Motor Unit.*

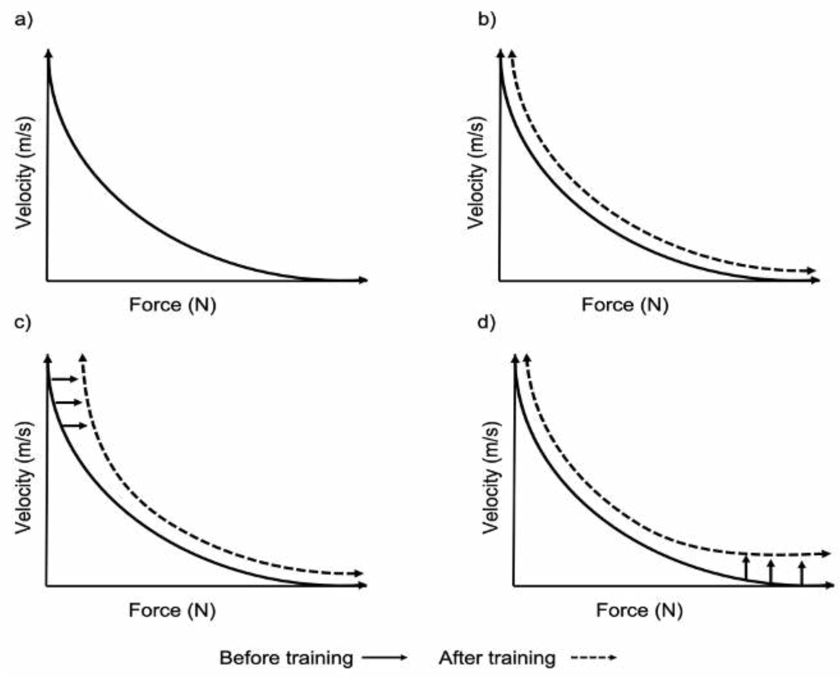


Note. Adapted from "Adaptations to Anaerobic Training Programs", by N.A. Ratamess in T. Baechle & R. Earle (Eds.), *Essentials of strength training and conditioning* (3rd ed., pp. 96–98). Human Kinetics.

Relationship between force, time, and velocity

Retrieving the equation established by Newton in the second law of motion (F = Mass x Acceleration), it is observed that to increase the acceleration of an external load, high force production is needed, and thus high speeds of motion will be achieved. On the other hand, if the external load is very high, the speed of movement decreases, establishing an inverse relationship between force and speed (Bompa & Haff, 2009), represented by the force-velocity curve (**Figure 8a**). Overall, strength training causes different adaptations in the neuromuscular system; concerning the production of force, a modification of the profile in the force-velocity curve is produced by moving the curve to the right (**Figure 8b**). When the explosive resistance training is performed, the theoretical modification to the force-velocity curve occurs in the high-velocity portion (**Figure 8c**), while if heavy resistance training is performed, the high-force portion increases (**Figure 8d**) (Harris et al., 2000; Rahmani et al., 2001).

Figure 8. *Relationship between Force Production and Velocity: a) Force-Velocity Curve, b) Alters of Force-Velocity Curve after Training, c) Theoretical Modifications of the Portion High-Velocity on Force-Velocity Curve after Explosive Resistance Training, and d) Theoretical Modifications of the Portion High-Force on Force-Velocity Curve after Resistance Training*



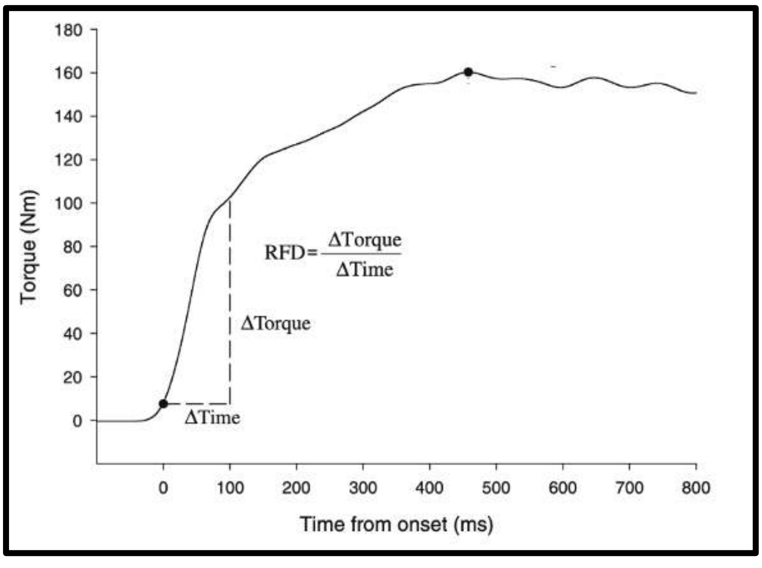
Note. Adapted from "Strength and Power Development, by T. Bompa and G. Haff, 2009, *Periodization. Theory and methodology of training*, p. 262-263

According to the training adaptations on the neuromuscular system, there is another relationship to be considered, which is the rate of force development (RFD), and it is expressed in the force-time curve. The RFD results from the ratio between the increase in force and the increase in time (**Figure 9**) and allows us to establish the speed of force production (Bompa & Haff, 2009). Therefore, it is an indicator of resistance training adaptations on explosive strength (Aagaard et al., 2002). The ability to produce a high RFD is essential for sports because it involves different explosive actions such as jumping, sprinting, or hopping, which must be performed quickly over approximately 50 ms to 250 m (Andersen & Aagaard, 2006). For this reason, the calculation of the RFD is made from the onset of the contraction to 250-300 ms in incremental periods (0-10 ms, 0-20 ms,…0-250 ms) to analyse the slope of the curve in the force-time curve (**Figure 10**) (Andersen & Aagaard, 2006).

Figure 9. *Rate of Force Development Equation*

$$\boxed{RFD = \Delta Force / \Delta Time}$$

Figure 10. *Maximum Voluntary Isometric Contraction Signal in Torque-Time Curve, where the RFD is Indicated as the Slope of the Torque-Time Curve Calculated in Intervals of 0-10, 0-20,…0-100 ms from the Onset of Contraction.*

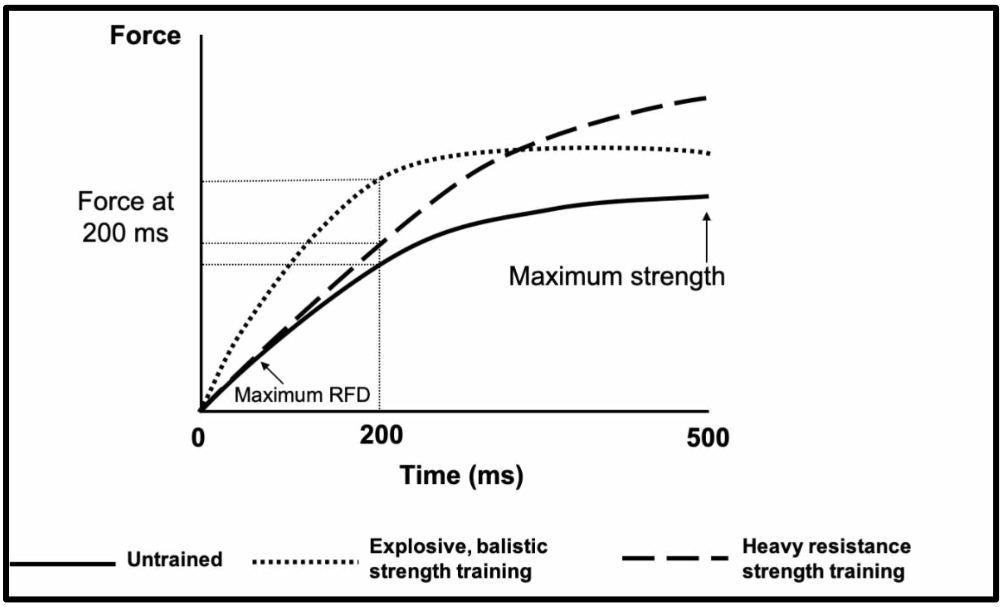


Note. From "Influence of maximal muscle strength and intrinsic muscle contractile properties on contractile rate of force development," by L. Andersen, and P. Aagaard, 2006, *Journal of Applied Physiology,* 96(1), p. 48 (https://doi.org/10.1007/s00421-005-0070-z)

Although RFD is used to indicate explosive strength, the relationship between maximal strength or muscular strength and RFD has also been studied. Suchomel et al. (2016) analysed 59 studies that established a positive correlation between maximal strength and RFD, obtaining that 97% of the studies analysed showed a significant Pearson's correlation with a moderate effect and 75% of the studies (44 studies) determined a large effect between the two variables. Although the execution times of explosive movements are substantially lower than the times for reaching maximum strength, these two variables are interrelated and linked to sports performance because they can produce acceleration and, therefore, affect the movement's speed. Furthermore, depending on the approach to resistance training, the slope of the RFD can be modified (**Figure 11**) (Häkkinen et al., 1985; Henriques, 2015).

Another concept associated with the force-time curve is the product between the production of force and the time this force is exerted, called impulse (the area under the curve), and can be a crucial factor in jumping ability or weightlifting performance (Garhammer & Gregor, 1992). In neuromuscular terms, there are different factors to consider that affect the production of force, such as the recruitment of motor units or the stretch-shortening cycle. The literature suggests that the factors influencing RFD are the rate coding and the motor unit synchronization (Suchomel et al., 2016, 2018). Thus, another critical factor for force production is a motor unit's firing rate (Hz), which EMG can detect. Conwit et al. (1999) examined the relationship between mean firing rate, mean surface-detected motor unit action potential area (motor unit size) and force production in the lower limb. As a result, these authors found that both mean firing rate and motor unit size had a positive linear trend with force production (ranged from 5% MVIC to 100%).

Figure 11. *Comparison of Slope and Determination of RFD on Force-Time Curve during Maximum Voluntary Isometric Contraction in Untrained, Heavy Resistance Strength Training, and Explosive Resistance Strength Training.*



Note. Adapted from "Changes In Isometric Force- And Relaxation-Time, Electromyographic And Muscle Fibre Characteristics Of Human Skeletal Muscle During Strength Training And Detraining," by K. Häkkinen, M. Aalén, and P.V. Komi, 1985, *Acta Physiologica Scandinavica,* 125(4), p. 581 (https://doi.org/10.1111/j.1748-1716.1985.tb07759.x)

Previously, force production and its relationship to velocity and time have been explained. In the next section, different methods for assessing force production will be explained.

Force assessment

The assessment of maximal dynamic strength is commonly evaluated from the 1 RM. As is known, the 1 RM is determined from the maximum weight that can be moved in one repetition of a specific exercise, such as the bench press or a back squat, and allows to establish strength gains. The 1 RM can be assessed directly or with submaximal loads using predictive equations like the ones from Brzycki, Eppley, or Lander (Naclerio Ayllón et al., 2009). Another method related to the estimation of 1 RM value is velocity-based endurance training. Unlike the traditional 1 RM test, this method evaluates barbell velocity with the lifted weights in exercises with a vertical displacement (Z-axis).

To determine the barbell velocity, different devices can be used, such as a) linear velocity transducers (T-Force System, Ergotech, Murcia, Spain) and linear position transducers (Chronojump-Boscosystem®, Barcelona, Spain; and Speed4Lift, Madrid, Spain), b) camera-based optoelectronic device (Velowin, DeporTeC, Murcia, Spain), c) inertial measurement units (PUSH band, PUSH Inc., Toronto, Canada; and Beast sensor, Beast Technologies S.r.l., Brescia, Italy) that determine the velocity through acceleration values and d) a smartphone application (My Lift) that establishes the velocity from the range of movement of the bar (e.g., in the bench press the distance between the chest and the bar keeping the elbows in extension) and the time of bar displacement (initial and final phase of the exercise) recording the exercise at 240 frames per second (Pérez-Castilla et al., 2019). These devices from a progressive load test (usually three loads) and a linear regression offer the estimation of the value of 1 RM indirectly, such as the predictive equation based on the mean propulsive velocity for the full back squat where *Load* (%1 RM) = - 5.961 MPV2 – 50.71 MPV + 117.0 (Sánchez-Medina et al., 2017). In addition, the above devices show force values as mean and peak, although indirectly from velocity. Likewise, some smartphone applications, such as My Lift or Iron Path, offer a tracking function that consists of superimposing a marker on the weight plate as an image that appears in the application and then records the exercise (e.g., bench press) in slow motion (240 frames per second) during the entire trajectory of the bench press to indirectly obtain data on displacement, speed, acceleration, and force.

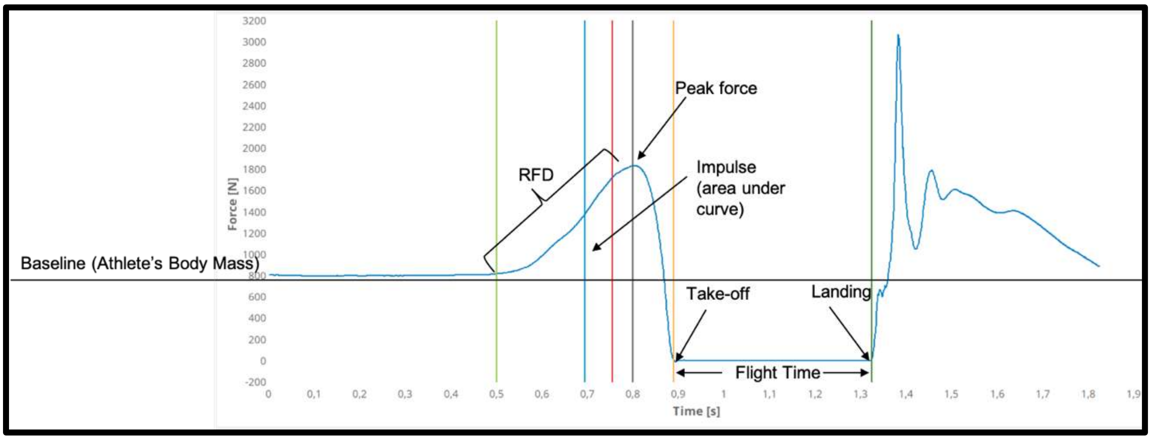
Other widely used methods for strength assessment are isometric and isokinetic dynamometry. Isometric dynamometry uses different devices, such as a handgrip or a back muscle dynamometer (Takei Scientific Instruments Co. Ltd, Niigata, Japan), to assess strength during isometric contraction. The handgrip, the gold standard to measure grip strength, is composed of an adjustable handle and records the peak force reached during isometric contraction by an electromechanical system that displays the result on a digital or analogue display (Gatt et al., 2018). To measure the isometric force with a handgrip, the participant stands with their arm fully extended, grasps the handle of the device, and squeezes the dynamometer, applying maximum force (Buscà et al., 2016; Gatt et al., 2018). Alternatively, isokinetic dynamometers such as the Biodex (BioDex Medical Inc., NY, USA) or the Cybex (Cybex International Inc., Medway, MA, USA) are motor-controlled devices that allow dynamic (concentric and eccentric) and isometric muscle strength to be assessed by offering controlled resistance at a constant speed to determine the moment of force exerted against the dynamometer (Baltzopoulos, 2008). These devices use angular speed (0º/s to 500º/s) and can be used for movements on the upper and lower extremities. The main difference between isokinetic dynamometers such as the Biodex or Cybex with other devices is that they allow working with different loads in different ranges of movement, even maximum loads, at all angles. Mainly, isokinetic dynamometry is used to assess muscle performance, injury prevention (muscle deficits and asymmetries), and determine the return to play or physical activity during a rehabilitation process (Zapparoli & Riberto, 2017).

Although the different methods mentioned above evaluate the force production, this PhD thesis evaluates the VGRF, and the forces exerted on a suspension strap. For this reason, this thesis dedicates one section to the force plate and another to the load cell.

Force plate

Movement can be described according to kinematic and kinetic characteristics. The kinetics observes the torque and the forces related to the movement, which the force plates can quantitatively measure. The evaluation of the forces is based on Newton's second and third laws. Under this theoretical principle, when an athlete applies vertical force against the ground, the ground applies a reactive force of the same magnitude. However, this force is demonized as VGRF and measured by force plates on three axes (vertical, anterior-posterior, and medial-lateral) (Beckham et al., 2014). In addition, force plates offer the analysis of other variables related to force production, such as peak force, RFD or impulse, in dynamic, isometric, bilateral or unilateral exercises (**Figure 12**).

Figure 12. *Data Acquisition of Vertical Ground Reaction Forces (Vertical Axis) and other Performance Variables while Performing a Squat Jump.*



Note. VGRF signal was recorded by a piezoelectric force plate (quartz crystal; Kistler 9260AA, Winterhur, Switzerland).

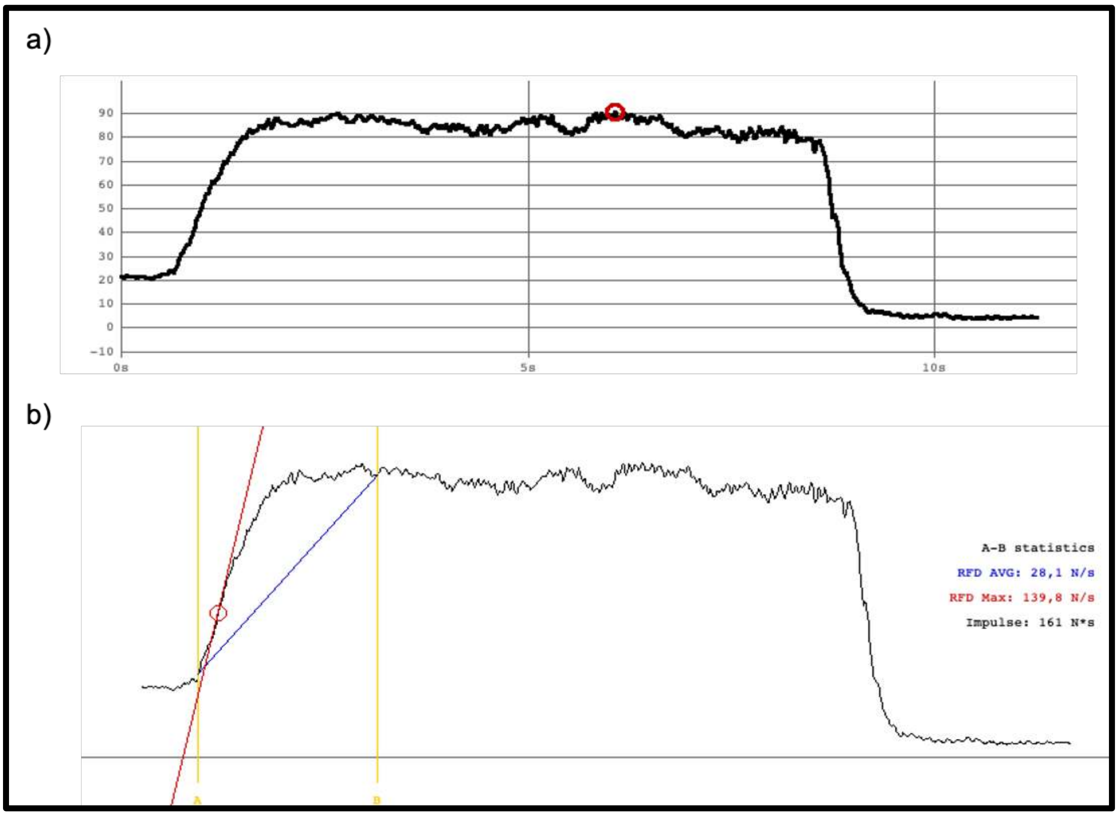
Generally, force plates measure the magnitude of the force from the changes in voltage recorded by the sensors when force is applied to the plate. These sensors are located at each corner of the force plate (4 sensors), some models of plates configure the sensors in different orientations to record both the direction and magnitude of the force (X, Y, and Z vector), the COP, the centre of force, and the moment (torque) (Scott, 2008).

Depending on the type of sensors used for VGRF acquisition, platforms with piezoelectric transducers or platforms with strain gauges and beam load transducers can be distinguished. Piezoelectric transducers measure the force magnitude using piezoelectric materials such as certain crystals, ceramics, or bone, which generate a load proportional to the force applied to the platform. On the other hand, strain gages and beam load transducers measure the force from the load that occurs when a material such as metal or another semiconductor is deformed. The main advantages of piezoelectric transducers respect strain gages and beam load transducers are a) an extremely wide measuring range, this means that piezoelectric transducers can measure minimal forces regardless of the measuring range and even with high preloads; b) protected against overloads because the piezoelectric sensor is rigid; also this sensor has c) high natural frequencies and damping in all three measurement directions, and d) the piezoelectric sensor feature does not show fatigue even after exposure to many measurements, so it provides stable functionality, reliability and accuracy over the long term (Bahra & Paros, 2010). In data acquisition, it is recommended to set the sampling frequency at 1000 Hz (at least 500 Hz) because, as stated by Lees and Lake (2008), this frequency is high enough to ensure the precision of measurement and reduction of signal aliasing while capturing the human motion.

Load cell

Strain gauge transducers used as sensors to measure VGRF on a force platform are also used in pull and compression actions to determine force production in dynamic and isometric testing or quantify weight. The most commonly used load cells or strain gauges are S-type load cells that measure the force production from the deformation on a metal material or a semiconductor (Beckham et al., 2014). The key element in this type of sensor is the deformation during traction and compression actions because the sensor measures the force based on the principle of Hooke's Law (the deformation of an elastic material is directly proportional to the force applied to it). Thus, the sensor determines the longitudinal deformation of the material on which the force is being applied (**Figure 13**). In sports science, load cells are valid and reliable instruments for measuring muscle strength (James et al., 2017; Moon et al., 2011; Steeves et al., 2019). Furthermore, load cells have been widely used in several studies to measure isometric strength (Anwer et al., 2011; Anwer & Alghadir, 2014; Bartolomei et al., 2015; Fortier et al., 2013; Khan et al., 2013; Mata et al., 2015; Rossi et al., 2014).

Figure 13. *Force-Time Curve Recording with S-Type Load Cell during an Isometric Pull: a) Raw Data and b) Smoothed Curve with RFD and Impulse Variables.*



Note. Force was recorded by an S-type load cell Chronojump-Boscosystem (Barcelona, Spain).

PERTURBATING STRENGTH AND CONDITIONING TASKS

Unstable surfaces

Background and performance assessment

The inclusion of unstable surfaces as a method of strength training arises from the functional exercises characterized by using body weight as a load, for its specificity and transference with the sports actions and for being a low-cost way of training efficiently (Thompson, 2016). Thus, unstable resistance training has progressively gained prominence over other traditional training methods. Likewise, coaches, athletes are continuously searching for new challenges to increase training demands through the complexity of the exercises, for instance, by modifying the amount of instability or intensity (Behm et al., 2015). Thus, the use of devices that create instability has become popular (i.e., BOSU® Ball, Wobble Board®). Primarily, unstable devices increase the load of traditional exercises by providing higher muscular demands through superior motor unit recruitment. Such devices also improve neuromuscular coordination to maintain balance during training exercises (Snarr et al., 2016). Hence, destabilizing environments provide more varied and effective training stimuli, enhancing neuromuscular adaptations (Kibele & Behm, 2009). Thus, the strength gains comparison between unstable and traditional resistance training has been investigated. Sparkes and Behm (2010) reported that after eight weeks of training, both the traditional training group and the functional training group (exercises on stability ball and stability disk) significantly increased jump capacity (CMJ) and strength (3RM) in back squat and bench press compared to the baseline, although non-significant differences were obtained between groups. The researchers concluded that training with unstable surfaces increased strength in untrained people with their bodyweight as a load. In addition, Kibele and Behm (2009) obtained that seven weeks of functional training using BOSUs and stability ball significantly increased the strength (1RM) in back squat in both the unstable and traditional training groups.

Similarly, Marinkovic et al. (2012) observed that eight weeks of training using BOSUs and stability balls significantly increased force gains (1RM) in the bench press and back squat in both groups (traditional and functional). Besides, Behm et al. (2015) conducted a systematic review with a meta-analysis reporting that unstable training provokes similar effects on muscle strength, power, and balance performance when compared to traditional resistance training. Recently, Saeterbakken et al. (2019) evaluated the short-term effects of three weeks of performing resistance training squats under different conditions (Smith Machine, free-weight and Wobble Boards), reporting that the Wobble Board group significantly increased 27.3% the strength gains (1 RM) in squat and also improved the jumping ability (CMJ height) in 8.5% compared to the other conditions. These authors also evaluated the long-term effects (7 weeks) of squatting on a smith machine, with free weight and on a Wobble Board. lthough CMJ performance was significantly higher for Wobble Board after seven weeks of training than the previous three weeks, non-significant differences were found between groups in strength gains and jump ability.

The research above indicates that training on unstable surfaces elicits the same adaptations in neuromuscular performance (1RM and CMJ) as traditional resistance training in an untrained or physically active population.

Regarding muscle activation, some evidence supports the idea that instability training elicits higher activity of several upper body and trunk muscles than traditional exercises such as push-ups, crunches, sit-ups, and back extensions. Anderson et al. (2013) recruited highly trained individuals to examine triceps brachii, erector spinae, rectus abdominis, internal oblique and soleus activation while performing traditional and unstable push-ups with hands or feet on the unstable surface. These researchers found that push-ups on the unstable surface significantly increased the activation of the triceps brachii by 34%, erector spinae by 50%, rectus abdominis by 61%, internal oblique by 61%, and soleus by 86% compared to the stable condition. Regarding abdominal muscles, different exercises performed on a stability ball (pike, knee-up, skier, roll-out) were analysed by Escamilla et al. (2010) in a sample of healthy young females to determine activation levels of rectus abdominis, internal and external oblique compared to traditional exercises (crunch and bent-knee sit-up). The roll-out obtained a very high activation (>60% MVIC) in the upper rectus abdominis, the roll-out and the pike a high activation (41-60% MVIC) in the lower rectus abdominis, the pike, the knee-up and the skier a very high activation (>60% MVIC) in the external oblique and a high activation (41-60% MVIC) in the internal oblique compared to the moderate activations (21-40% MVIC) of the traditional exercises. In particular, the lower rectus abdominis, the internal and external oblique activations in the pike exercise were significantly greater than in the crunch and bent-knee sit-up exercises. Likewise, Cosio-Lima et al.'s study (2003) showed that after five weeks of sit-up and back extension unstable training (stability ball) in untrained college women, muscle activity of rectus abdominis and erector spinae significantly increased (67% and 63%, respectively) compared to the control group.

For the studies mentioned previously, the effect of unstable surfaces on the activation of the abdominal musculature in an analytical exercise such as a pike or sit-up is evident. However, including unstable surfaces in upper or lower body exercises can provoke distinct effects on the agonist, antagonist, or stabiliser musculature. For instance, in Anderson et al.'s (2013) research, push-ups performed in instability elicited a higher effect on the stabilising (abdominal) musculature than on the triceps brachii (synergist). Likewise, Torres et al. (2017) found a significantly higher activation of the stabilising musculature of the scapula (anterior serratus, upper, middle and lower trapezius) in push-ups under unstable than stable conditions. Furthermore, these authors found that the inclusion of unstable surfaces in push-ups inhibited the synergist (anterior deltoid) and antagonist (posterior deltoid) musculature compared to stable surfaces. However, the activation of the agonist musculature (pectoralis major) was similar between conditions; the triceps brachii (synergist) under unstable conditions registered the highest activation value among the other muscle. Accordingly, the inclusion of the unstable surfaces and their effects on the muscle groups should be considered. On the other hand, it could be interesting to evaluate the effect of the unstable surfaces in terms of the activation value (%MVIC) rather than the role of the muscle groups in the different exercises.

Apart from the upper-body and trunk musculature, the effects of unstable surfaces on lower extremity muscle recruitment have also been examined. Isometric squat exercise under stable and unstable conditions (balance disks) was investigated by McBride et al. (2006), who reported significantly greater activation of the vastus medialis and lateralis in stable squat compared to unstable squat (stable vs unstable: 182.35±63.28 mV vs 119.56±54.53 mV; 206.72±66.56 mV vs 129.65±53.83 mV, respectively), although the activation of biceps femoris in unstable squat condition had an increase of 32% was not significant,; also there were no significant differences between conditions and activation of medial gastrocnemius. Another research conducted by McBride et al. (2010) examined the muscle activity of vastus lateralis, biceps femoris and erector spinae while performing dynamic squats performed with absolute and relative loads under different conditions. These authors reported that stable squat provides a significantly greater strength gain (1 RM) than unstable squat (stable vs unstable: 128.0±31.4 Kg vs 83.8±17.3 Kg), and the muscle activity was significantly higher for vastus lateralis in both relative (70%, 80%, and 90% 1 RM) and absolute (59 Kg, 67 Kg, and 75 Kg) loads under stable than unstable condition, for biceps femoris the relative load (90% 1 RM) was also significantly higher under stable squat condition. Moreover, there were no significant differences in terms of activity for erector spinae between exercise condition and load lifted (absolute or relative). Likewise, Harput et al. (2014) studied muscle activation of the vastus medialis, lateralis, medial hamstring, and biceps femoris in a group of healthy subjects to determine the effects of gender and quadriceps:hamstring (Q:H) ratio when performing forward-lunge, single-leg-stance, side-lunge, and single-leg-squat exercises on a Wobble Board. The analysed muscles obtained a moderate to low activation (<40% MVIC), except for the vastus lateralis in the side lunge exercise that achieved a very high activation (>60% MVIC), but the exercises performed on the unstable surface were not compared with traditional exercises. However, the Q:H ratio was significantly higher for women than for men, and the Q:H ratio was significantly lower in single-leg-stance exercises than in the other exercises in both genders. Another study conducted by Youdas et al. (2007) found that surface (stable vs unstable) and sex have a significant effect on the activations of rectus femoris (women vs men in a stable surface: 33.9% MVIC vs. 20.1% MVIC, respectively; $p = 0.04$) and hamstring (men vs women in an unstable surface: 37.9% MVIC vs 19.9% MVIC, respectively; $p = 0.04$) during the extension of a standard lunge in healthy recreational athletes. On the other hand, Andersen et al. (2014) examined the effect of performing a standardized Bulgarian squat (6-RM loaded) under stable (front leg on the floor) and unstable (front leg on a foam cushion) conditions on the hip and thigh muscles of healthy trained participants. Bulgarian squats significantly increased the activation of biceps femoris under stable conditions compared to those under unstable conditions (stable vs unstable: 215.5 ± 106.7% MVIC vs 193.3 101.5% MVIC, $p = 0.030$), and there were no significant differences for rectus femoris, vastus medialis, vastus lateralis, and gastrocnemius, and all of them achieved a high activation (>60% MVIC) under both exercise conditions. Recently, Monajati et al. (2019) investigated the effect of different squat conditions (single-leg on a bench, double-leg on the floor, and double-leg on BOSU (dome side up)) on quadriceps and hamstring muscle activation in a sample of female elite soccer players. The single-leg squat condition significantly increased the activation of biceps femoris, vastus lateralis, and vastus medialis compared to double-leg squat (biceps femoris: 5% MVIC vs 15 % MVIC, $p = 0.046$; vastus lateralis: 60% MVIC vs 80% MVIC, $p = 0.040$; vastus medialis: 65% MVIC vs 100% MVIC, $p = 0.021$) but not for double-leg squat on BOSU. Despite, the single-leg squat significantly increased the activity of semitendinosus compared to double-leg squat (8% MVIC vs 15% MVIC, $p = 0.040$) and double-leg squat on BOSU (5% MVIC vs 15% MVIC, $p = 0.010$).

The effects of unstable surfaces on force production have been examined for lower body exercises. Previous studies have shown that an unstable environment decreases force output (Behm et al., 2002). McBride et al. (2006) recorded that when an isometric squat is performed on balance disks, the peak force and RFD decreased significantly by 45.6% and 40.5%, respectively, compared to stable squat (on the floor). Likewise, Saeterbakken & Fimland (2013) examined the isometric force output while performing squat under stable (on the floor) and unstable conditions (power board, balance cone and BOSU). As a result, force output significantly decreased under BOSU and balance cone conditions compared to squat on the floor (Floor vs BOSU: 749±222 N vs 603±208 N, $p = 0.003$; Floor vs Balance cone: 749±222 N vs 570±257 N, $p = 0.001$) and power board (Power Board vs BOSU: 694±220 N vs 603±208 N, $p = 0.037$; Power Board vs Balance cone: 694±220 N vs 570±257 N, $p = 0.001$). Moreover, with respect to the floor, squatting on BOSU and balance board decreased the force output by 19% and 24%, respectively. Compared to a power board, the force output decreased 13% while performing a squat on BOSU and 18% for a balance cone. In this vein, another investigation reported that BOSU® and T-Bow® deadlift conditions significantly decreased force production in deadlift on the floor (Floor vs BOSU: -34.19% $p < 0.005$; Floor vs T-Bow: -8.80% $p = <0.05$) (Chulvi-Medrano et al., 2010). It seems that the studies above show a consensus on the effects of unstable surfaces on the upper body and trunk muscle activity and force production. However, evidence that unstable surfaces increase muscular demands during lower body exercises, such as squat or Bulgarian squats, is weak.

Suspension training

Background and performance assessment

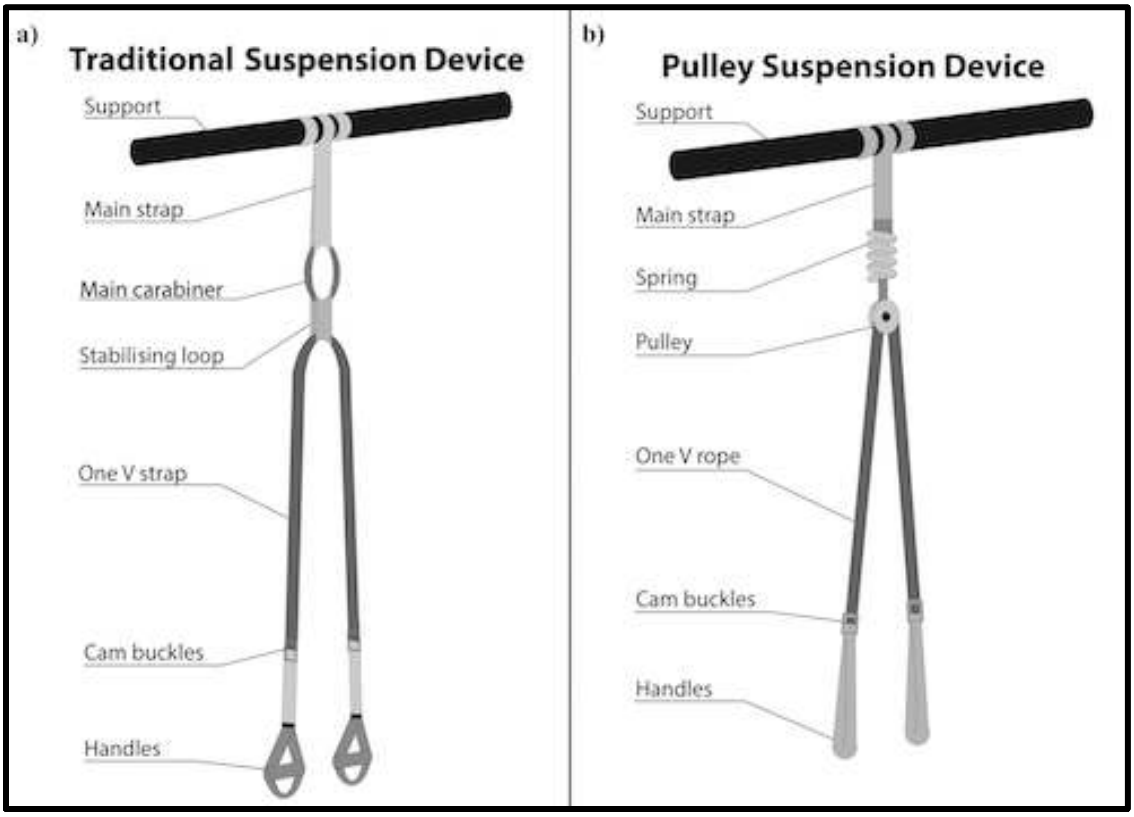
One of the most popular materials for training anywhere, performing a wide range of whole-body exercises in multiple planes with bodyweight as a load are suspension devices, which are also an essential element of functional training equipment (Boyle, 2017). These devices can be used to perform different exercises to strengthen the whole body, and thus simultaneously train multiple muscle groups from multi-directional movements and using body weight as resistance to overcome (Harris et al., 2017; Hetrick, 2006). The instability that suspension training offers is because the device is based on a system of straps with handles on the bottom and attached to a single anchor point that acts as a pendulum by rotating around the singular anchor point (Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014). In addition, to increase the recruitment of motor units, suspension devices are based on the fundamental principles of suspension training (Bettendorf, 2010). The first principle is the resistance vector, which means that the angle of body inclination is used to modify the resistance (the exercise is easier when the person's body is slightly inclined). The second principle is stability, which states that concerning the base of support and the balance, the difficulty level of an exercise will be lower when the body's contact points with the ground are increased, and the amplitude concerning the device will be reduced. Finally, the third principle is the pendulum, which established that the device acts in stable equilibrium in front of the different perturbations on the stability line (perpendicular to the ground).

The literature shows some evidence about the effects of a training program with both suspended exercises and unstable surfaces on jumping performance and muscle strength. Tomljanovic et al. (2011) compared the performance of a traditional training group with a functional training group (suspension device, stability ball, flowing, power-wheel), observing that the functional training group improved their postural control and coordination for the standing overarm medicine ball throw and the agility test Hexagon, while the traditional training group improved their ground contact time and peak power for the explosive jumping strength (CMJ). Although, non-significant differences were found for the jumping height performance (CMJ) between training groups. Likewise, Maté-Muñoz et al. (2014) used a suspension device and a BOSU® to perform different exercises of the functional training group for seven weeks. By previous studies, jumping performance (squat jump (SJ) and CMJ) increased in both groups but did not differ significantly between groups. Both training programs effectively improved muscular strength (1RM), peak power and average power in the back squat and bench press. In agreement with the previous studies, it seems that training programs combining suspension devices with other unstable surfaces do not significantly increase performance parameters compared to traditional training, even though the suspension device could produce a higher degree of instability or level of perturbation compared to other unstable materials. Thus, Janot et al. (2013) and Dannelly et al. (2011) studied the effects of a suspension training program compared to a traditional training program, for seven and 13 weeks, respectively. Both studies agree on the improvement in strength gain (leg press, back squat, and press bench) achieved by both groups.

In contrast, a 6-week intervention in young high-standard handball players performing glenohumeral joint rotation exercises with a suspension device significantly improved the muscle strength of the internal and external rotators compared to the traditional training group (Genevois et al., 2014). Furthermore, a group of elite San Shou athletes (Chinese boxing speciality) carried out a 10-week training program with suspension devices to strengthen the trunk muscles. As a result, the explosive power of the trunk extensors and flexors in the suspension training group significantly increased compared to the traditional training group (Ma et al., 2017).

The degree of instability of the suspension devices and its effect on muscle activation of the upper body have been analysed in some traditional exercises such as push-ups, inverted row, and prone bridge. The use of a suspension device in the push-up exercise increased the activation of most of the muscles involved in this exercise (pectoralis major, anterior deltoid, upper trapezius, triceps brachii, latissimus dorsi, and serratus anterior) in comparison with a push-up on the floor (Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015; Borreani, Calatayud, Colado, Tella, et al., 2015; Snarr & Esco, 2013b). Some muscles such as the anterior deltoid or anterior serratus showed inhibition in recruitment due to lateral instability caused by the pulley suspension device (**Figure 14**) compared to the activation recorded during push-ups on the floor (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014). Conversely, the increased instability caused by the pulley suspension device resulted in increased recruitment of the upper trapezius, triceps brachii, and posterior deltoid compared to push-ups on the floor and suspended push-ups with a traditional suspension device (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014).

Figure 14. *Suspension Training Devices and their Main Features: a) Traditional Suspension Device and b) Pulley Suspension Device.*



Regarding the inverted row, Snarr and Esco (2013a) found no significant differences in muscle activity of latissimus dorsi, upper trapezius, and posterior deltoid when comparing the suspended condition with the traditional one. However, the biceps brachii significantly increased the activation during the inverted row. Likewise, McGill et al. (2014b) obtained no difference in the activation of latissimus dorsi in both traditional and suspended inverted row conditions. Contrarily, Snarr et al. (2014) revealed that variations in suspended inverted row grip (neutral, pronated, supinated position) resulted in increased latissimus dorsi, posterior deltoid, and biceps brachii except for the middle trapezius. Another back-strengthening exercise that has been studied by Snarr et al. (2017) is the suspended pull-ups. These researchers compared the traditional pull-ups with the suspended and towel conditions in the primary muscles involved in this exercise (latissimus dorsi, biceps brachii, middle trapezius, and posterior deltoid), reporting high (41-60% MVIC) and very high (>60% MVIC) activations for the muscles analysed without being significant among pull-up conditions. Concerning the suspended prone bridge exercise, the activation of the trunk muscles (rectus abdominis, external oblique, and erector spinae) was significantly higher than the traditional exercise. Moreover, the prone bridge condition with the forearms in suspension was the most demanding with the activity of the rectus abdominis (Atkins et al., 2015; Byrne et al., 2014; Snarr & Esco, 2014).

The effects of the suspension device on the activation of the core muscles have been widely studied. In push-ups, rectus abdominis and external oblique showed values above 32% MVIC and 26% MVIC in suspended push-ups compared to push-ups on the floor (Beach et al., 2008; Fong et al., 2015; McGill et al., 2014a; Mok et al., 2014), and very high activation (>60% MVIC) when push-ups were performed with a pulley suspended push-up (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014). However, for the suspended inverted row, McGill et al. (2014b) found no significant differences in the activation of rectus abdominis, external and internal oblique compared to inverted row. Although, for the suspended prone bridge, Fong et al. (2015) and Mok et al. (2014) recorded moderate (21%-40% MVIC) to very high (>60% MVIC) activation in the rectus abdominis and external oblique, these researchers did not compare the activation performing the exercise under stable conditions. Another exercise that has been studied under suspended conditions is the bilateral and unilateral supine bridge (Calatayud, Casaña, Martín, Jakobsen, Colado, Gargallo, et al., 2017). These researchers reported that rectus abdominis and external oblique recorded higher muscle activity in the pulley suspended supine bridge, but these values were not significantly different from the activation in the stable exercise (on the floor). However, activation of the erector spinae and lumbar multifidus was significantly greater in the unilateral pulley suspended supine bridge than in other exercise conditions. In line with the previous research, the pulley suspended prone bridge and variations (roll out and unilateral prone bridge) significantly increased the activation of the rectus abdominis, the external oblique, the erector spinae and the lumbar multifidus in comparison with its counterparts (Calatayud, Casaña, Martín, Jakobsen, Colado, & Andersen, 2017). Furthermore, other researchers (Cugliari & Boccia, 2017) have reported that dynamic suspended exercises, to strengthen the trunk stabilizing muscles such as roll-out and body-saw, significantly increased rectus abdominis and external oblique activity (>60% MVIC) compared to pike and knee tuck. In addition, Snarr et al. (2016) compared the pike exercise in different conditions (floor, BOSU dome side down, stability ball, core coaster, and suspension). They found that the suspended pike significantly increased the rectus abdominis activity by 46%, the external oblique by 68%, and the erector spinae by 52% compared to the stable pike (on the floor) condition.

The effects of using a suspension device on lower body activation have been investigated primarily in upper body exercises such as rectus femoris in push-ups (Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014;

Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014), prone bridge (Byrne et al., 2014) and pike (Snarr et al., 2016), or the gluteus maximus in push-ups (Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Harris et al., 2017), inverted row, and prone and supine bridge (Harris et al., 2017). However, some research has studied the effect of suspension devices on the activation of prime movers in lower body exercises such as hamstring curl, lunge or Bulgarian squat. Thus, Malliaropoulos et al. (2015) examined the effect of ten hamstring loading exercises (standard lunge, single-leg Romanian deadlift T-drop, kettlebell swing, bridge, suspended hamstring curl, hamstring bridge, curl, Nordic exercise, stability ball flexion and slide leg exercise) on biceps femoris and semitendinosus activity in elite female track and field athletes reporting a very high activation in biceps femoris and semitendinosus (>60% MVIC) in the suspended hamstring curls compared to the high-to-low activity (<60% MVIC) for the standard lunge, single-leg Romanian deadlift T-drop, kettlebell swing, bridge, hamstring bridge, curl, and Nordic exercise. However, the suspended hamstring curl was less demanding for the biceps femoris (84% MVIC) and semitendinosus, (75% MVIC) than the stability ball flexion and the slide leg exercise, both with muscle activity >90% MVIC. Recently, Krause et al. (2018) assessed the activation of hip and thigh muscles during a suspended lunge (rear leg leaning on the suspension device cradles) and its counterpart. The suspended lunge exercise significantly increased hamstrings, gluteus maximus, gluteus medius and adductor longus activation between 34% and 42% compared to standard lunge. However, the authors did not find significant differences in the rectus femoris because the suspension condition was not enough demanding compared to the standard lunge with a 10% increase in the rectus femoris activity. Following the previous study, Miller et al. (2019) examined rectus femoris and gluteus maximus in Bulgarian squat and suspended lunge (rear leg leaning on a bench and rear leg leaning on the suspension device cradles, respectively), obtaining that the average of muscle activity (millivolts (mV)) for rectus femoris and gluteus maximus was higher but non-significant for the suspended lunge than its counterpart. However, concerning the Bulgarian squat, the gluteus maximus significantly increased by 14.19% the average of muscle activity compared to the rectus femoris (2.78%) under the suspended condition.

Other studies have assessed the VGRF in upper body suspended exercises (McGill et al., 2014a, 2014b), such as push-ups and inverted rows, to examine the effects of trunk-leg inclination on force production. Likewise, Melrose and Dawes (2015) measured the force exerted on the suspension strap while performing an isometric suspended inverted row in college students. These authors found that the percentage of body mass resistance on the suspension strap increases from 37.4% to 79.4% when the trunk-leg inclination is closer to the floor (from 30º to 75º). In this vein, Cayot et al. (2017) assessed the force exerted on the suspension strap in the biceps curl exercise, registering lower dynamic and isometric force production (35% and 31%, respectively) for the suspended biceps curl compared to the biceps curl with barbell. Furthermore, Gulmez (2017) recruited male sport sciences students to examine the force on the suspension strap and VGRF while performing isometric suspended push-ups under two conditions (elbow flexion and elbow extension). The study found that when trunk-leg inclination is modified (from 45º to 0º), the percentage of body mass resistance increases (elbow flexion: 36.8% to 75.3%; elbow extension: 11.9% to 50.4%), while VGRF decreases (elbow flexion: 80.7% to 32.2%; elbow extension: 97.5% to 46.6%). According to the previous study, Giancotti et al. (2018) modified the length of the suspension straps (from 178 cm to 238 cm) and observed that when the length of the suspension strap was longer, and therefore the trunk-leg inclination increased, the percentage of body mass resistance increases (elbow flexion: 41.5% to 58.2%; elbow extension: 19.1% to 43.5%) while VGRF decreases (elbow flexion: 58.5% to 41.8%; elbow extension: 80.9% to 56.5%).

The literature shows that the effects of suspension devices have been widely studied in traditional exercises such as push-ups, inverted rows, prone bridge, or supine bridge. The additional instability provided by a suspension device mainly has been assessed in the prime movers and stabilizing muscles activity on the upper body. Likewise, force production under suspended conditions also has been measured in upper-body exercises like push-ups and inverted rows. Nevertheless, the evidence about the effects of suspension devices on muscle activation and force production for the prime movers in lower-body exercises (squats, hamstring curls, Bulgarian squats, lunges, single-leg squat) are insufficient.

Whole-body vibration

Background and performance assessment

Other devices such as whole-body vibration (WBV) platforms are commonly used to increase neuromuscular performance in strength training. These platforms modify workloads through vibration (side-alternating vibration or synchronous vibration), frequency (in Hz), and amplitude (peak to peak displacement, in mm) and, as a consequence, the magnitude of acceleration following the muscle tuning paradigm (Cardinale & Bosco, 2003; Cardinale & Wakeling, 2005). WBV is applied to the muscle or tendon to elicit tonic vibration reflex (Issurin, 2005), and the beneficial effects of WBV on muscle strength and jump ability have been demonstrated in lower limb exercises (squat, half-squat, Bulgarian squat, or lunge) in different cohorts such as untrained, recreationally active, and older adults (Osawa et al., 2013; Rehn et al., 2006; Rittweger, 2010). However, the effect of vibration training on dynamic exercises with heavy loads is not completely clear. Thus, Rønnestad (2004) found that five weeks using WBV (40 Hz) did not improve maximal strength or jumping ability compared to a traditional dynamic squat (6-10RM). Recently, Hammer et al. (2018) found that WBV training (50 Hz at < 1 mm of amplitude) combined with dynamic squat resistance training (85%-95% 1RM) did not elicit higher performance in maximal strength and standing broad jump compared to squat resistance training. In both studies, the participants were healthy, recreationally resistance-trained men. In contrast, Bush et al. (2015) reported a post-activation potentiation effect on knee extension torque after exposing healthy participants to a WBV dynamic squat with bodyweight resistance (30 Hz and 4 mm of amplitude).

As for muscle activation, vastus lateralis recruitment significantly increases when performing 60 s of static half-squat with 100º of knee flexion at three different WBV frequencies (30, 40, and 50 Hz) with 10 mm of amplitude compared to half-squat with no vibration (Cardinale & Lim, 2003). Likewise, Di Giminiani (2013) reported that performing 20 s of static half-squat in four different positions (knee flexion angle ranging from 90° to 120°) with WBV (45-55 Hz and 1 mm of amplitude) increased the activation of vastus lateralis compared to a half-squat with no vibration applied in male sport sciences students. Moreover, Ritzmann et al. (2013) found that a progressive increase in WBV frequencies (from 5 to 30 Hz) and amplitudes (from 2 to 4 mm) causes a progressive increase in the activation of vastus medialis, rectus femoris, and biceps femoris while performing 10 s of static half-squat. Thus, frequencies ranging from 30 to 55 Hz and amplitudes from 2 to 5 mm elicited the highest response in the muscles mentioned above (Hazell et al., 2010; Osawa et al., 2013; Ritzmann et al., 2013). Recently, Marín & Cochrane (2021) found a significantly greater biceps femoris and semitendinosus activation performing 30 s static supine bridge in WBV at 30 Hz and 40 Hz than non-vibration, but for gluteus maximus and multifidus lumbar, authors did not find significant differences among conditions.

The evidence of the effects of WBV on muscle activation in lower body exercises suggests that it is an excellent method to enhance muscle recruitment. However, literature shows that most of the exercises exposed to WBV have been isometric, even in randomized control trials such as those by Osawa & Oguma (2013). This study established a cadence of 4 s for the concentric, 2 s for isometric (lower position), and 4 s for the eccentric phase in leg and trunk exercises (squat, Bulgarian squat, trunk curl, leg raise, and back extension). It seems that the effects of WBV on neuromuscular performance variables and muscle activity in dynamic lower-body exercises have not been widely investigated.

Dual Condition

Background and performance assessment

Vibratory platforms, flywheels, rubber bands, or pulley machines have been used together with other devices such as Pielaster®, stability Balls, Freeman plates, and BOSU® to create instability because some evidence suggested that the combination of two training methods (dual condition) or two unstable surfaces (dual instability) enhance the muscle activation, mainly the synergistic and stabilising musculature. As well as have positive effects on balance, postural control, and intermuscular coordination. Thus, Norwood et al. (2007) compared the muscle activation of latissimus dorsi, rectus abdominis, internal oblique, erector spinae, soleus, and biceps femoris in bench press exercise under conditions of stability, single instability (upper body on a stability ball or lower body on BOSU dome side down), and dual instability (both upper and lower body on a stability ball and BOSU, respectively). Results showed a significant linear effect between the amount of instability provided (stable, single, and dual instability) and the level of muscle activity in latissimus dorsi, internal oblique, erector spinae, and biceps femoris, but not the rectus abdominis and the soleus. Besides, the dual instability improved the activity of rectus abdominis (14%), internal oblique (58%) and erector spinae (90%) in comparison with stable condition, and the activity achieved under the dual instability for latissimus dorsi, soleus and biceps femoris was significantly higher than the stable condition, with an increase of 64%, 84% and 90%, respectively. Following the previous study, Anderson et al. (2013) recruited highly trained individuals to examine triceps brachii, erector spinae, rectus abdominis, internal oblique and soleus activation while performing traditional and unstable push-ups in the single (hands on extreme balance board or feet on stability ball) or dual (hands and feet on extreme balance board and stability ball, respectively) instability. The authors showed increased muscle activity as the percentage of change for single and dual instability compared to the stable condition. For dual instability, all the analysed muscles reached the highest increase of percentage of change (>150%) compared to the other conditions and also, under the dual instability recorded significantly higher activity when comparing the activity of the push-ups on the floor for triceps brachii (stable vs dual instability: 57.48±23.51 mV vs 162.14±61.03 mV), erector spinae (stable vs dual instability: 11.80±5.94 mV vs 31.79±16.78 mV), rectus abdominis (stable vs dual instability: 27.37±16.77 mV vs 100.29±88.14 mV), internal oblique (stable vs dual instability: 38.56±23.64 mV vs 151.10±79.54 mV) and soleus (stable vs dual instability: 9.03±15.97 mV vs 29.23±34.73 mV). Besides, a significant linear effect was found between the amount of instability provided and the level of muscle activity in all muscles and exercise conditions. On the other hand, Freeman et al. (2006) examined the effects of dual instability in the upper body (two unstable surfaces in the same body region) on push-ups, obtaining that the push-ups on basketball balls (two hands on two balls) caused greater recruitment of the prime movers (pectoralis major and biceps brachii) and the stabilizers (rectus abdominis, external oblique, internal oblique, and latissimus dorsi,) with increases in activation over 24% and 34% respectively, compared to push-ups under stable conditions. In this vein, Byrne et al. (2014) examined the effects of dual instability on the core muscles (rectus abdominis, external oblique, rectus femoris, and serratus anterior), creating instability in the upper and lower body with a suspension device while performing prone bridge exercise. These researchers found that the dual instability (both forearms and feet suspended) elicits a significantly very high activation (>60% MVIC) for the rectus abdominis and external oblique compared to the prone bridge on the floor. However, they did not find a significant difference in any of the muscles analysed while compared dual and single instability (suspended forearms or suspended feet). Whereas single instability significantly increased rectus femoris [from low (<20% MVIC) to moderate (21- 40% MVIC)], rectus abdominis and external oblique [from moderate (21-40% MVIC) to very high (>60% MVIC)] muscular activity when compared to the prone bridge on the floor.

The effects of the dual condition on the lower limb have been less studied. Despite this, Moras et al. (2019) compared a dynamic half-squat in stable conditions using a flywheel machine with the same exercise in dual condition (flywheel and standing on Pielasters), without finding significant differences in force production, even though force values were slightly lower for the dual condition (632.43±159.13 N) compared to the stable condition (658.37±156.35 N). Contrarily, performing dynamic squat training (6 sets of 6 reps; with an individual optimal load) on a WBV (30 Hz at 4 mm of amplitude) combined with repeated sprint training (3 sets of 6 reps of 20 meters shuttle run with 180º change of direction) (Suarez-Arrones et al., 2014) or functional eccentric-overload exercises (8 exercises between 6 to 10 reps with an inertial load ranged from 0.27 $Kg{\cdot}m^{-2}$ to 0.11 $Kg{\cdot}m^{-2}$) (Tous-Fajardo et al., 2016) elicited higher performance than traditional resistance training (lunges, half-squats, and calf raise; 50-100% body mass) on sprint, change of direction, and jumping performance. Furthermore, blood flow restriction training combined with WBV resistance training (30 Hz and parallel squat with dynamic loading) improved critical power, overall capillary-to-fibre ratio, and total lean body mass in endurance-trained men (Mueller et al., 2014).

Combining methods in the upper body suggests a greater activation in push exercises, mainly if the combined methods are unstable surfaces (dual instability). On the other hand, the dual condition does not increase force production on the lower body, but the combination of WBV and other training methods shows a similar result to those obtained in previous studies on the upper body with unstable surfaces. However, the effects of dual instability, understood as a perturbation, on the lower body have been studied under the superimposed vibration condition.

Superimposed vibration

The studies mentioned above support the idea that the improvement in neuromuscular performance comes from the boosting effect of WBV in combination with other training methods. When training with vibration, athletes commonly stand on the platform and perform the exercise over the lower body while receiving the vibratory stimulus from the bottom. For this reason, superimposed vibration devices were designed to transmit the vibration to other training materials such as handles or straps (Issurin, 2005). Thus, Moras et al. (2010) examined the effects of using a vibration bar in bench press exercise in 2 isometric positions (extended elbows and flexion elbows) and at 3 vibration frequencies (0 Hz, 25 Hz and 45 Hz), obtaining that the activation of the triceps brachii (at 25 Hz and 45 Hz in both positions), the pectoralis major and the anterior deltoid (both at 25 Hz and 45 Hz in an extended elbow and 45 Hz in flexion elbow) was significantly increased compared to the non-vibration condition. Moreover, other superimposed vibrations have been used in the past in devices such as dumbbells (Bosco et al., 1999; Cochrane & Hawke, 2007), bars (Mischi & Cardinale, 2009; Poston et al., 2007; Xu et al., 2013), and cables (Issurin, 2010; Issurin & Tenenbaum, 1999). Likewise, superimposed vibration has been used to study the training effects on the lower body. Thus, the addition of vibration (30 Hz at 2.5 mm of amplitude) had no effects during four weeks of dynamic calf-raise on a seated rig (75-90% 1RM) (Carson et al., 2010). However, superimposed vibration on a BOSU (35-40 Hz and 2 to 4 mm of amplitude) enhanced the reaction time of peroneus brevis, longus, and tibialis anterior in athletes with chronic ankle instability during six weeks of training (Sierra-Guzmán et al., 2017). Furthermore, sEMG has been used to evaluate the activity of different muscles during an exercise with superimposed vibration (Xu et al., 2015). Thus, Marín and Hazell (2014) found higher activation of the gastrocnemius medialis, vastus medialis, and multifidus during 60° knee flexion static half-squats with superimposed vibration on a BOSU (30 Hz and 50 Hz and 1 mm of amplitude) in comparison to the stable condition. The gastrocnemius medialis, vastus medialis, and lumbar multifidus activation increased significantly between 20% and 35% under the superimposed vibration condition compared to the stable condition. Although there is not enough evidence to agree on the effects of the combination of methods on the lower body. In this sense, it could be interesting to study the combination of suspension devices with other exercise training methods such as the half-squat or the Bulgarian squat.

Currently, to our knowledge, there are only four devices with superimposed vibration allowing the lower body training. Two of these devices are similar to vibration platforms, consisting of a small platform to improve flexibility in gymnasts (Kinser et al., 2008; Sands et al., 2006) and a platform with a bi-engine that provides vibration on a leg press machine (Pujari et al., 2019). The other two devices are Vibrosphere (ProMedvi), a superimposed vibration wobble board (Cloak et al., 2013), and Vibalance (Viequipment), a platform that combines vibration with different degrees of instability even though neither of these devices superimposed vibration on suspension straps.

Thus, the invention of a vibratory system for suspension training could improve the effects of suspension training on lower-body exercises by the superimposed vibration on the suspension device. Additionally, several superimposed vibration devices have been patented as the muscle stimulation device (patent number: US9174079B2), the barbell or dumbbell with vibration device (patent number: EP244127), or the vibratory exercise equipment (patent number: US2007259759).

Methods for assess the perturbations

Surface electromyography

Several studies (G. Anderson et al., 2013; Escamilla et al., 2010; Krause et al., 2018; Snarr & Esco, 2014) sustained that sEMG is an excellent method to quantify the effects of perturbed loads on muscle recruitment. Besides, in several studies, sEMG has been used to quantify the muscle activation under suspension conditions, primarily in the upper body (Atkins et al., 2015; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014; McGill et al., 2014b; Snarr et al., 2016), and secondly for the lower body (Krause et al., 2018; Malliaropoulos et al., 2015; Miller et al., 2019), despite the limited amount of studies.

Methodological considerations must be considered in the use of sEMG to ensure the validity and reliability of the muscle activation during measurements with perturbed loads. Firstly, the protocol is used to normalize the electromyographic signal through MVIC. In this vein, the explosiveness to obtain MVIC matters (Vigotsky et al., 2018). Compared to explosive actions, the use of isometric and slow dynamic movements for achieving a maximal contraction results in more reliable and easy-to-compare electromyographic signals (Alizadehkhaiyat & Frostick, 2015). Secondly, the electrode placement and attachment of the electrodes, following the SENIAM (Hermens et al., 2000) guidelines. Thirdly, the use of filters to record the muscle signal and avoid noise in the electrical signal, such as when performing vibratory exercise conditions (Borges et al., 2017). Lastly, whether muscle activity has been normalized (MVIC), it is recommended to categorize the magnitude of activation as low (<20% MVIC), moderate (21-40% MVIC), high (41-60% MVIC) and very high (>60% MVIC) as described in previous studies (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Escamilla et al., 2010; Mok et al., 2014). On the other hand, EMG data from the Root Mean Square signal can be expressed as integrated electromyography (IEMG) in millivolts (mV). This signal processing has the disadvantage that the EMG signal is not normalised, and it will not be possible to categorise the activation into low, moderate, high, and very high.

About the technique of exercises performed on unstable surfaces or in suspended conditions, due to the sensitivity of the electrodes and connectors of the electromyographic signal acquisition systems, the exercises must have a controlled pace, and the posture must be maintained as consistently as possible (McGill et al., 2014b; Mok et al., 2014; Snarr et al., 2014). In dynamic exercises on unstable or suspended surfaces, alterations in the timing of excitation can occur, which means a delay in the acquisition of the EMG signal reading by the electrode due to the velocity of the movement (Vigotsky et al., 2018). Likewise, due to changes in joint angle during dynamic contractions, the electrodes can move concerning the muscle fibres and cause changes in tissue conductivity (Farina, 2006). All these factors are inherent, and therefore very difficult or impossible to control. In this vein, recording sEMG in dynamic exercises under controlled conditions is essential (Enoka & Duchateau, 2015). Moreover, in explosive dynamic exercises, or those involving jumps, notch filtering should be used to reduce the amplitude of possible movement artefacts, as Mackala et al. (2013) indicated during the EMG signal analysis in the take-off phase the SJ and CMJ.

Force plates and load cell

The study and quantification of VGRF using force plate has been carried out in upper body exercises such as bench press (Jandacka & Uchytil, 2011; McMaster et al., 2014; van den Tillaar et al., 2012) or push-ups (Dhahbi et al., 2017; Koch et al., 2012; Zalleg et al., 2020) or lower body exercises such as a single-leg squat, forward and reverse lunges or squats (Comfort et al., 2015; Kellis et al., 2005; McMaster et al., 2014), in jumping ability (SJ and CMJ) (Mackala et al., 2013; McMaster et al., 2014; Souza et al., 2020) and in COD tasks (de Hoyo et al., 2014; Maloney et al., 2016; Spiteri et al., 2015). Likewise, VGRFs have been measured to examine the effects of perturbed loads in exercises such as single-leg squat on wobble board (P. de B. Silva et al., 2018) or in push-ups and inverted rows under suspended conditions (McGill et al., 2014a, 2014b). In suspended push-ups and inverted rows, VGRFs have mainly been analysed to determine the effects of the fundamental principle of the resistance vector on trunk inclination through changes in the angle or length of the suspension straps (Giancotti et al., 2018; Gulmez, 2017). On the other hand, the force production has been evaluated with a load cell in the isometric half-squat under stable and unstable conditions (Saeterbakken & Fimland, 2013). The load cell seems to be a useful tool to quantify the force production in exercises with perturbation such as suspension devices. Various researchers (Giancotti et al., 2018; Gulmez, 2017; Melrose & Dawes, 2015) placed a load cell between the anchor point and the suspension strap to quantify the forces in push-up inverted row exercises. In these studies, the forces of isometric exercises were measured using load cells. On the other hand, dynamic exercises, such as suspended biceps curl (Cayot et al., 2017), were also measured with these sensors. The quantification of load in suspended exercises using load cells constitute a more affordable and practical way to evaluate force production because it records the forces exerted on the suspension straps and the magnitude of the forces generated depends on the degree of instability caused by the suspension device and body position (Maté-Muñoz et al., 2014). Currently, apart from the data acquisition equipment to acquire data in the laboratory as the BIOPAC system (BIOPAC System, Inc., Goleta, CA), there are more portable devices as the load cell of Chronojump-Boscosystem (Barcelona, Spain), the force sensor of Musclelab (Ergotest Innovation AS, Stathelle, Norway) or a completely portable and wireless as the load cell Suiff (Estel S.L., Barcelona, Spain), the S-Beam load cell (wireless version; Biometrics Ltd, Newport, UK), the Musclelab force sensor (wireless version; Ergotest Innovation AS, Stathelle, Norway) and the Powrlink sensor (Aerobis fitness GmbH, Cologne, Germany).

The force production values recorded with force plates or load cells can be expressed in absolute values (Kg or Newtons), although it is usually expressed in relative values (percentage of body mass resistance), and so these values are normalized from the bodyweight of the subject by this equation: body mass resistance (%) = load/body weight x 100.

Accelerometer

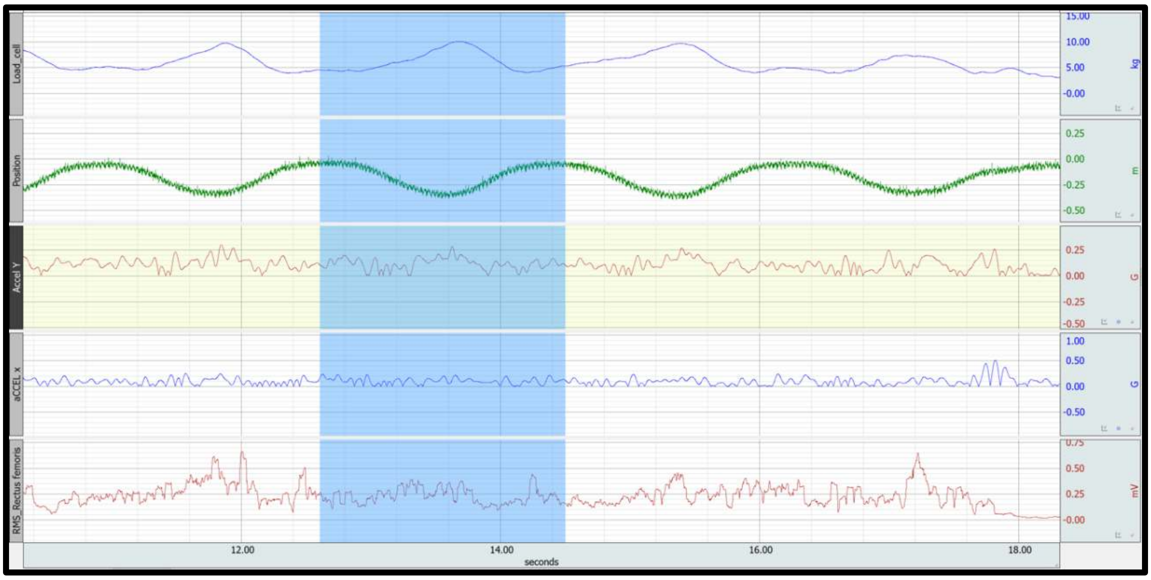
The previous sections have shown that different researchers have studied the influence of perturbed loads on muscle activation and force production (Drinkwater et al., 2007; McBride et al., 2010; Saeterbakken & Fimland, 2013) but not on the amount of instability. Including the perturbation loads is challenging for the athletes because it increases task demands and provides the 4D force effect. However, the destabilising material must provide the appropriate amount of instability because the perturbation effect on athletes could be different, in terms of the level of difficulty and be more or less challenging, despite the task being performed on the same destabilising material, such as a suspension device or a Wobble Board (Moras, 2017). Thus, it is relevant to determine the magnitude of the effect of the perturbation on both the athlete and performance variables, such as force production, muscle activity or balance. Accordingly, inertial accelerometers could be useful to determine the amount of perturbation involved in performing an exercise. Despite this, to our knowledge, no studies have quantified the amount of instability using inertial accelerometers in the strength and conditioning context (**Figure 15**). Only Moras and Vázquez-Guerrero (2015) described the amount of instability when comparing the force output under different stability conditions of a flywheel squat using an accelerometer. In other contexts, Thiel et al. (2014) used different accelerometers to assess professional dancers' quality of the movements. Moreover, Johnston et al. (2018) used an inertial sensor to detect minor changes when performing the Y Balance test in healthy adults, and Barbado et al. (2018) proposed the smartphone's accelerometer to describe core stability in different unstable environments.

Some investigations have quantified or altered the balance with good validity and reliability using different methods such as force platforms (Gopalai et al., 2011), stabilometres (Kovacikova et al., 2015) and pressure mats (Goetschius et al., 2018) in the context of ankle and knee rehabilitation processes, fall prevention and postural balance in different populations. However, all the methods mentioned above have several limitations in assessing the amount of instability. All of them, displayed on the floor, interfere with the signal and, therefore, the validity of the assessment itself. Other studies (Barbado et al., 2018; Vázquez-Guerrero et al., 2016) suggested that the use of mean accelerations values might not be the best way to describe the amount of instability due to mean or peak root mean square acceleration values do not reflect the ability to maintain the posture, because the moments when the participants are balanced are taken into consideration for the calculations (Thiel et al., 2014). Despite this, inertial accelerometers can be a suitable measurement device for quantifying the amount of instability caused by perturbation load in an unstable or suspended environment.

To our knowledge, recently, two methods are being used for measuring the amount of instability. One of the proposals is the sum of peaks of acceleration during the entire movement. This consists of placing the inertial accelerometer at the body's centre of mass and measuring all the variations in the three axes (X, Y, Z) during each repetition of an exercise. For instance, a set of 5 repetitions of single-leg squat on BOSU (dome side down). During each repetition, the destabilising material's peaks of acceleration to maintain the position and perform the task are registered. Then, the acceleration signal is calculated as the root mean square and the sum of the peaks of the entire set is obtained. Finally, the acceleration vector is calculated by the quadratic combination of the three axes (X, Y, Z). This approach, which considers the quadratic combination of the acceleration in anteroposterior and mid-lateral axes (Moras & Vázquez-Guerrero, 2015), seems to provide an accurate approach for quantifying the amount of instability (BCMA) in different unstable and suspended environments (Romero-Franco et al., 2013). The other proposal is based on entropy, which describes signal predictability in postural control (Deffeyes et al., 2011). Specifically, sample entropy measures the variability of the movement (Richman & Moorman, 2000), determining that low variability (constancy in the movement) results in a low entropy score and a high level of variability means a high entropy score (Stergiou et al., 2006). Thus, Moras et al. (2019) used sample entropy (nonlinear technique) to determine variability in force production while performing half-squat under stable and unstable conditions (Pielasters) in a flywheel in physically active participants, without finding differences between surfaces and force production using sample entropy (X^2 = 3.420 p = 0.527).

This finding confirmed the complexity of selecting the optimal destabilizing material based on the exercise features and confirmed the importance of knowing the magnitude of the perturbation. Recently, research is being carried out to determine the amount of perturbation through accelerometers placed on the athlete's body and analysing this signal using sample entropy (Moras, 2017).

Figure 15. *Methods to Quantify the Perturbation Elicit in a Suspended Lunge*



Note. Integrated and synchronized acquisition of electromyographic signal (root mean square of the rectus femoris), force production (load cell (kg)), and variations in the body centre of mass acceleration (BCMA) from X and Y axis (accelerometer (g)).

The blue shaded area represents the entire suspended lunge phase (eccentric-concentric repetition).

OBJECTIVES AND HYPOTHESIS

The overall objective of this PhD dissertation is to quantify the force production, muscle activity and the magnitude of instability in the Bulgarian squat and other lower body exercises in an unstable environment (unstable surfaces and suspension devices)— secondly, the development of a technological invention based on a vibratory system for suspension training. The objectives and hypotheses of each of the studies are shown below.

STUDY 1

Objective:

To identify the level of activation of the muscles involved in the most studied exercises under suspended conditions and compare the activation levels of the different exercises performed under stable and suspended conditions.

STUDY 2

Objective:

Determine the force exerted on the suspension strap when performing a suspended lunge in different positions, paces and contraction patterns.

Hypothesis:

The force exerted on the suspension strap would progressively increase with the variation of the position (suspension strap height and distance between the suspended leg and the front leg) and the speed of execution.

The force would be better in the dynamic contraction scheme than in the isometric one.

STUDY 3

Objective:

Analyse the effects of an unstable environment (suspension and unstable surfaces) in Bulgarian squat exercise on muscle activity and force production (VGRF and force exerted on the suspension strap).

Hypothesis:

The unstable environment would increase muscle activity and VGRF compared to the stable condition (Bulgarian squat on the floor).

The force exerted on the suspension strap would be lower on the suspended lunge-BOSU than under the other unstable conditions.

STUDY 4

Objective:

Establish the relationship between the BCMA, muscle activity and the force exerted on the suspension strap during different suspended lunge conditions.

Hypothesis:

The relationship between the BCMA and muscle activation would have a positive trend, while the BCMA and the force exerted on the suspension strap would have a negative trend.

STUDY 5

Objective:

Examine the amount of instability, the prime movers' muscle activity, and the OMNI- Res while executing a loaded free barbell half-squat under different unstable conditions.

Hypothesis:

The BOSU-down condition would elicit both a higher BCMA and a higher OMNI-Res, whereas this would elicit lower muscular activity of primary squat movers than the others conditions.

The relation between BCMA and the highest performance limb muscle activity would have a positive trend for all exercise conditions.

STUDY 6

Objective:

To assess the effects of the vibration device on muscle activation and OMNI-Res in the dynamic suspended supine bridge and hamstring curl exercises.

Hypothesis:

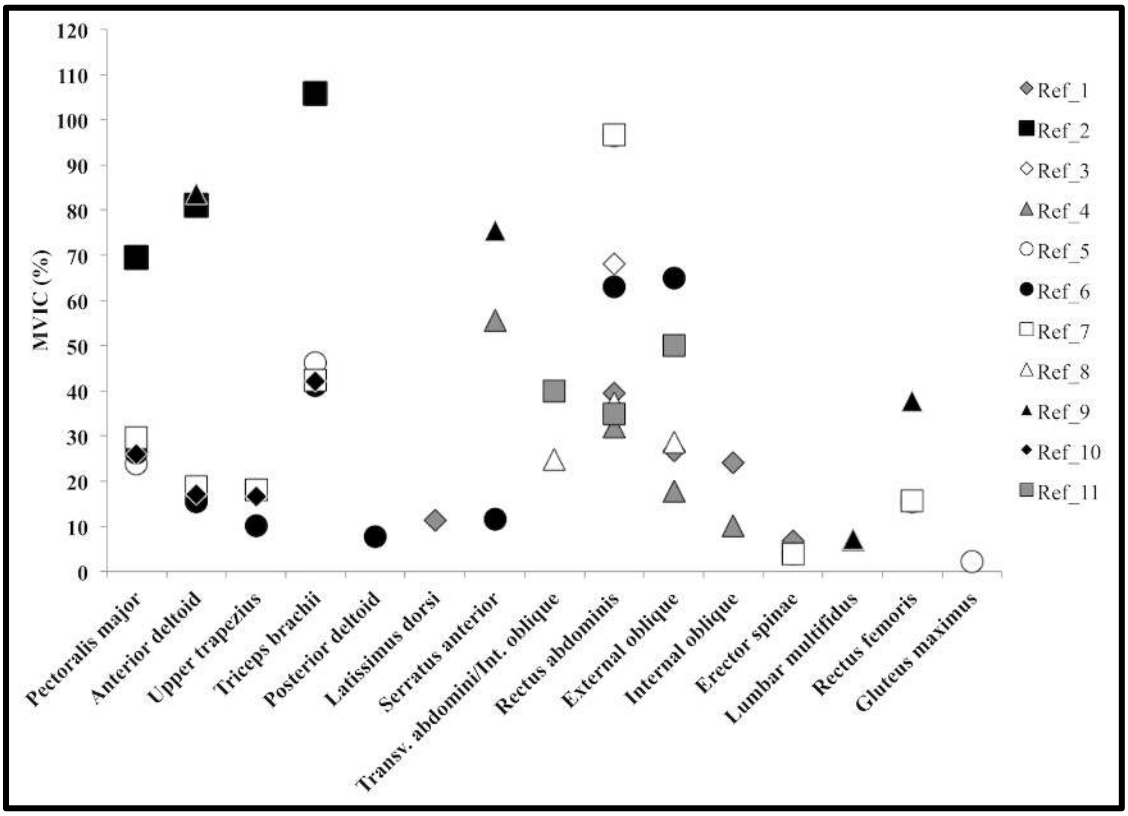
The superimposed vibration on the suspension device would elicit a higher muscle activity than the suspended condition without vibration in both exercises.

OMNI-Res would be higher in the superimposed vibration suspended condition than those without vibration in the supine bridge and hamstring curl.

RESULTS

A systematic review of the most studied exercises under suspended conditions was carried out in study 1. This study showed that activation of the upper body and core muscles ranged from low (<21% MVIC) to very high (> 60% MVIC) when performing push-ups, inverted rows, prone bridges and hamstring curls under suspended conditions (**Figure 16**, **Figure 17**, **Figure 18**, **Figure 19**). The comparison between suspended and stable conditions revealed that activations of pectoralis major, anterior deltoid, upper trapezius, triceps brachii, latissimus dorsi, serratus anterior, rectus abdominis, external oblique, internal oblique, lumbar multifidus and rectus femoris were significantly greater in suspended push-up compared to push-up (Beach et al., 2008; Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015; Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; McGill et al., 2014a; Snarr & Esco, 2013b). The pulley suspended device caused significant increases in activation of upper trapezius, triceps brachii, posterior deltoid, rectus abdominis, external oblique, erector spinae, rectus femoris and gluteus maximus relative to traditional push-up (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014). The traditional push-up resulted in significantly higher pectoralis major and anterior deltoid activations than suspension push-up with pulley (Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015). However, for certain muscles, like pectoralis major (Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015), anterior deltoid (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014) and serratus anterior (Borreani, Calatayud, Colado, Tella, et al., 2015; McGill et al., 2014a), significantly greater activation was found in traditional push-up in comparison with suspension push-up in the aforementioned studies. Furthermore, a comparison between suspended and traditional inverted row, prone bridge, and hamstring curl showed that the activations of the middle trapezius, posterior deltoid, rectus abdominis, internal oblique, external oblique and erector spinae were higher in suspension inverted row compared to inverted row; however, the increases were not statistically significant (McGill et al., 2014b; Snarr et al., 2014; Snarr & Esco, 2013a). Activation of latissimus dorsi was significantly greater in an inverted row compared to a suspended inverted row, but biceps brachii activity was significantly higher in suspension inverted row compared to inverted row (Snarr et al., 2014). Activations of core muscles (rectus abdominis, external oblique, erector spinae and rectus femoris) were significantly greater in suspension prone bridge than prone bridge (Atkins et al., 2015; Byrne et al., 2014; Snarr & Esco, 2014). Activations of biceps femoris and semitendinosus were significantly greater in suspension hamstring curl compared to traditional exercise with and without destabilising devices (Malliaropoulos et al., 2015).

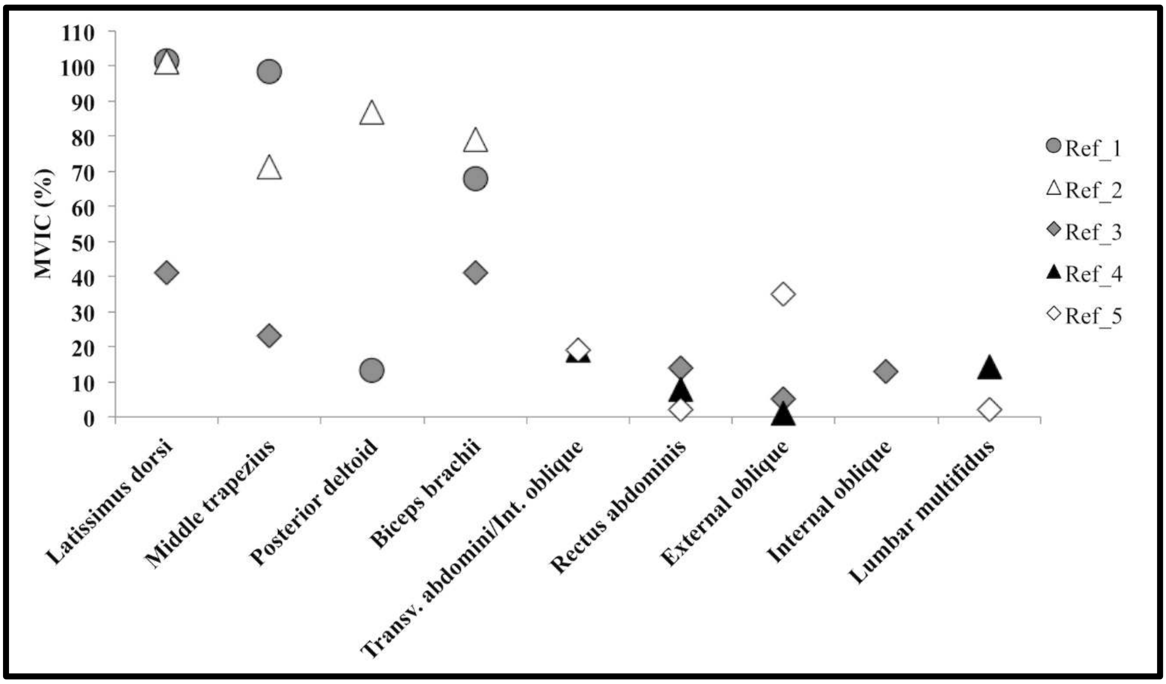
Figure 16. *Percentage of Maximum Voluntary Isometric Contraction Achieved for Each Muscle in Suspended Push-Ups Studies.*



Note. **REF_1:** Beach et al. (2008); **REF_2:** Snarr and Esco (2013b); **REF_3:** Snarr et al. (2013); **REF_4:** McGill et al. (2014a); **REF_5:** Calatayud, Borreani, Colado, Martin, Batalha, et al. (2014); **REF_6:** Calatayud, Borreani, Colado, Martin, and Rogers (2014); **REF_7:** Calatayud, Borreani, Colado, Martin, Rogers, et al. (2014); **REF_8:**

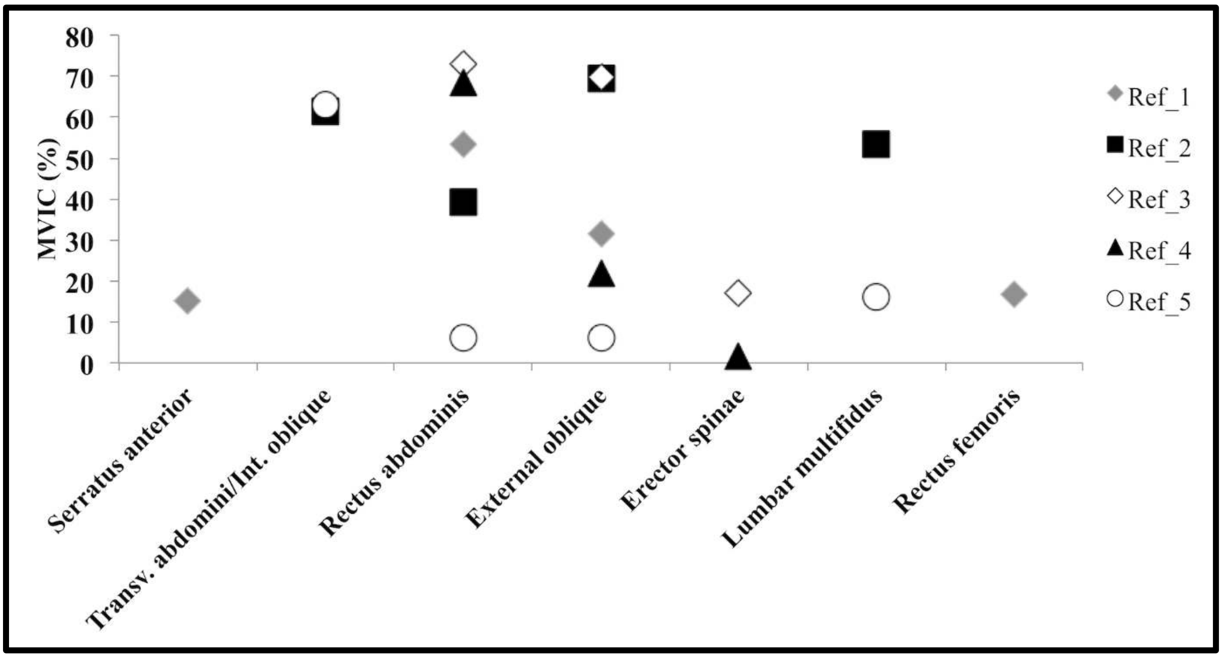
Mok et al. (2014); **REF_9:** Borreani, Calatayud, Colado, Moya-Nájera, et al. (2015); **REF_10:** Borreani, Calatayud, Colado, Tella, et al. (2015); **REF_11:** Fong et al. (2015)

Figure 17. *Percentage of Maximum Voluntary Isometric Contraction Achieved for Each Muscle in Suspended Inverted Row Studies.*



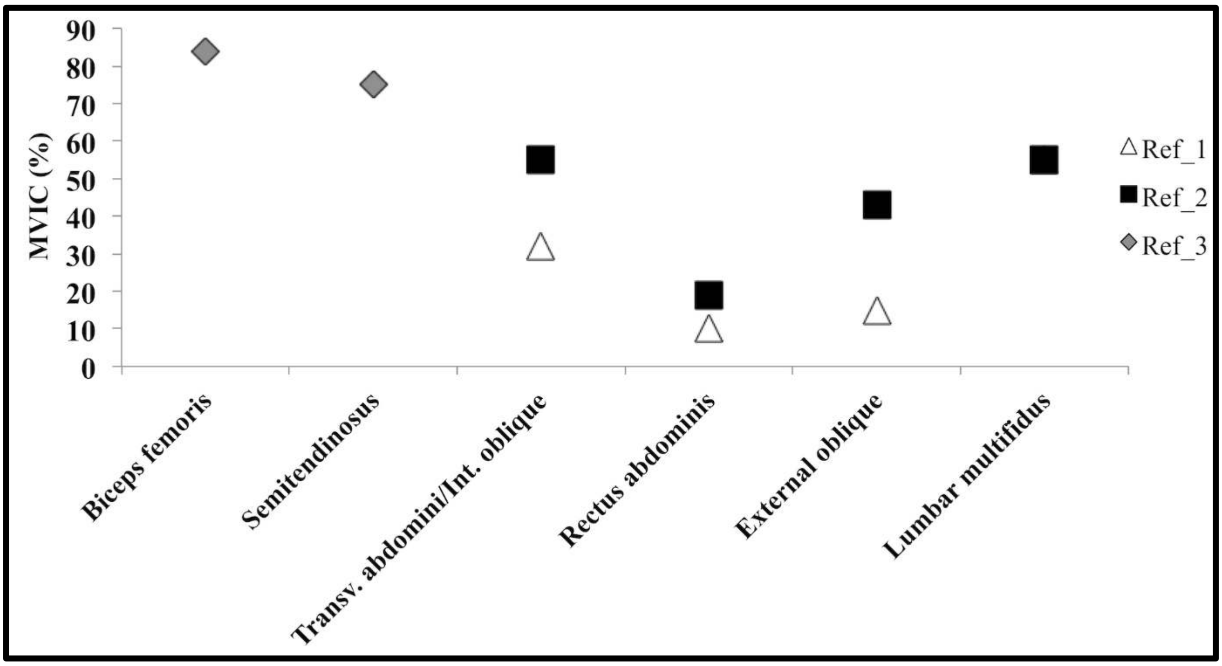
Note. **REF_1:** Snarr and Esco (2013a); **REF_2:** Snarr et al. (2014); **REF_3:** McGill et al. (2014b); **REF_4:** Mok et al. (2014); **REF_5:** Fong et al. (2015).

Figure 18. *Percentage of Maximum Voluntary Isometric Contraction Achieved for Each Muscle in Suspended Prone Bridge Studies.*



***Note*.** **REF_1:** Byrne et al. (2014); **REF_2:** Mok et al. (2014); **REF_3:** Snarr and Esco (2014); **REF_4:** Atkins et al. (2015); **REF_5:** Fong et al. (2015)

Figure 19. *Percentage of Maximum Voluntary Isometric Contraction Achieved for Each Muscle in Suspended Hamstring Curl Studies.*



Note. **REF_1:** Mok et al. (2014); **REF_2:** Fong et al. (2015); **REF_3:** Malliaropoulos et al. (2015)

The study 2 showed that performing a dynamic suspended lunge an interaction effect was not found between position (TRX_40-60, TRX_40-80, TRX_60-60 and TRX_60-80) and frequency (60, 70, and 80 bpm) for concentric force [$F_{(6, 54)} = 0.663$, $p = 0.681$, $\eta^2 = 0.06$], eccentric force [$F_{(2.41, 21.70)} = 0.834$, $p = 0.467$, $\eta^2 = 0.08$], average force [$F_{(6, 54)} = 0.799$, $p = 0.575$, $\eta^2 = 0.08$] and peak force [$F_{(2.68, 24.18)} = 0.594$, $p = 0.607$, $\eta^2 = 0.06$]. Despite, a main effect was found for position on concentric force [$F_{(3, 27)} = 8.284$, $p = 0.000$, $\eta^2 = 0.47$], average force [$F_{(3, 27)} = 6.565$, $p = 0.002$, $\eta^2 = 0.42$], and for frequency on peak force [$F_{(1.22, 11.04)} = 7.776$, $p = 0.004$, $\eta^2 = 0.46$]. Pairwise comparison indicated that TRX_60-80 significantly increases concentric and average force exerted on the suspension strap compared to TRX_40-60 ($p = 0.008$; $p = 0.007$) and TRX_60-60 ($p = 0.021$; $p = 0.020$) at the frequency of 70 bpm and also the average force exerted on the suspension strap was greater for TRX_40-80 than TRX_40-60 ($p = 0.036$) at frequency of 80 bpm. The frequency of 70 bpm and 80 bpm significantly increased the peak force exerted on the suspension strap compared to 60 bpm for TRX_60-80 ($p = 0.006$) and TRX_40-80 ($p = 0.035$), respectively (**Table 1**). A non-interaction effect was found between contraction type and position [$F_{(3,36)} = 0.862$, $p = 0.469$, $\eta^2 = 0.07$] but a main effect was found for contraction type on peak force exerted on the suspension strap [$F_{(1, 36)} = 52.346$, $p = 0.000$, $\eta^2 = 0.59$] (**Figure 20**). Furthermore, isometric suspended lunge indicated a significant main effect for position [$F_{(3, 36)} = 21.103$, $p = 0.000$, $\eta^2 = 0.64$] on relative force exerted on the suspensions strap (percentage of body mass resistance) (**Figure 21**).

Table 1. *Force Exerted on the Suspension Strap (N) during Dynamic Suspended Lunge at Four Different Positions and Three Different Frequencies. Values Showed in Mean ± SD*

	Position	Dynamic frequency			Interaction effect	
		60 bpm	70 bpm	80 bpm		
		Mean ±SD	Mean ±SD	Mean ±SD	p ($p<.05$)	η_p^2
Concentric Force	TRX_40-60	116.21±37.15	115.81± 32.61*	112.97±39.06	0.681	0.06
	TRX_40-80	120.23±33.02	122.98±38.40	122.99±43.88		
	TRX_60-60	114.74±32.41	117.40±36.83†	119.48±41.34		
	TRX_60-80	123.42±36.87	131.04±36.84* †	128.41±35.22		
Eccentric Force	TRX_40-60	159.47±43.69	160.18±41.64	156.53±49.67	0.467	0.08
	TRX_40-80	160.89±47.19	166.90±45.78	163.52±50.31		
	TRX_60-60	156.82±38.38	167.05±47.04	164.49±52.96		
	TRX_60-80	162.57±46.37	178.31±48.68	166.87±47.04		
Average Force	TRX_40-60	130.56±39.94	130.03±36.11*	126.62±42.91 ¶	0.575	0.08
	TRX_40-80	133.26±37.77	136.50±39.56	136.07±45.94 ¶		
	TRX_60-60	128.57±34.06	132.45±40.45 †	133.96±45.63		
	TRX_60-80	136.48±40.84	146.43±42.06* †	141.21±39.54		
Peak Force	TRX_40-60	205.85±63.40	215.85±64.12	221.94±83.44	0.607	0.06
	TRX_40-80	207.84±61.82 §	223.14±78.70	233.14±78.70 §		
	TRX_60-60	199.14±51.15	221.50±67.47	226.45±81.88		
	TRX_60-80	210.63±61.60 ✝	233.24±68.04 ✝	229.65±72.97		

Note. (N) = Newton; bpm = Beats per minute

* Significant differences between TRX_40-60 and TRX_60-80

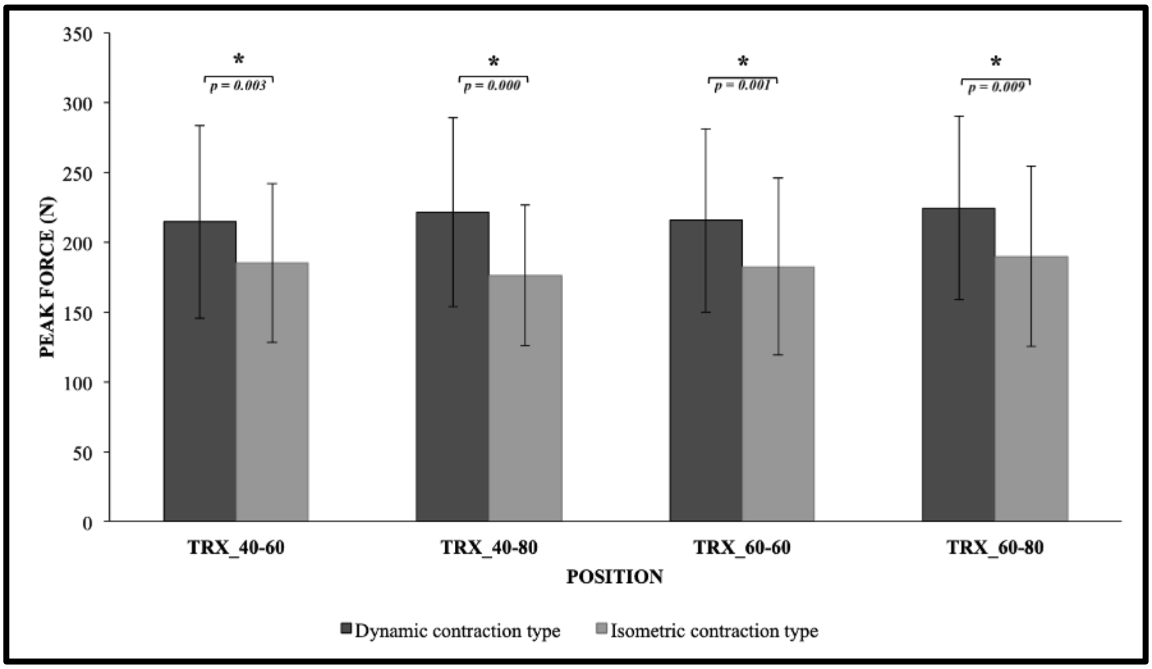
†Significant differences between TRX_60-60 and TRX_60-80

¶ Significant differences between TRX_40-60 and TRX_40-80

✝ Significant differences between frequency 60 bpm and 70 bpm

§ Significant differences between frequency 60 bpm and 80 bpm

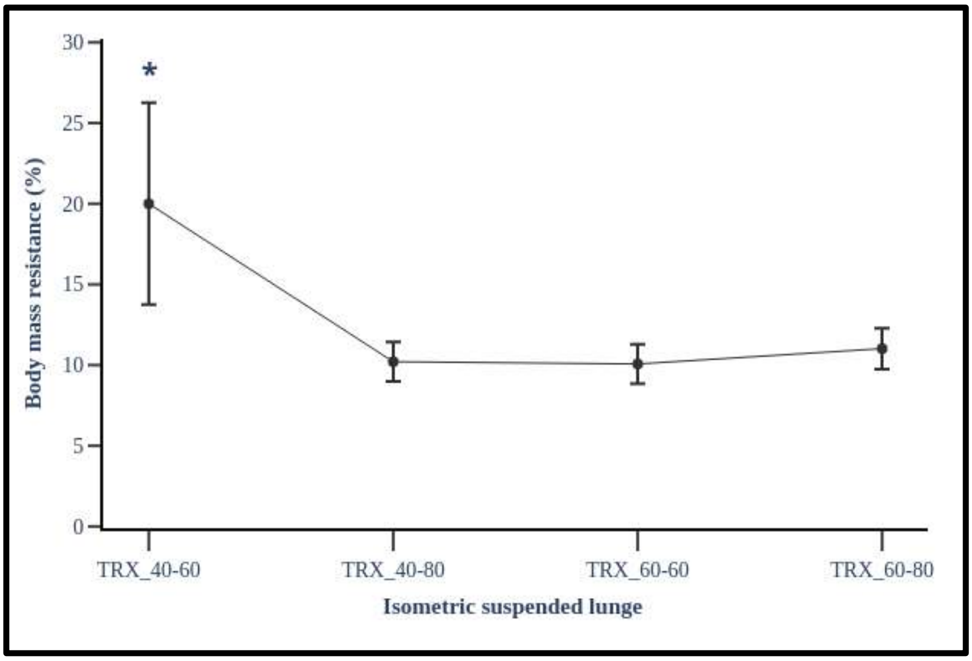
Figure 20. *Peak Force Comparison Between Dynamic and Isometric Suspended Lunge at Four Different Positions.*



Note. Each bar represents the mean, and the error bar is the standard deviation (SD).

* Significant differences (p<0.05) between dynamic and isometric contraction type

Figure 21. *Percentage of Body Mass Resistance Exerted on The Suspension Strap for Each Position Under Isometric Suspended Lunge Condition.*

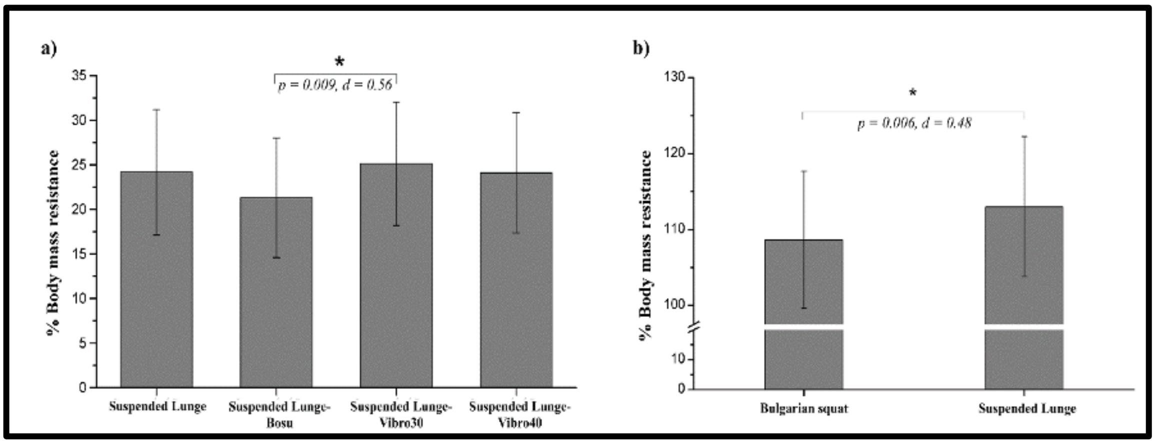


Note. Each black bullet represents the mean, and the error bar is the standard deviation (SD).

* Significantly greater (p = 0.000) than TRX_40-80, TRX_60-60, and TRX_60-80

Study 3 has also shown a main effect for exercise condition on the force exerted on the suspension strap under different suspended lunge conditions [$F_{(3,57)} = 5.106$ $p = 0.003$, $\eta_p^2 = 0.21$] and the VGRF on the front leg was significantly higher for suspended lunge than Bulgarian squat ($t_{(19)} = -3.106$) (**Figure 22**). The comparison between exercise condition and muscle activity showed that the activity of all analysed muscles (except rectus femoris) was lower. However, non-significant for suspended lunge than Bulgarian squat, nevertheless significant differences for muscle activity were found for the dual condition (suspended lunge-BOSU, -Vibro30, -Vibro40) compared to single instability (suspended lunge) and Bulgarian squat (**Table 2** and **Figure 23**).

Figure 22. *Force Values During the Bulgarian Squat and Suspended Lunge Conditions*



Note. a) Comparison between forces exerted by the rear leg on the suspension strap and exercise condition, b) Front leg force production comparison between Bulgarian squat and suspended lunge. Each bar represents the mean, and the error bar represents the standard deviation (SD).

* Significant difference (p<0.05)

Table 2. *Normalised Electromyographic Activation for Each Lower Body Muscle Under Different Lunge Conditions as a Percentage of Maximum Voluntary Isometric Contraction (%MVIC). Values are expressed as Mean ± Standard Error of the Mean (SE).*

	Bulgarian Squat (a)	**Suspended Lunge (b)**	**Suspended Lunge-BOSU (c)**	**Suspended Lunge-Vibro30 (d)**	**Suspended Lunge-Vibro40 (e)**	P-value (effect size *d*)					
RF_FL	32.72 ± 3.48†	33.50 ± 3.45†	45.30 ± 4.28	35.16 ± 3.96†§	44.90 ± 5.72	a-c 0.010 (0.72)	b-c 0.002 (0.68)	d-c 0.001 (0.55)	d-e 0.012 (0.44)		
BF	24.50 ± 2.40	21.48 ± 2.14‡§	27.21 ± 2.21	28.07 ± 2.30	26.92 ± 2.38	b-d 0.044 (0.66)	b-e 0.014 (0.54)				
Gmed	46.53 ± 4.18†§	45.54 ± 3.15†‡§	65.67 ± 4.85	55.73 ± 4.67	65.59 ± 4.98	a-c 0.000 (0.95)	a-e 0.001 (0.93)	b-c 0.000 (1.10)	b-d 0.022 (0.57)	b-e 0.000 (1.08)	
VM	64.58 ± 3.75§	62.18 ± 3.90§	67.61 ± 2.87	69.05 ± 4.45	76.23 ± 4.57	a-e 0.014 (0.62)	b-e 0.006 (0.74)				
VL	72.34 ± 4.81	64.92 ± 4.13‡§	76.79 ± 3.80	81.13 ± 6.31	87.63 ± 5.49	b-d 0.038 (0.68)	b-e 0.03 (1.05)				
RF_RL	33.51 ± 3.76	24.69 ± 3.87	23.61 ± 2.56*	26.31 ± 3.09	28.60 ± 3.00	c-a 0.019 (0.69)					
GL_FL	47.94 ± 1.40†§	45.52 ± 1.31†‡§	56.31 ± 1.96	53.83 ± 1.89§	60.26 ± 2.32	a-c 0.005 (1.10)	a-e 0.000 (1.44)	b-c 0.000 (1.44)	b-d 0.001 (1.14)	b-e 0.000 (1.75)	d-e 0.043 (0.68)
GL	46.75 ± 1.48§	42.76 ± 1.33†‡§	50.64 ± 2.20	50.53 ± 1.46	54.37 ± 2.03	a-e 0.010 (0.96)	b-c 0.012 (0.97)	b-d 0.001 (1.26)	b-e 0.000 (1.51)		

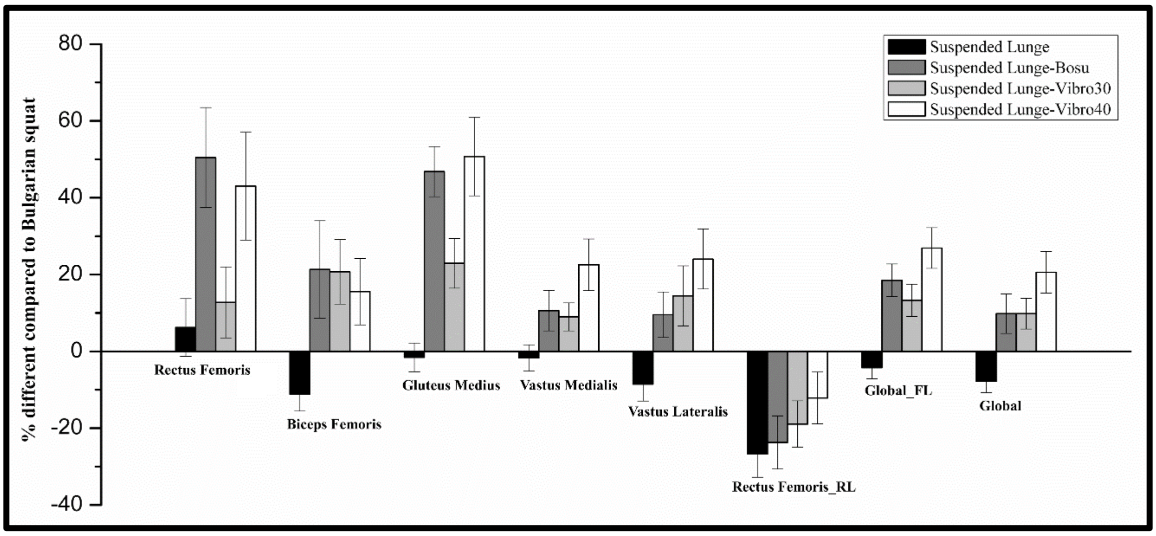
Note. RF_FL = Rectus femoris front leg; BF = Biceps femoris; Gmed = Gluteus medius; VM = Vastus medialis; VL = Vastus lateralis;

RF_RL = Rectus femoris rear leg; GL_FL = Global mean of the five front leg muscles; GL = Global mean of the six muscles

* = Significantly lower than Bulgarian squat; † = Significantly lower than Suspension lunge-BOSU

‡ = Significantly lower than Suspension lunge-Vibro30**;** § = Significantly lower than Suspension lunge-Vibro40

Figure 23. *Electromyographic Activations for all Conditions Relative to the Bulgarian Squat.*



Note. Each bar represents the mean, and the error bar represents the standard error of the mean (SEM).

FL = Front leg; RL = Rear leg.

The relationship between BCMA, muscle activity, and force exerted on the suspension strap was examined in study 4 while performing suspended lunge under unstable conditions. A significant Pearson correlation was found between BCMA, muscle activity (range from r = 0.393 to r = 0.826; p < 0.05) and force exerted on the suspension strap (range from r = -0.595 to r = -0.797; p < 0.05) for all the exercises (**Table 3** and **Table 4**) with moderate to very large effect.

Table 3. *Pearson's Correlation (r) between Forces Exerted on the Suspension Strap and Body Centre of Mass Acceleration under Suspended Lunge Conditions.*

	Suspended lunge	**Suspended lunge-Foam**	**Suspended lunge-Bosu up**	**Suspended lunge-Bosu down**
r	-0.595**	-0.797**	-0.776**	-0.741**
p-value	0.001	0.000	0.000	0.000
LC	Large	Very large	Very large	Very large

Note. LC: Level of correlation

** Statistical significance at p<0.01

Table 4. *Pearson's Correlation (r) between Muscle Activity Values for Each Muscle Analysed and Body Centre of Mass Acceleration under Suspended Lunge Conditions.*

	Suspended lunge	Suspended lunge-Foam	Suspended lunge-Bosu up	Suspended lunge-Bosu down
Rectus femoris	-0.050	0.192	0.283	-0.087
p-value	0.794	0.310	0.130	0.649
LC	Trivial	Small	Small	Trivial
Vastus medialis	**0.699****	**0.632****	**0.650****	**0.588****
p-value	0.000	0.000	0.000	0.001
LC	Large	Large	Large	Large
Vastus lateralis	**0.393***	**0.689****	**0.629****	**0.506****
p-value	0.031	0.000	0.000	0.004
LC	Moderate	Large	Large	Large
Gluteus maximus	**0.477****	**0.553****	**0.611****	**0.558****
p-value	0.008	0.002	0.000	0.001
LC	Moderate	Large	Large	Large
Gluteus medius	**0.526****	**0.749****	**0.826****	**0.646****
p-value	0.003	0.000	0.000	0.000
LC	Large	Very large	Very large	Large
Biceps femoris	**0.468****	-0.216	0.250	-0.158
p-value	0.009	0.251	0.183	0.403
LC	Moderate	Small	Small	Small

Note. LC: Level of correlation

*Statistical significance at p<0.05

** Statistical significance at p<0.01

In this vein, study 5 demonstrated a significant Pearson correlation between the highest performance limb activity and BCMA for half-squat floor (r = 0.446, p = 0.003), foam (r = 0.322, p = 0.038), BOSU-up (r = 0.500, p = 0.001), and BOSU-down (r = 0.495, p = 0.001) exercises, all of them with a moderate effect (r = 0.3 to 0.5). Additionally, the linear mixed model showed a significant fixed effect for exercise condition [$F_{(3,42)}$ = 6.706, p = 0.001] and BCMA [$F_{(1,46)}$ = 19.209, p = 0.000] on global activity (**Table 5**). Likewise, a fixed effect of exercise condition was found for the BCMA [$F_{(3,42)}$ = 30.873, p = 0.000], vastus medialis [$F_{(3,42)}$ = 6.350, p = 0.001], vastus lateralis [$F_{(3,42)}$ = 6.039, p = 0.002] , biceps femoris [$F_{(3,42)}$ = 10.051, p = 0.000] and global activity [$F_{(3,42)}$ = 10.028, p = 0.000] (**Table 6** and **Table 7**). Also, a main effect of exercise condition was found for the OMNI-Res [$X^2_{(3)}$ = 35.667 p = 0.000]. Post-hoc analysis between exercise condition and muscle activity, BCMA and OMNI-Res is showed in **Figure 24** and **Table 8**.

Table 5. *Linear Mixed Model with Exercise Condition and BCMA as the Fixed Effects and Global Activity as the Dependent Variable.*

	Parameter	ES	SE	95%CI Lower	95%CI Upper	Test (df)	*p*
Global activity	Intercept	0.83	0.24	0.35	1.31	t (54) = 3.460	0.001
	Half-squat Floor	0.76	0.12	−0.17	0.32	t (45) = 0.620	0.539
	Half-squat Foam	0.09	0.12	−0.15	0.34	t (45) = 0.728	0.470
	Half-squat BOSU-up	0.34	0.10	0.12	0.55	t (44) = 3.229	0.002
	BCMA	0.03	0.01	0.02	0.05	t (46) = 4.383	0.000
	σ_u				0.30		
	σ_ϵ				0.20		

Note. ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t– value; *p* = *p*–value; BCMA = body centre of mass acceleration; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

Table 6. *Linear Mixed Model with Exercise Condition as the Fixed Effects and BCMA as the Dependent Variable.*

	Parameter	ES	SE	95%CI Lower	95%CI Upper	Test (df)	*p*
BCMA	Intercept	26.59	1.10	24.37	28.81	t (50) = 24.043	0.000
	Half-squat floor	−11.69	1.41	−14.52	−8.85	t (42) = −8.307	0.000
	Half-squat foam	−11.69	1.41	−14.53	−8.85	t (42) = −8.309	0.000
	Half-squat BOSU-up	−8.67	1.41	−11.51	−5.83	t (42) = −6.166	0.000
	σ_u				1.80		
	σ_ϵ				3.72		

Note. ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t– value; *p* = *p*–value; BCMA = body centre of mass acceleration; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

Table 7. *Linear Mixed Model with Exercise Condition as the Fixed Effects and Muscle Activity (Vastus Medialis, Vastus Lateralis, Biceps Femoris, and Global Activity) as the Dependent Variable.*

	Parameter	ES	SE	95%CI		Test (df)	*p*
				Lower	Upper		
Vastus medialis	Intercept	0.73	0.06	0.60	0.85	t (20) = 12.116	0.000
	Half-squat Floor	−0.09	0.04	−0.17	−0.01	t (42) = −2.393	0.021
	Half-squat Foam	−0.11	0.04	−0.19	−0.03	t (42) = −2.886	0.006
	Half-squat BOSU-up	0.03	0.04	−0.05	0.11	t (42) = 0.721	0.475
	σ_u				0.19		
	σ_ϵ				0.10		
Vastus lateralis	Intercept	0.74	0.05	0.63	0.84	t (25) = 14.605	0.000
	Half-squat Floor	−0.15	0.04	−0.24	−0.06	t (42) = −3.532	0.001
	Half-squat Foam	−0.14	0.04	−0.22	−0.05	t (42) = −3.236	0.002
	Half-squat BOSU-up	−0.03	0.04	−0.12	0.05	t (42) = −0.821	0.416
	σ_u				0.15		
	σ_ϵ				0.11		
Biceps femoris	Intercept	0.33	0.02	0.27	0.38	t (31) = 11.875	0.000
	Half-squat Floor	−0.09	0.02	−0.15	−0.04	t (42) = −3.519	0.001
	Half-squat Foam	−0.07	0.02	−0.13	−0,02	t (42) = −2.763	0.008
	Half-squat BOSU-up	0.03	0.02	−0.02	0.08	t (42) = 1.199	0.237
	σ_u				0.07		
	σ_ϵ				0.07		
Global activity	Intercept	1.79	0.11	1.56	2.02	t (24) = 16.115	0.000
	Half-squat Floor	−0.34	0.09	−0.53	−0.16	t (42) = −3.794	0.000
	Half-squat Foam	−0.33	0.09	−0.51	−0.14	t (42) = −3.645	0.001
	Half-squat BOSU-up	0.02	0.09	−0.15	0.21	t (42) = 0.297	0.768
	σ_u				0.33		
	σ_ϵ				0.24		

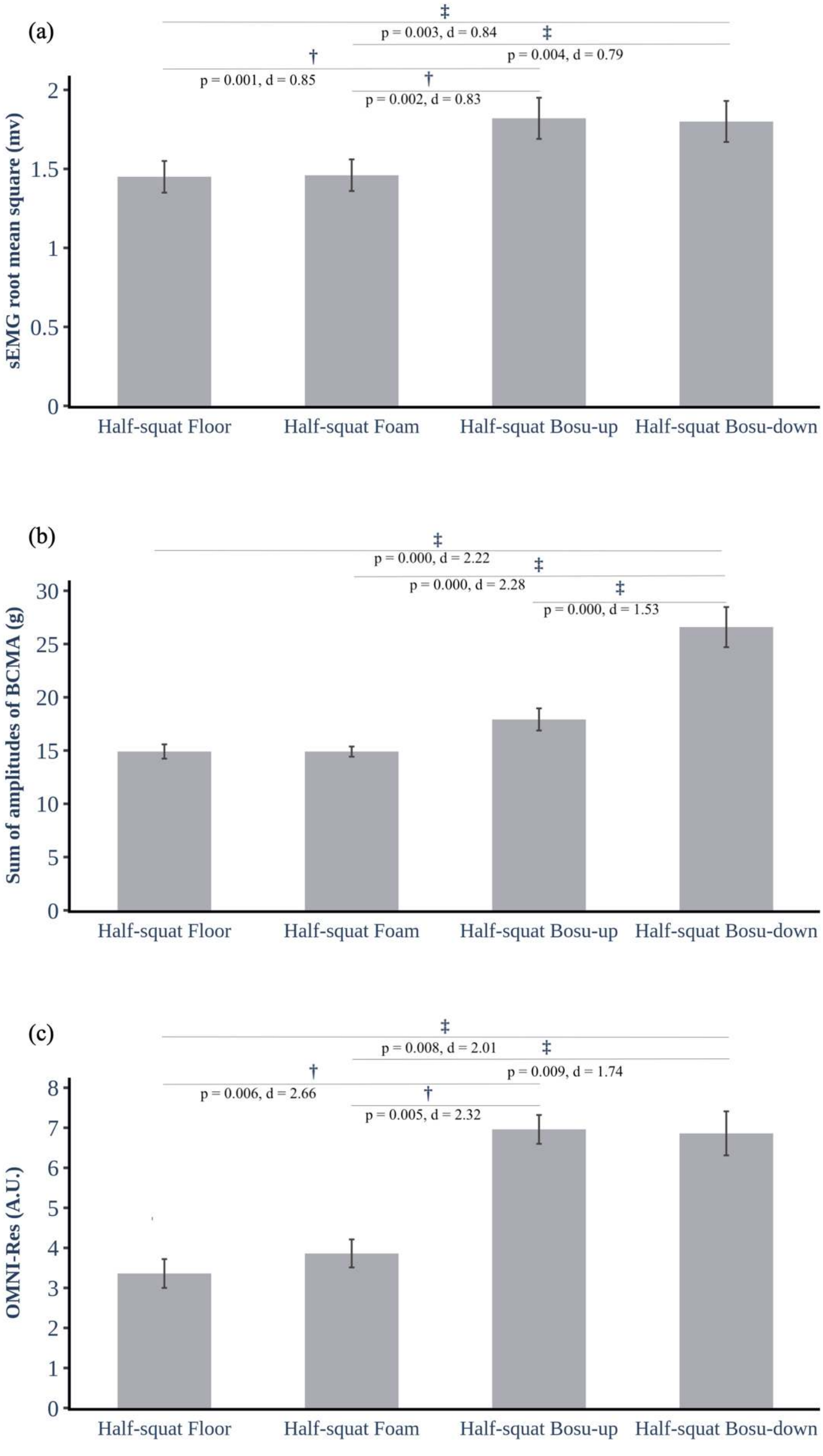
Note. ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t– value; *p* = *p*–value; BCMA = body centre of mass acceleration; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

Table 8. *Root Mean Square Surface Electromyography Values (mV) for each Muscle Analysed under Half-Squat Conditions. Values are expressed as Mean ± Standard Error of the Mean (SE).*

	Half-squat Floor (a)	Half-squat Foam (b)	Half-squat Bosu-up (c)	Half-squat Bosu-down (d)	p-value (effect size)		
Vastus medialis	0.63±0.06†	0.61±0.06†‡	0.76±0.07	0.73±0.06	c-a 0.020 (0.56)	c-b 0.005 (0.60)	d-b 0.037 (0.53)
Vastus lateralis	0.59±0.04‡	0.60±0.05‡	0.70±0.06	0.74±0.07	d-a 0.006 (0.75)	d-b 0.014 (0.67)	
Biceps femoris	0.23±0.03†‡	0.25±0.03†	0.36±0.03	0.33±0.03	c-a 0.000 (1.23)	d-a 0.006 (1.00)	c-b 0.002 (1.00)

Note. mV = millivolts.
†Significantly different than half squat-Bosu up; ‡Significantly different than half squat-Bosu down

Figure 24. *Comparison of the Collected Data under Half-Squat Conditions: (a) Global Activity§, (b) BCMA, and (c) OMNI-Res.*



Note. Each bar represents the mean, and the error bar represents the standard error of the mean (SE).

§ = Sum of the activity of the vastus medialis, lateralis and biceps femoris; sEMG = surface electromyography; mV = millivolts; BCMA = body centre of mass acceleration; A.U. = Arbitrary units

†Significantly different than half-squat BOSU-up

‡Significantly different than half-squat BOSU-down

Study 6 results reported the effect of performing a suspended supine bridge and suspended hamstring curl under non-vibration, 25 Hz and 40 Hz vibration condition on muscle activity and OMNI-Res. The linear mixed model showed a significant fixed effect for suspended supine bridge condition during the concentric phase on semitendinosus [$F_{(2,42)} = 9.05$, $p = 0.001$], gastrocnemius medialis [$F_{(2,42)} = 9.71$, $p = 0.000$], gastrocnemius lateralis [$F_{(2,42)} = 5.19$, $p = 0.010$], and global activity [$F_{(2,42)} = 16.51$, $p = 0.000$], but not on rectus femoris [$F_{(2,42)} = 0.20$, $p = 0.815$], biceps femoris [$F_{(2,42)} = 0.72$, $p = 0.490$] and gluteus maximus [$F_{(2,42)} = 1.79$, $p = 0.178$]. For eccentric phase, the exercise condition showed a significant fixed effect on semitendinosus [$F_{(2,42)} = 4.73$, $p = 0.014$], gastrocnemius medialis [$F_{(2,42)} = 8.91$, $p = 0.001$], and global activity [$F_{(2,42)} = 7.39$, $p = 0.002$], but no such effect was found on rectus femoris [$F_{(2,42)} = 0.25$, $p = 0.780$], biceps femoris [$F_{(2,42)} = 3.11$, $p = 0.055$], gluteus maximus [$F_{(2,42)} = 0.19$, $p = 0.822$] and gastrocnemius lateralis [$F_{(2,42)} = 1.24$, $p = 0.29$]. A non-significant fixed effect for suspended hamstring curl condition was found during the concentric phase on rectus femoris [$F_{(2,42)} = 1.13$, $p = 0.330$], biceps femoris [$F_{(2,42)} = 0.04$, $p = 0.955$], semitendinosus [$F_{(2,42)} = 0.72$, $p = 0.490$], gluteus maximus [$F_{(2,42)} = 0.16$, $p = 0.848$], gastrocnemius medialis [$F_{(2,42)} = 1.61$, $p = 0.210$], gastrocnemius lateralis [$F_{(2,42)} = 1.88$, $p = 0.165$], and global activity [$F_{(2,42)} = 2.60$, $p = 0.086$]. For the eccentric phase, there were also no significant fixed effect for exercise condition on rectus femoris [$F_{(2,42)} = 1.14$, $p = 0.32$], biceps femoris [$F_{(2,42)} = 1.61$, $p = 0.211$], semitendinosus [$F_{(2,42)} = 2.01$, $p = 0.146$], gluteus maximus [$F_{(2,42)} = 3.48$, $p = 0.060$], gastrocnemius medialis [$F_{(2,42)} =$

0.17, $p = 0.838$], gastrocnemius lateralis [$F_{(2,42)} = 0.06$, $p = 0.940$], and global activity [$F_{(2,42)} = 1.85$, $p = 0.169$]. Moreover, for suspended supine bridge Friedman test showed a significant main effect [$X^2_{(2)} = 26.46$, $p = 0.000$] but not for suspended hamstring curl [$X^2_{(2)} = 6.33$, $p = 0.052$] on the OMNI-Res.

Post hoc analysis between exercise condition and muscle activity and OMNI-Res is shown in **Table 9** and **Figure 25**, respectively. Additionally, **Figure 26** and **Figure 27** show the standardized differences, expressed as Cohen *d* effect size*,* for suspended supine bridge and suspended hamstring curl with significant small (*d* from 0.2 to 0.6) to moderate (*d* from 0.6 to 1.2) effects for suspended supine bridge under non-vibration, vibration at 25 Hz and 40 Hz condition on muscle activity.

Table 9. *The sEMG Activity for each Analysed Muscle under Suspended Supine Bridge and Hamstring Curl Conditions during Concentric and Eccentric Phase.*

Exercise phase	Suspended supine bridge condition	Mean ±SE	***p*-value (effect size)** Vs. Non-Vibration *p (ES)*	Vs. 25 Hz vibration *p (ES)*
		Rectus femoris		
Concentric	Non-Vibration	1.7±0.3	-	-
	Vibration at 25 Hz	1.8±0.4	1.0 (0.05)	-
	Vibration at 40 Hz	2.0±0.5	1.0 (0.15)	1.0 (0.10)
		Biceps femoris		
	Non-Vibration	19.1±1.6	-	-
	Vibration at 25 Hz	20.2±1.6	0.706 (0.15)	-
	Vibration at 40 Hz	19.6±1.8	1.0 (0.07)	1.0 (-0.08)
		Semitendinosus		
	Non-Vibration	19.7±1.4	-	-
	Vibration at 25 Hz	22.9±1.5†	0.003 (0.47)	-
	Vibration at 40 Hz	23.2±1.7†	0.001 (0.46)	1.0 (0.03)
		Gluteus maximus		
	Non-Vibration	14.8±1.7	-	-
	Vibration at 25 Hz	16.1±2.3	0.545 (0.14)	-
	Vibration at 40 Hz	16.6±2.2	0.226 (0.19)	1.0 (0.04)
		Gastrocnemius medialis		
	Non-Vibration	30.2±2.0	-	-
	Vibration at 25 Hz	37.4±2.1†‡	0.000 (0.75)	-
	Vibration at 40 Hz	32.8±1.8	0.364 (0.30)	0.025 (-0.50)
		Gastrocnemius lateralis		
	Non-Vibration	36.5±3.1	-	-
	Vibration at 25 Hz	41.7±3.1†	0.008 (0.36)	-
	Vibration at 40 Hz	38.6±3.1	0.633 (0.14)	0.181 (-0.22)
		Global activity		
	Non-Vibration	20.3±1.1	-	-
	Vibration at 25 Hz	23.4±1.0†	0.000 (0.60)	-
	Vibration at 40 Hz	22.1±1.1†	0.005 (0.34)	0.073 (-0.24)

Table 9. *(Continued).*

Exercise phase	Suspended supine bridge condition	Mean ±SE	*p*-value (effect size)	
			Vs. Non-Vibration	Vs. 25 Hz vibration
			p (ES)	*p (ES)*
		Rectus femoris		
Eccentric	Non-Vibration	2.0±0.3	-	-
	Vibration at 25 Hz	1.9±0.3	1.0 (-0.08)	-
	Vibration at 40 Hz	2.0±0.3	1.0 (-0.02)	1.0 (0.07)
		Biceps femoris		
	Non-Vibration	14.5±1.3	-	-
	Vibration at 25 Hz	16.5±1.7	0.081 (0.28)	-
	Vibration at 40 Hz	14.7±1.4	1.0 (0.04)	0.156 (-0.24)
		Semitendinosus		
	Non-Vibration	16.5±1.3	-	-
	Vibration at 25 Hz	18.1±1.2†	0.046 (0.28)	-
	Vibration at 40 Hz	18.3±1.3†	0.024 (0.29)	1.0 (0.03)
		Gluteus maximus		
	Non-Vibration	8.6±1.0	-	-
	Vibration at 25 Hz	8.3±0.8	1.0 (-0.07)	-
	Vibration at 40 Hz	8.6±1.0	1.0 (-0.01)	1.0 (0.07)
		Gastrocnemius medialis		
	Non-Vibration	24.4±1.8	-	-
	Vibration at 25 Hz	29.9±1.9†	0.000 (0.63)	-
	Vibration at 40 Hz	27.5±1.9	0.068 (0.35)	0.215 (-0.27)
		Gastrocnemius lateralis		
	Non-Vibration	37.6±3.2	-	-
	Vibration at 25 Hz	39.0±2.9	1.0 (0.10)	-
	Vibration at 40 Hz	36.4±2.8	1.0 (-0.08)	0.368 (-0.19)
		Global activity		
	Non-Vibration	17.3±0.9	-	-
	Vibration at 25 Hz	18.9±0.9†	0.001 (0.40)	-
	Vibration at 40 Hz	17.9±0.9	0.440 (0.15)	0.073 (-0.24)

Table 9. *(Continued)*

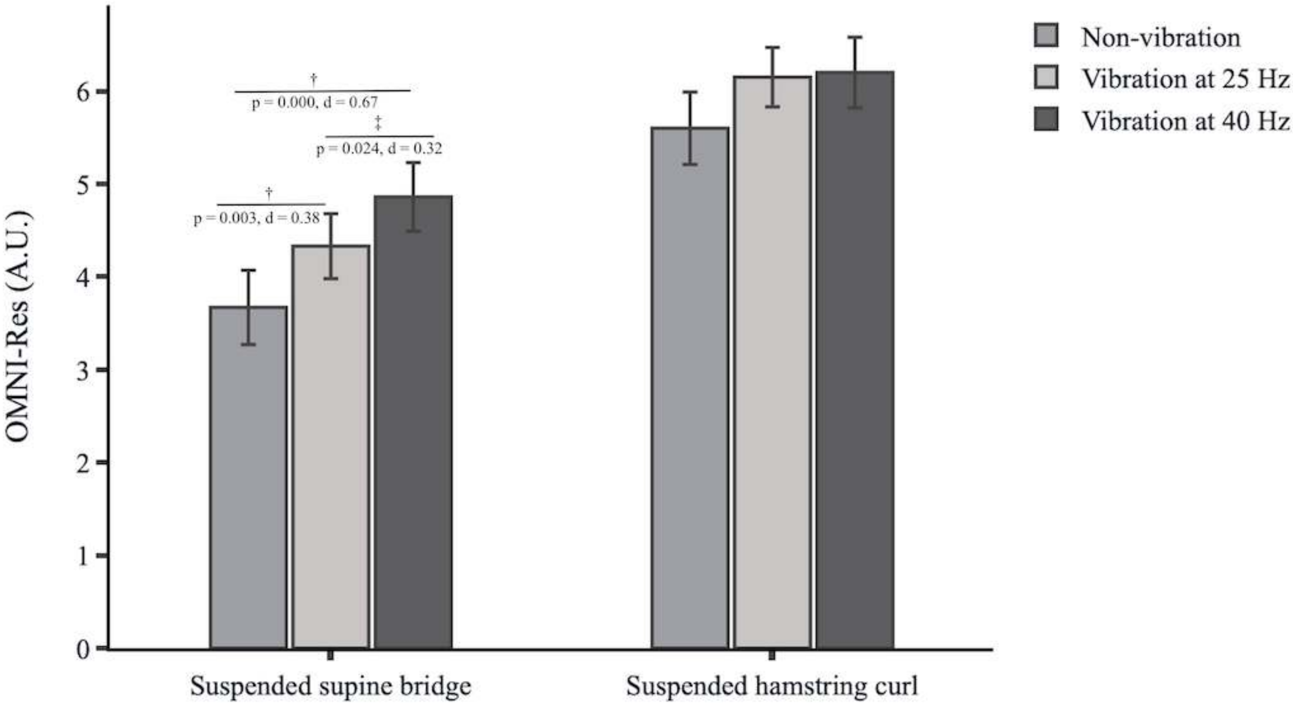
Exercise phase	Suspended hamstring curl condition	Mean ±SE	*p*-value (effect size) Vs. Non-Vibration	Vs. 25 Hz vibration
			p (ES)	*p (ES)*
		Rectus femoris		
Concentric	Non-Vibration	1.3±0.1	-	-
	Vibration at 25 Hz	1.4±0.1	0.945 (0.12)	-
	Vibration at 40 Hz	1.2±0.1	1.0 (-0.06)	0.445 (-0.19)
		Biceps femoris		
	Non-Vibration	23.6±1.4	-	-
	Vibration at 25 Hz	23.7±1.3	1.0 (0.01)	-
	Vibration at 40 Hz	24.0±1.6	1.0 (0.05)	1.0 (0.04)
		Semitendinosus		
	Non-Vibration	24.9±1.7	-	-
	Vibration at 25 Hz	26.2±1.6	0.736 (0.17)	-
	Vibration at 40 Hz	25.8±1.7	1.0 (0.11)	1.0 (-0.05)
		Gluteus maximus		
	Non-Vibration	12.7±1.1	-	-
	Vibration at 25 Hz	13.1±1.4	1.0 (0.07)	-
	Vibration at 40 Hz	12.9±1.1	1.0 (0.03)	1.0 (-0.05)
		Gastrocnemius medialis		
	Non-Vibration	37.0±3.0	-	-
	Vibration at 25 Hz	37.6±2.0	1.0 (0.05)	-
	Vibration at 40 Hz	40.8±3.4	0.310 (0.25)	0.486 (0.25)
		Gastrocnemius lateralis		
	Non-Vibration	52.8±3.7	-	-
	Vibration at 25 Hz	57.5±3.8	1.0 (0.19)	-
	Vibration at 40 Hz	56.2±3.9	1.0 (-0.08)	0.368 (-0.06)
		Global activity		
	Non-Vibration	25.4±1.1	-	-
	Vibration at 25 Hz	26.5±1.0	0.276 (0.22)	-
	Vibration at 40 Hz	26.8±1.2	0.110 (0.26)	1.0 (0.05)

Table 9. *(Continued)*

Exercise phase	Suspended hamstring curl condition	Mean ±SE	*p*-value (effect size)	
			Vs. Non-Vibration	Vs. 25 Hz vibration
			p (ES)	*p (ES)*
		Rectus femoris		
Eccentric	Non-Vibration	1.4±0.2	-	-
	Vibration at 25 Hz	1.5±0.2	1.0 (0.12)	-
	Vibration at 40 Hz	1.8±0.3	0.444 (0.31)	0.930 (0.21)
		Biceps femoris		
	Non-Vibration	22.0±1.4	-	-
	Vibration at 25 Hz	24.5±1.7	0.276 (0.34)	-
	Vibration at 40 Hz	22.6±1.6	1.0 (0.09)	0.600 (-0.25)
		Semitendinosus		
	Non-Vibration	20.6±1.1	-	-
	Vibration at 25 Hz	22.9±1.5	0.201 (0.38)	-
	Vibration at 40 Hz	22.5±1.9	0.389 (0.26)	1.0 (-0.05)
		Gluteus maximus		
	Non-Vibration	10.0±0.8	-	-
	Vibration at 25 Hz	11.7±1.1	0.053 (0.36)	-
	Vibration at 40 Hz	11.4±1.0	0.144 (0.33)	1.0 (-0.06)
		Gastrocnemius medialis		
	Non-Vibration	36.3±2.1	-	-
	Vibration at 25 Hz	37.0±2.2	1.0 (0.07)	-
	Vibration at 40 Hz	37.1±2.2	1.0 (0.09)	1.0 (0.02)
		Gastrocnemius lateralis		
	Non-Vibration	51.5±3.7	-	-
	Vibration at 25 Hz	50.8±3.6	1.0 (-0.04)	-
	Vibration at 40 Hz	51.2±4.4	1.0 (-0.01)	1.0 (0.02)
		Global activity		
	Non-Vibration	23.6±0.9	-	-
	Vibration at 25 Hz	24.7±1.0	0.211 (0.24)	-
	Vibration at 40 Hz	24.4±1.1	0.534 (0.17)	1.0 (-0.06)

Note. Data presented as normalized muscle activity (%MVIC); SE = standard error of the mean; Global activity = mean of the six muscles; † = significantly different with the non-vibration condition;‡ = significantly different with vibration at 40 Hz condition.

Figure 25. *OMNI-Res (Mean± SE) for Suspended Supine Bridge and Suspended Hamstring Curl under Non-Vibration, Vibration at 25 Hz and Vibration at 40 Hz Conditions.*



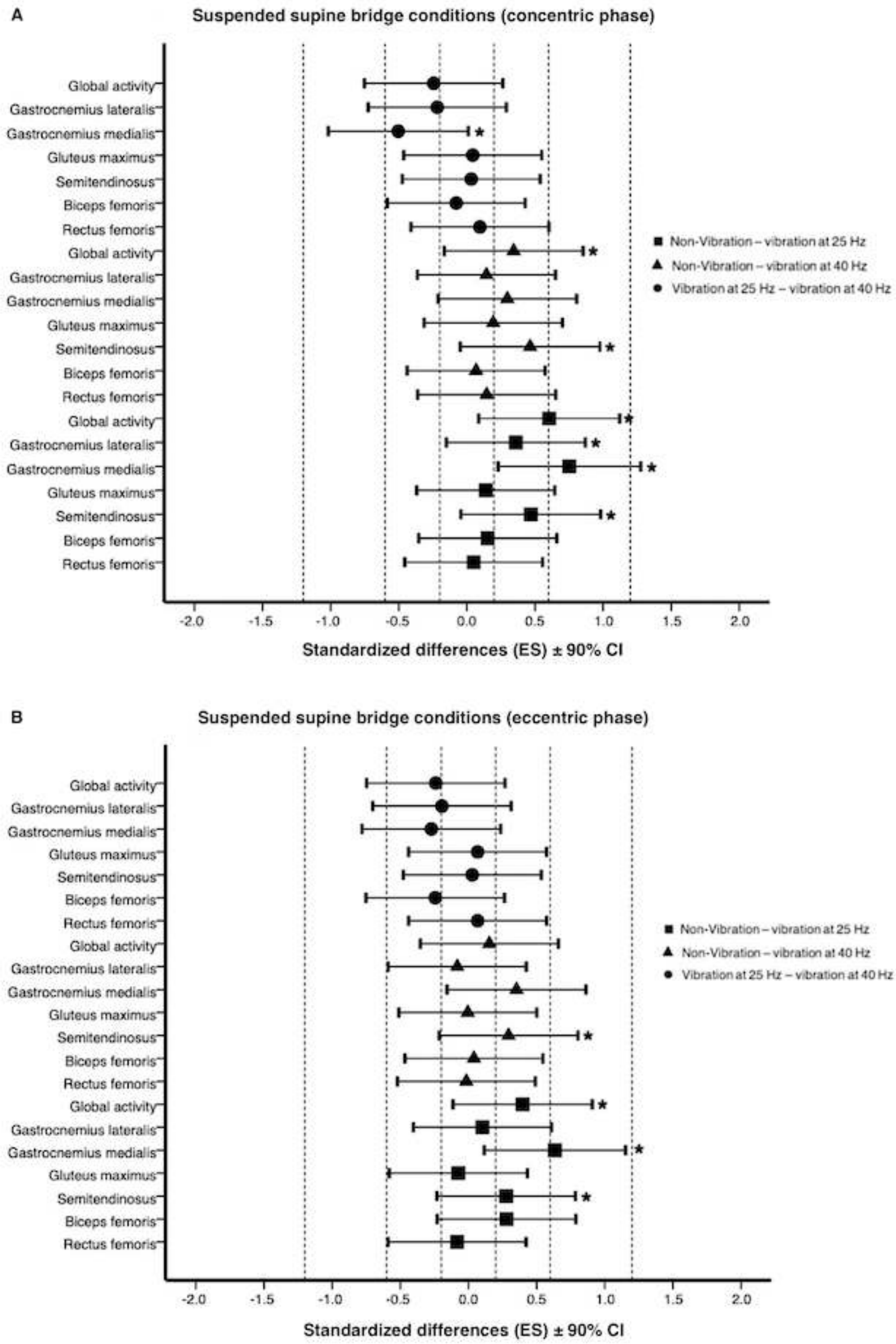
Note. Each bar represents the mean, and the error bar represent the standard error of the mean (SE).

A.U. = Arbitrary units

† = significantly different with non-vibration condition

‡ = significantly different with vibration at 25 Hz condition

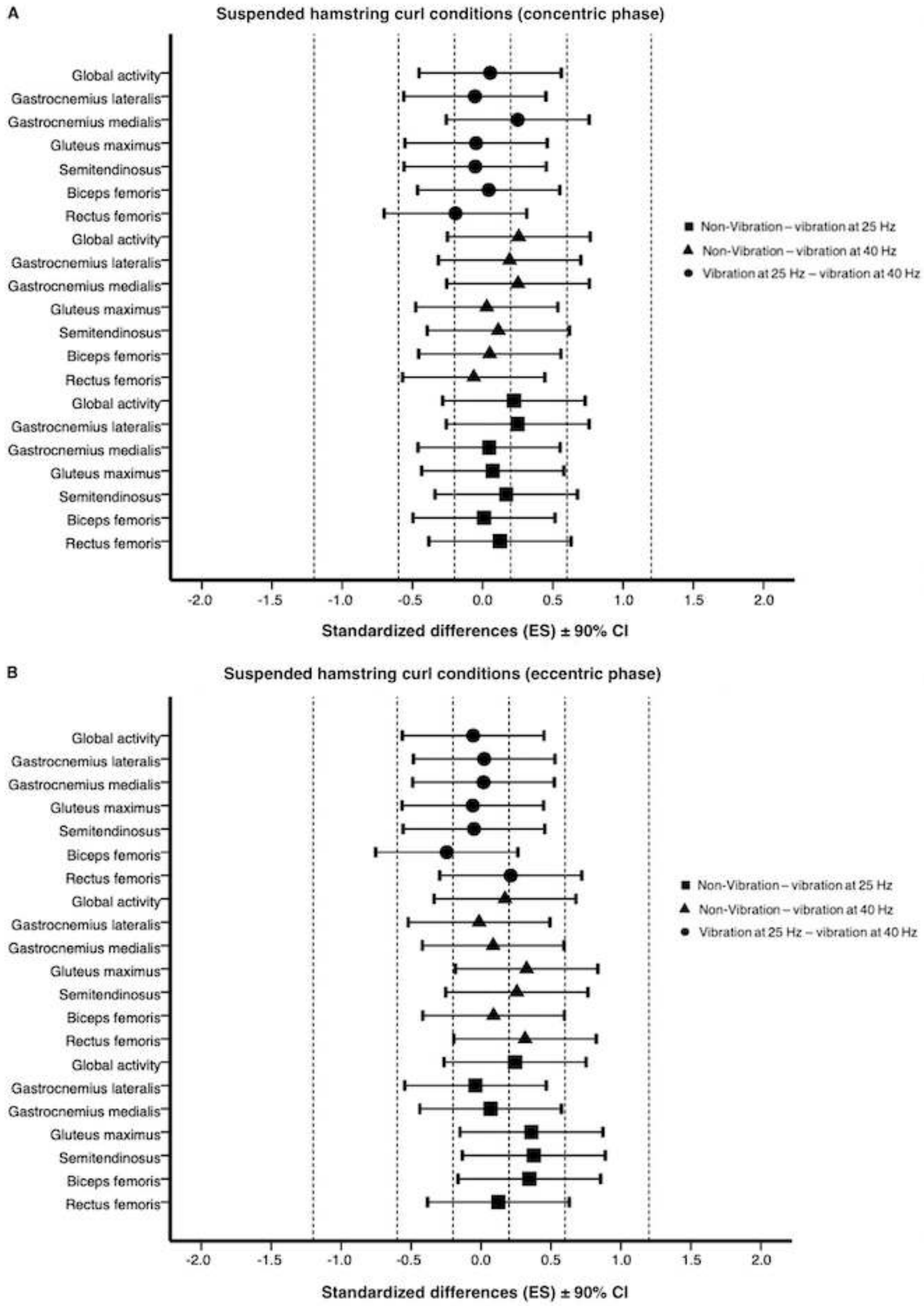
Figure 26. *Effects of Suspended Supine Bridge Conditions on Muscle Activity (%MVIC) at Concentric (A) and Eccentric Phase (B) Expressed as Standardized Differences (Cohen's d) ± 90% CI.*



Note. The dotted line represents the effect size thresholds. ES = effect size; CI = confidence interval.

* Significant differences at $p < 0.05$.

Figure 27. *Effects of Suspended Hamstring Curl Conditions on Muscle Activity (%MVIC) at Concentric (A) and Eccentric Phase (B) Expressed as Standardized Differences (Cohen's d) ± 90% CI.*

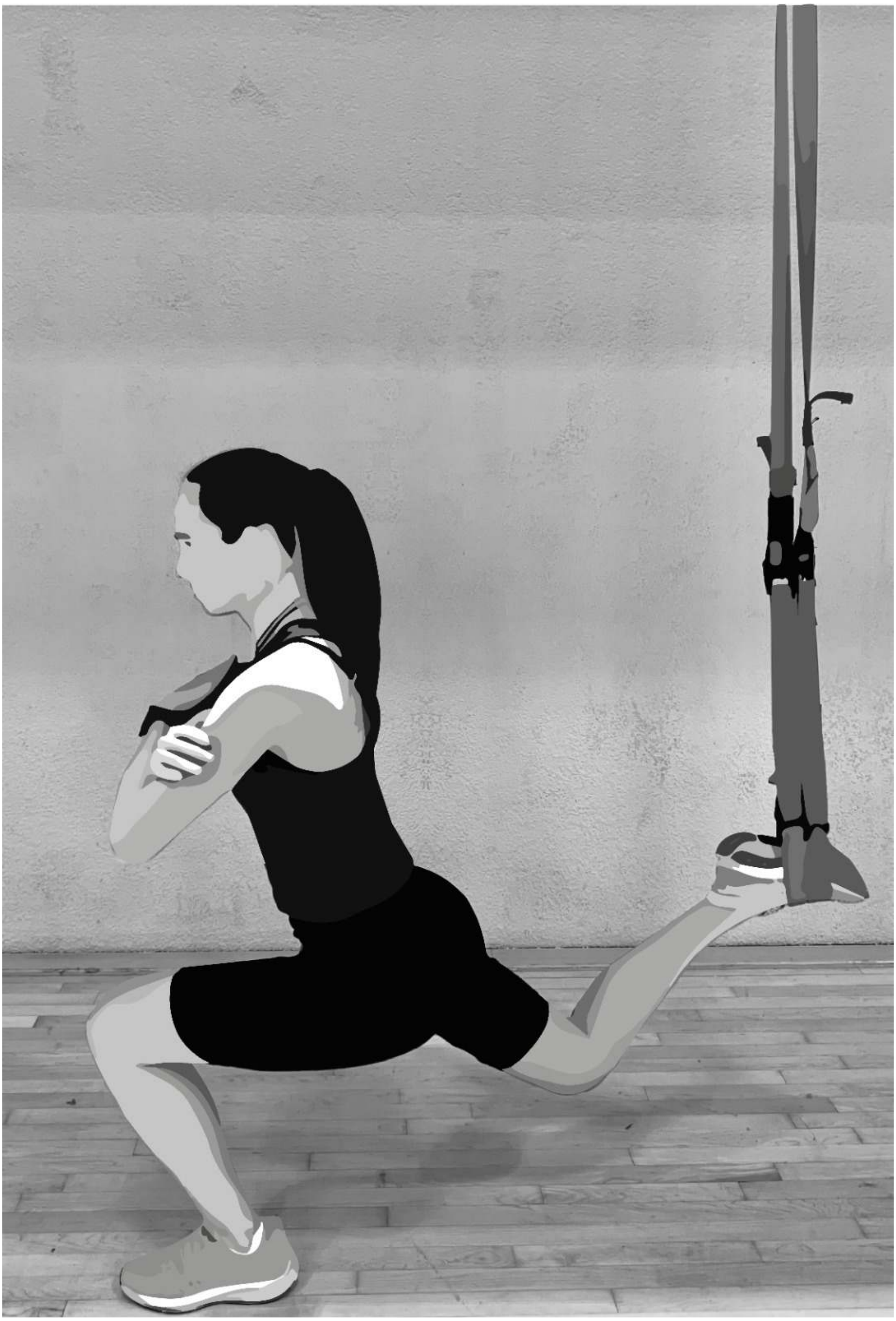


Note. The dotted line represents the effect size thresholds. ES = effect size; CI = confidence interval.

In addition, with the aim of placing superimposed vibration on suspension straps, a vibratory system for suspension training was used in study 6. As a result, the vibratory system provided vibration to the suspension device by converting the rotary motion of an electric motor into a vertical motion, which caused the displacement of a connecting rod with an amplitude of 8 mm (peak to peak), and the motor rotation frequency was regulated with a potentiometer. The prototype is shown in **Figure 28**. The detailed description of the invention and the proof of application for National Patent number 202030652 requested to OEPM can be found in the appendix (Patent document and proof of application for national patent number 202030652).

Figure 28. *Vibratory System for Suspension Training: Innovation Scheme*

This Figure is not shown for confidentiality reasons.



DISCUSSION

The main findings were that the levels of activation when a suspended device is used on the principal and stabilizing musculature, in particular on upper extremity exercises, are similar but some studies show differences in activation for the same muscle group analysed, these differences can be attributed to differences in body position during exercise or other parameters such as range of motion, height of the suspension device, type of the grip or the method used to normalize the electromyographic signal (study 1); for the suspended lunge the position has a significant effect on the values of force exerted on the suspension strap (concentric and average force) and the pace of 80 bpm on the peak force exerted on the suspension strap, the comparison between contraction patterns had an effect on the peak force exerted on the suspension strap and the closest body position (TRX_40-60) reached the significantly higher value of relative force compared to the other isometric suspended lunge conditions (study 2); the recruitment of the front leg musculature did not increase in the suspended lunge and the Bulgarian squat was as demanding as the suspended exercise, the need to introduce an unstable surface or vibration (dual condition) into the front leg to provoke greater muscle activation, the unstable surface (BOSU) significantly decreases the force exerted on the suspension strap and leaning the leg on a suspension strap increases the VGRF of the front leg (study 3); the amount of instability correlates with a positive trend in muscle activation and a negative trend in force exerted on the suspension strap in single and dual instability conditions (suspended lunge conditions) (study 4); the sum of the amplitudes (peak) and the quadratic combination of the acceleration is a useful approximation to quantify the amount of perturbation in unstable environment using an accelerometer, the amount of instability correlates with positive trend with the muscle activation (the highest performance limb) in the loaded free barbell half-squat, the BOSU conditions elicited an increase in the muscle activation of the prime half-squat movers, with the BOSU-down condition being the most unstable surface (study 5); superimposed vibration on a suspension device significantly increased global activity in the suspended supine bridge at frequencies of 25 Hz and 40 Hz compared to the non-vibration condition. The 25 Hz frequency in the suspended supine bridge elicited significantly greater activation than vibration at 40 Hz compared to the non-vibration condition for the more proximal muscles exposed to vibration (gastrocnemius medialis, gastrocnemius lateralis and semitendinosus) but not for the prime movers. For the suspended hamstring curl, non-significant differences were found between muscle activity and exercise condition because the pendulum motion damped the effect of 25 Hz and 40 Hz frequencies. Moreover, superimposed vibration progressively increased the value of OMNI-Res while increasing the vibration frequency (non-vibration, 25 Hz and 40 Hz) in suspended supine bridge and suspended hamstring curl (study 6).

Muscle activation in suspension training: a systematic review (study 1)

Muscle activation during suspended exercises

The activation of pectoralis major in suspended push-ups ranges from moderate (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014) to very high (Snarr & Esco, 2013b). These differences in activation can be explained by the type of suspended device used such as the traditional suspended device (Borreani, Calatayud, Colado, Tella, et al., 2015; Snarr & Esco, 2013b) or the pulley suspended device (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014), or by the height of the suspended device that affected the fundamental principle of the resistance vector, with the trunk-legs inclination or the angle between the body and the floor (Borreani, Calatayud, Colado, Tella, et al., 2015; Melrose & Dawes, 2015). Similarly, the anterior deltoid decreases its activation when the amount of instability produced by the suspension device is high. For abdominal muscles, the use of a pulley suspension device increases the value of the rectus abdominis (Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014) although other muscles such as the external oblique, serratus anterior, and erector spinae are inhibited or decreased in their activation when compared to the traditional suspension device exercise (Calatayud, Borreani, Colado, Martin, & Rogers, 2014). This is because suspension push-ups increase the activation of the latissimus dorsi to stabilise the shoulder joint. Additionally, the use of a suspension device requires the abdominal muscles and latissimus dorsi to be sufficiently activated to achieve mechanical equilibrium around the lower back (Beach et al., 2008). However, triceps brachii showed high or very high activation values regardless of the type of suspension device used (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014; Snarr & Esco, 2013b). As mentioned above, the level of recruitment of the anterior muscles appears to be conditioned by the type of device and the length of the strap (Melrose & Dawes, 2015).

For the suspended inverted row, the type of grip has determined the activation level of the prime movers (latissimus dorsi, middle trapezius, posterior deltoid, and biceps brachii). The pronated and supinated grip significantly influenced the recruitment of prime inverted row movers, especially for posterior deltoid and biceps brachii (Snarr et al., 2014; Snarr & Esco, 2013a) compared to the activation levels reported by McGill et al. (2014b). The prone handgrip could be enhanced the extensor role of the posterior chain musculature. The biceps brachii activity observed by Snarr et al. (2014) differed from the other studies. This could be because they used a supine handgrip, which increases the recruitment of the biceps brachii during elbow flexion. Additionally, the trunk-legs inclination and the hip flexion angle influence the latissimus dorsi and middle trapezius activation. For core muscles (rectus abdominis, external oblique and lumbar multifidus), the inverted row under suspended conditions was insufficient to offer a challenge to the recruitment (<20% MVIC) of the muscles mentioned above, and this suggests that the instability created by traditional suspension devices does not engage the trunk muscles. According to Bettendorf (2010), the modification in the fundamental principles of stability and pendulum lead to differences in the trunk muscles activation (rectus abdominis, external oblique, and erector spinae) in the exercise of suspended prone bridge with hip abduction (Fong et al., 2015; Mok et al., 2014), arms extension (Atkins et al., 2015), suspension of feet or arms (Snarr & Esco, 2014), or suspension of feet and arms (Byrne et al., 2014). Moreover, the suspended hamstring curl technique utilised by Fong et al. (2015) in chronic back pain patients, which entails positioning the supine trunk with a lumbopelvic retroversion, could explain the differences in the activation of trunk muscles (transversus abdominis and internal oblique, rectus abdominis and external oblique) in comparison with Mok et al. (2014).

Muscle activation in suspended exercises compared to traditional exercises

Studies that compared muscle activation during suspension and traditional exercises utilised physically active participants between 15 and 28 years of age. Most studies presented activations as %MVIC; these studies used different procedures to obtain the MVIC. For example, Malliaropoulos et al. (2015) performed three five-second MVICs with an isokinetic dynamometer, whereas other studies performed MVIC trials against a matched resistance (Beach et al., 2008; Borreani, Calatayud, Colado, Moya-Nájera, et al., 2015; Byrne et al., 2014), with trial lengths ranging from three to ten seconds (Atkins et al., 2015; Byrne et al., 2014; Mok et al., 2014). The MVIC trials for each analysed muscle differed by the protocol used (e.g. the protocols described by Konrad (2006) compared to Escamilla et al. (2010).

The reviewed studies show a higher activation in suspended push-ups than in traditional push-ups, however the type of suspension device as the absence of a stabilising loop will significantly improve the activation of the pectoralis major (Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014) but an increase in instability such as a pulley suspension device will inhibit the activation of the pectoralis major and anterior deltoid (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014) while significantly increasing the activation of the triceps brachii, because acts as a stabiliser under unstable conditions, especially when instability is lateral (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014; Snarr & Esco, 2013b), and the core muscles (rectus abdominis, internal oblique, and external oblique), particularly the rectus abdominis because acts as a grater stabiliser during suspended push-ups compared to traditional push-ups (Beach et al., 2008; Calatayud, Borreani, Colado, Martin, & Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014; McGill et al., 2014a; Snarr et al., 2013). The lateral instability seems to have caused an inhibitory effect on the activation of the anterior serratus, and although there is no consensus on the activation levels recorded in the different studies analyzed, the variations in the suspension device height suggested that the higher the position of the suspension device, the greater the activation of the serratus anterior (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, &

Rogers, 2014; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014; Calatayud, Borreani, Colado, Martin, Rogers, et al., 2014).

Variations in the grip type used in studies investigating inverted row exercise make it challenging to compare muscle activations. Along with the grip type, variations in trunk-legs inclination and the hip flexion angle result in variations in the activation of the involved muscles (latissimus dorsi, middle trapezius, posterior deltoid, and biceps brachii), which cause significant differences in activations between suspended inverted row and inverted row (McGill et al., 2014b; Snarr et al., 2014; Snarr & Esco, 2013a). It is difficult to tell if this difference is due to the addition of instability to the exercise. For the suspended prone bridge, instability of the arms may increase the difficulty of the exercise more than the instability of the legs, which may cause higher rectus abdominis activation (Byrne et al., 2014). In contrast, the activation of trunk muscles reported by Atkins et al. (2015) differs from other studies because they studied elite swimmers with particular neuromuscular characteristics. Finally, although there is little evidence in the literature, the hamstring musculature was more activated during the suspended hamstring curl than other bilateral hamstring exercises (Malliaropoulos et al., 2015).

In this review, most of the references are from 2013 or later, which indicates that the study of muscle activation during suspension training exercises is a new topic in strength and conditioning. The manufacturer of the most popular suspension devices patented its first device in 2006. There were certain variations in the execution of the exercises, the muscles assessed for activation, the methods used to assess muscle activity and the EMG signal normalisation procedures. As a result, readers should be cautious in interpreting the magnitudes of muscle activations between the same muscles and amongst exercises. Moreover, the participants in many of the studies were physically active male non-athletes, which makes it difficult to generalise findings to both females and athletes.

Suspended lunge exercise: assessment of forces in different positions and paces (study 2)

From the four studied positions in the suspended lunge, the position that generated the highest force exerted on the suspension strap for all the analysed variables (concentric, eccentric, average and peak force variables) was the TRX_60-80 compared to the TRX_40-60. This finding suggests that the position of the suspended lunge that appears to be more unstable (TRX_60-80) will require greater recruitment of motor units due to the instability generated by the suspension device and body position in consensus with the fundamental principle of the resistance and stability vector (Bettendorf, 2010; Maté-Muñoz et al., 2014). Hence, the force exerted on the suspension strap reported by Gulmez (2017) and Melrose and Dawes (2015) during the suspended push-up and the suspended inverted row it gradually increased from the less unstable position (trunk-inclination far to the floor) to the more unstable position (trunk-inclination closer to the floor).

Related to the variation of the position in length and distance, the modification of the strap length in 20 cm between positions (from 40 cm to 60 cm) was not sufficiently demanding to cause significant changes in the force exerted on the strap (in any variable), although for upper body exercises, such as suspended push-up, the muscle activations analysed were greater in suspension strap lengths very close to the ground (10 cm) than in lengths far from the ground (65 cm) (Borreani, Calatayud, Colado, Tella, et al., 2015; Calatayud, Borreani, Colado, Martin, Batalha, et al., 2014), in the same way, that a succession of increases in the length of the strap (from 178 to 238 cm) increased the force exerted on the suspension strap (from 40% to 70%) in the suspended push-up (Giancotti et al., 2018). On the other hand, the variation on feet distance (60 cm to 80 cm) significantly increased the force exerted on the suspension strap in average force for TRX_40-80 compared to TRX_40-40 and in concentric and average force for TRX_60-80 compared to and TRX_60-80. This finding follows Melrose and Dawes (2015), who indicated a higher force exerted on the suspension strap when increasing the distance between the anchor point of the suspension device and the feet fulcrum (3 to 6 increments of 30.5 cm) in the suspended inverted row. These results suggest that an increment in the feet distance also increases force production when performing a suspended lunge.

Regarding the frequency, the positions with the highest feet distance (TRX_40-80 and TRX_60-80) also revealed a higher peak force exerted on the suspension strap at the highest frequencies (80 bpm and 70 bpm, respectively) compared to 60 bpm. Furthermore, the results show a trend towards improvement in the force production at 70 bpm compared to 60 bpm. These differences may be explained by the need to apply more force on the suspension devices. Stability needs to increase at the same time as the frequency of the movement also does it. Peak force values obtained between 70 bpm and 80 bpm are very similar at TRX_60-80. Although insufficient evidence exists to make a comparison, this finding agrees with Jakobsen et al. (2013),who reported that performing lunges under ballistic conditions elicited higher muscle activity than in slowly controlled contractions.

The comparison of peak force exerted on the suspension strap between contraction types and positions showed a significantly higher peak force during dynamic contraction in all positions. Despite the lack of studies analysing the above variables in the suspended lunge, the literature shows that high speeds are associated with high levels of muscle activation (Jakobsen et al., 2013; Jönhagen et al., 2009), and low levels of muscle activity are often caused by isometric contractions (Ekstrom et al., 2007) in lunge exercise. It appears that dynamic contraction in comparison with isometric leads to peak force production improvements because an increase of the pace could boost the recruitment of the motor unit, thus increasing muscle activity.

The percentage of body mass resistance (relative force exerted on the suspension strap) was 9.93% higher in the less unstable position (closest position TRX_40-60) than in the other positions. This may be because the TRX_40-60 position provides greater support on the suspension device handles than other positions, and other positions require more force on the forward foot than on the suspended foot, probably due to increases in strap height and feet distance. In contrast, previous results for isometric suspended exercises (push-up and inverted row) reported higher values of relative force exerted on the suspension strap in the most unstable position, which means in those positions where the body angle is closer from the floor (Giancotti et al., 2018; Gulmez, 2017; Melrose & Dawes, 2015).

There were some limitations associated with this study. Firstly, only the strap length variations were established to modify the degree of instability from the suspension device. No comparison between different strap angles was conducted. Another limitation of our data was the lack of quantification of force on the forward leg compared to the rear leg (suspended). Finally, another limitation may be the no normalized distance forward step during lunge execution, as Boudreau et al. (2009) recommended. However, the thigh and leg length were measured following the ISAK (2001) ensuring the participants' homogeneity.

Muscle activity of Bulgarian squat. Effects of additional vibration, suspension, and unstable surface (study 3)

The instability in the rear leg in the suspended lunge did not cause an increase in muscle activity in the front leg, although it presented a slightly higher but non-significant activation for the rectus femoris compared to the Bulgarian squat. The same recruitment patterns for the rectus femoris were found by Krause et al. (2018), who reported non-significant differences in the activation of the rectus femoris in the standard lunge compared to that in the suspended lunge (22.1 ± 22.2 %MVIC vs 24.5 ± 22.0 %MVIC, p = 0.434). It seems that performing a unilateral lower limb exercise with a suspension strap on the rear leg causes higher demands for the rectus femoris. This is because the primary role of the rectus femoris in the Bulgarian squat and suspended lunge could be the control of the hip flexion and knee extension movements, instead of stabilizing the abduction, adduction, and rotational movements of the hip and pelvis (Krause et al., 2018). For the other muscles of the front leg, activation was moderate (biceps femoris), high (gluteus medius) and very high (vastus medialis and vastus lateralis) for both the Bulgarian squat and the suspended lunge. Similar results were reported by Mausehund et al. (2019) and DeForest (2014) for the exercise of loaded Bulgarian squat in the muscles mentioned above. In contrast, Krause et al. (2018) reported that the suspended lunge increases significantly the muscle recruitment for the hamstring and gluteus medius (13.1 ± 20.1 % MVIC; 24.1 ± 15.1 %MVIC, respectively) compared to a standard lunge (hamstring: 8.7 ± 13.2 %MVIC, p = 0.01; gluteus medius: 15.3 ± 11.4% MVIC, p = 0.01). This suggests that the exercise technique could influence the level of activation of the hip and thigh muscles. As a result, performing a Bulgarian squat with the front leg on the floor demands a higher hip and thigh muscle recruitment than a standard lunge (V. Andersen et al., 2014; Boudreau et al., 2009; Ekstrom et al., 2007; Mausehund et al., 2019) and therefore, the difference in the muscle activation between the traditional and suspended exercises is higher in case of a standard lunge than the Bulgarian squat. Furthermore, leaning the rear leg on the suspension strap appears to decrease the recruitment of these muscles.

Regarding the dual condition, the two conditions eliciting higher activation of the rectus femoris and gluteus medius in the front leg were suspended lunge-BOSU (45.30 ± 4.28 %MVIC; 65.67 ± 4.85 %MVIC, respectively) and suspended lunge-Vibro40 (44.90 ± 5.72 %MVIC; 65.59 ± 4.98 %MVIC, respectively). For these muscles, the stimulus provoked by the BOSU® conditions could be equivalent, in terms of muscle activation, with those offered by the WBV platform at 40 Hz-high, but not at 30 Hz-high. Pollock et al. (2010) found that in healthy participants standing on a WBV platform at 30 Hz of frequency and 5.5 mm of amplitude, the rectus femoris recruitment was significantly higher than when WBV was set at 5 Hz of the frequency the same amplitude. These authors indicated that muscle recruitment for the rectus femoris depends on the frequency and amplitude of vibration. This finding suggests that dual conditions with WBV and compliant environments compromised postural stability, increasing muscle tuning mechanisms and muscle contraction (Choi & Kang, 2013; Marín & Hazell, 2014).

On the other hand, the role of gluteus medius as a stabilizer of the body during the dynamic flexo-extension of the front leg under suspended-BOSU condition and the ability to absorb the vibration offered by the vibration plate might be caused by the contribution of multiple neural pathways with distinct functional roles to rapid motor control response to a perturbation (Shemmell et al., 2010). Furthermore, the reflex motor response during the BOSU condition and the vibratory tonic reflex on the WBV platforms might induce similar activation in the involved muscles. This finding, also reflected in the global activation (the mean of all analysed muscles), might be explained by the specific requirements of absorbing the vibration or maintaining the stability on a BOSU®. Hence, performing dynamic tasks on a BOSU®, subjects experiment muscular trembling (micro amplitude changes), provoked by body mass variations projected on the forward leg, leaned on a compliant surface like this during the whole range of movement. These micro amplitude changes are described as one of the muscle tuning mechanisms for vibration training (Cardinale & Bosco, 2003).

The percentage of body mass resistance exerted by the rear leg on the suspension strap was significantly lower for suspended lunge-BOSU than suspended lunge-Vibro30 and lower than suspended lunge and suspended lunge-Vibro40. This finding suggests that force exerted on the suspension strap could not be influenced by the front leg lean, but leaning the front leg under unstable conditions (BOSU) provokes an increase in the amount of instability and the force production decreases (Behm et al., 2002). Likewise, Saeterbakken & Fimland (2013) reported that performing a squat under unstable conditions, BOSU, significantly decreased the force produced (603 ± 208 N) compared to the stable squat on the floor (749 ± 222 N) or less unstable surfaces as squats on the power board (694 ± 220 N). On the other hand, the VGRF was significantly higher for suspended lunge than for the Bulgarian squat (113.01% ± 9.24 vs 108.65% ± 9.05, p = 0.006). Thus, leaning the rear leg on a suspension strap provokes a transfer of a certain amount of body mass resistance towards the front leg, maintaining the trunk position, which exerts a force on the ground to attempt to keep the posture. Moreover, the low activation of rectus femoris of the rear leg could explain that maintaining the rear leg on a suspension device inhibits the role of the rectus femoris as a hip flexor and contribute to the increase of the VGRF in the front leg.

There were some limitations associated with this study. Results of the present study may be influenced by subjects' experiences with similar exercises to those performed in the present investigation. Each individual has a different level of motor control for the same task, which might be considered when assessing muscle electrical signals. Therefore, participants' characteristics might constitute a limitation to infer the results of the present study. This study did not use functional tests to determine participants' laterality, together with their neuromuscular and performance levels.

Moreover the lack of quantification about the amount of instability produced by the device should be considered. Another limitation may be that a goniometer did not control the knee flexion angle. However, the displacement during each repetition of the Bulgarian squat and suspended lunge conditions was measured with a positional encoder.

Correlational data concerning body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions (study 4)

For this study, the use of the accelerometer and its relationship with muscular activity and the force exerted on the suspension strap can be justified based on the study conducted by Barbado et al. (2018) showed a moderate to high correlation (r = 0.520 - 0.810) between the values of acceleration and the displacement of the COP in different variations of the trunk stabilisation exercise, as well as a moderate to high reliability (ICC: 0.71-0.88) in the values obtained by the accelerometer in the different conditions of the trunk stabilisation exercises. According to data from Barbado et al. (2018) it is suggested that the accelerometer could be used to quantify the level of disturbance caused by an unstable environment.

The performance of a Bulgarian squat in single instability (suspended lunge) and dual instability (suspended lunge-Foam, -BOSU up, and -BOSU down) conditions obtained a significant Pearson's correlation with a moderate to very large effect between muscle activation (vastus medialis, vastus lateralis, gluteus maximus, gluteus medius, and biceps femoris) and the BCMA (r = 0. 393 to r = 0.826, p = 0.008 - 0.000), except for the rectus femoris. However, no previous studies associate muscle activation with variations in the BCMA as an indicator of the amount of perturbation produced by the unstable environment (unstable surfaces and suspension devices). This finding suggests that regardless of if the perturbation is generated in the rear leg, in the front leg or both, rear and front leg, the displacement of the body centre of mass, caused by an increase in instability, has a linear relationship with the increase in muscle recruitment of the front leg in a unilateral exercise such as the suspended lunge. Previous studies, in upper extremity exercises such as bench press and push-ups, showed a significant linear effect between the amount of instability (stable, single and dual instability) and the level of muscle activity (G. Anderson et al., 2013; Norwood et al., 2007), although the level of disturbance was established by the increase of instability and not through the magnitude of perturbation produced by the condition of the exercise measured by accelerometer.

Another finding was that the force exerted on the suspension strap (percentage of body mass resistance) revealed a Pearson's correlation with the BCMA ($r = -0.595$ to $r = 0.797$, $p = 0.001$- 0.000) with a large to very large effect in the different conditions of the suspended lunge. This suggests that whether the source of instability is single (suspended lunge) or dual (suspended lunge-Foam, -BOSU up, -BOSU down), the increase in the magnitude of the perturbation (variation in the BCMA) decreases the force production. This finding reinforces the argument of Behm et al. (2002) that the degree of stability or instability affects limb force production directly, and provides quantitatively with the accelerometer the magnitude of the degree of instability. Likewise, other studies (Moras et al., 2019; Saeterbakken & Fimland, 2013) obtained lower values of force production on unstable surfaces (BOSU and power board) or dual condition (inertial flywheel device and Pielasters) compared to the squat under stable conditions. Although the previous studies did not correlate force production with the magnitude of instability, the data obtained from the decrease in force on unstable surfaces could reinforce the relationship between the BCMA and force production in an unstable environment. On the other hand, Moras and Vàzquez-Guerrero (2015) used an accelerometer to quantify the instability offered by the Pielasters in the squat exercise, although it did not correlate the acceleration values with the force production levels; however the force values were lower in the squat under unstable than in stable conditions.

he study's main limitation was that the electromyographic signal was not normalized, expressed in % MVIC, and the signal was given in root mean square, so readers should be cautious in interpreting the correlations obtained between muscle activation and the BCMA for the same muscle and amongst exercises. Another limitation was that the force assessment was performed only in the suspended leg (rear leg) for all exercise conditions, with no assessment of the VGRF for a complete analysis of the relationship between force and the BCMA during different stability conditions in the suspended lunge, where the variation of the perturbance was carried out in the forward leg (floor, foam, BOSU-up and - down).

Influence of the amount of instability on the leg muscle activity during a loaded free barbell half-squat (study 5)

The amount of instability during the half-squats under different stability conditions was measured with an accelerometer. The proposed methodology was the sum of the peaks, considering the quadratic combination of the acceleration in anteroposterior and mid-lateral axes (Moras & Vázquez-Guerrero, 2015), seems to provide an accurate approach for quantifying the amount of instability (BCMA) in different unstable resistance training environments (Romero-Franco et al., 2013). Previous studies (Barbado et al., 2018; Vázquez-Guerrero et al., 2016) suggested that the use of mean accelerations values might not be the best way to describe the amount of instability due to mean or peak root mean square acceleration values do not reflect the ability to maintain the posture, because the moments when the participants are balanced are taken into consideration for the calculations (Thiel et al., 2014). The quantification of instability revealed that the BOSU-down condition provided a significantly higher BCMA than the other conditions. Furthermore, the BCMA was higher in the BOSU-up than in floor and foam (stable conditions). By Saeterbakken and Fimland (2013) criteria to establish the magnitude of instability (unstable dimensions and magnitude of contact with the floor).

The exercise condition obtained a main effect on the global muscle activity, the ratio of perception of exertion, and the BCMA. The global activity was significantly higher on the BOSU than on the floor and foam. In this regard, several studies (V. Andersen et al., 2014; Drinkwater et al., 2007) found that performing a loaded squat on a foam pad did not significantly increase muscular activation compared to the exercise performed on the floor. According to the studies mentioned earlier, the present results showed that the inclusion of foam pads during a squat might not be worthwhile for high-standard athletes, at least for increasing the activity of the knee extensor muscles. Furthermore, the use of high loads seemed to play a stabilizing role under unstable conditions, allowing higher muscle activation and, therefore, higher force production (Drinkwater et al., 2007; Saeterbakken & Fimland, 2013). Concerning BOSU conditions, muscle activation values (vastus medialis, lateralis, biceps femoris, and global activity) and BCMA were significantly higher than both stable and foam conditions. However, McBride et al. (2006) reported lower activation values in the vastus medialis and lateralis in the squat exercise on Dyna disc. Likewise, Saeterbakken and Fimland (2013) found no differences in different lower limb muscles when comparing squats on the floor with other unstable surfaces (Power Ball, BOSU, or Balance cone). It could be speculated that the tendency to avoid the dynamic knee valgus explains the high activation of the vastus medialis on BOSU conditions. Indeed, although the BOSU-down condition created higher global instability, it offered a flat and rigid surface that compelled the participants to act differently in avoiding the knee valgus position. Although this study did not test this muscle, the role of the gluteus medius in stabilizing the posture can probably explain the lower activation of the vastus medialis in the BOSU-down actions (Aguilera-Castells et al., 2019; Krause et al., 2009; Mausehund et al., 2019). Overall, it seems that the effects of BOSU on muscle activation are not clear.

On the other hand, the role of the biceps femoris co-contraction in the most unstable conditions seemed to be clear in a half-squat, even if previous studies (V. Andersen et al., 2014; McBride et al., 2010; Wahl & Behm, 2008) demonstrated a lower activation on the squat exercise under unstable conditions in comparison with more stable conditions. The main reason for the high activation of the biceps femoris on BOSU conditions is the high anteroposterior instability created by the unstable surface. Only the study conducted by Saeterbakken and Fimland (2013) follows the results obtained, showing a higher percentage of biceps femoris activation under the BOSU condition than the stable condition. Furthermore, the athletes' experience in the present study and their ability to maintain balance, even in the most perturbed conditions, might explain these differences. The contemporary trend of introducing unstable environments in training programs for experienced athletes might change the inhibiting effect of instability on the primary squat movers and become a challenge for intramuscular coordination in highly trained and coordinated populations. Thus, using unstable resistance training exercises would force accommodation to an unstable environment, diminishing the loss of force and the extent of co-contractions (Anderson & Behm, 2005).

The OMNI-Res was significantly higher for BOSU conditions than the stable and less unstable surface (foam). Even though the validity of the OMNI-Res has been demonstrated in terms of metabolic resistance training (Robertson et al., 2003) and velocity-based training (Naclerio & Larumbe-Zabala, 2017), no research has studied the relationship between the amount of instability and muscle activity. This study shows a strong similarity between OMNI-RES and muscle activation under different half-squat conditions. However, for the comparison between OMNI-Res and BCMA, the OMNI-

Res values were very similar between BOSU conditions, but the BOSU-down obtained a significantly higher BCMA than the rest of half-squat conditions. This finding is consistent with Brown et al. (2014), which despite not quantifying the amount of instability, reported significantly greater OMNI-Res values for the stability ball bench press than for traditional exercise. Conversely, Panza et al. (2014) obtained non-significant differences for the OMNI-Res while performing a bench press under unstable (stability ball) and stable conditions. This would suggest that the relationship between OMNI-Res and the resistance training exercises performed on unstable surfaces is not clear, and further research about OMNI-Res, unstable surfaces and the quantification of instability is needed.

There are several limitations to the present study. The characteristics of the sample, demonstrating high neuromuscular performance, prevents extrapolation of the results to the general population. Although the statistical power is acceptable, the sample size was also limited, as too was the number of analysed muscles. Moreover, the present study was conducted using dynamic half-squats at 60 bpm. This controlled pace allowed an efficient and balanced execution, but the present results cannot be generalized to other rhythms and, of course, other motor skills. Additionally, only the BCMA has been considered, but no acceleration measurements were obtained from other body parts such as the knee or the ankle. The data processing still requires developing a proper algorithm for obtaining the BCMA in real-time, while executing the movements.

sEMG activity in superimposed vibration on suspended supine bridge and hamstring curl (study 6)

Superimposed vibration in a suspension device increased lower limb muscle activity in the supine bridge but not in the hamstring curl exercise. In the suspended supine bridge, a significant moderate increase of 14.8% (concentric phase) and a small increase of 9.7% (eccentric phase) was found under the 25 Hz vibration condition compared to the non-vibration global activity. Likewise, 40 Hz vibration significantly increased global activation by 8.7% (a small increase) during the concentric phase. Similarly, Marín and Hazell (2014) applied superimposed 30 Hz vibration on an unstable surface (BOSU) and found a higher muscle activity between 23.5% and 35% in the isometric half-squat compared to the unstable condition. The effect of additional vibration (30 Hz and 40 Hz with an amplitude of 4mm) on unstable surfaces and suspension devices increased the exercise demands. Thus, eliciting a greater activation of the lower limb muscles (vastus medialis and lateralis, biceps femoris, and gluteus medius) during the suspended lunge combined with 40 Hz WBV than in unstable or suspended exercises without vibration (Aguilera-Castells et al., 2019).

The effect of two different frequencies was studied in the present study, finding a small to moderate significant increase in semitendinosus, gastrocnemius medialis, and lateralis activation under 25 Hz vibration compared to the non-vibration condition. Likewise, there was a significantly small decrease in the gastrocnemius medialis activity at 40 Hz. Overall, this study showed that performing the 25 Hz suspended supine bridge elicits a greater activation than at 40 Hz vibration in almost all the analysed muscles. In the same vein, a progressive increase in vibration frequency (5 Hz to 30 Hz) gradually enhanced the neuromuscular response for the lower limb muscles (soleus, gastrocnemius, tibialis anterior, biceps femoris, vastus medialis, and rectus femoris), achieving the highest activations at 25 to 30 Hz frequencies (Ritzmann et al., 2013). On the other hand, 25 Hz vibration was consistently more demanding than 40 Hz vibration, per Cardinale and Lim (2003), who found lower but not significant muscle activity of 40 Hz vibration compared to 30 Hz.

Regarding the effect of the different frequencies on the analysed muscles, higher activation was found for the more proximal muscles exposed to the vibration. The additional effect of vibration at 25 Hz compared to the non-vibration suspended condition was significantly higher for the gastrocnemius (medialis and lateralis) and semitendinosus in the concentric and eccentric phase (from 9.8% to 23.8% with trivial to moderate effect). Previous studies also demonstrated that the more proximal to the vibration experimented higher activities than the more distal muscles (Hazell et al., 2010; Ritzmann et al., 2013). In this regard, the present study showed that in both vibration conditions (25 Hz and 40 Hz), the muscle excitation sequence (Neto et al., 2019), from higher to lower activation, was gastrocnemius lateralis, gastrocnemius medialis, semitendinosus, biceps femoris, gluteus maximus, and rectus femoris. Thus, the magnitude of the neuromuscular response to the vibratory stimulus in those muscles closer to the most proximal joints (ankles) dissipates the effects of vibration for the more distal muscles, acting as a damper (Abercromby et al., 2007b). Indeed, the vibration induces different reflexes that favour increased muscle activation on the most proximal muscles, such as the tonic vibration reflex (Issurin, 2005; Ritzmann et al., 2010) or the stretch reflex on the soft tissues (Cardinale & Lim, 2003; Cochrane et al., 2009).

Of all analysed muscles, gastrocnemius lateralis (41-60% MVIC) achieved a high activation under 25 Hz vibration and slightly lower (37.4 % MVIC) for gastrocnemius medialis. Participants were asked to perform an ankle plantar flexion on the strap cradles in the suspended supine bridge. Although the feet remained in plantar flexion in the three suspended supine bridge conditions in the current study, the percentage of gastrocnemius activity significantly increased (14-23%, from small to moderate increase) under 25 Hz vibration to the non-vibration condition. Similarly, Ritzmann et al. (2013) found that the gastrocnemius medialis activity increased up to 48% between forefoot stance and normal stance foot position on the vibration platform. The lack of differences between the 40 Hz vibration and the non-vibration suspended condition could be explained because gastrocnemius is more predominantly activated at frequencies below 40 Hz (20, 25, and 30 Hz) (Di Giminiani et al., 2013), according to the findings of the present study.

The hamstrings (biceps femoris and semitendinosus) muscle activity ranged from moderate to low (<24% MVIC), with significant differences in semitendinosus activity at 25 Hz and 40 Hz in comparison to the non-vibration condition. However, following Abercromby et al. (2007a), the biceps femoris activity was slightly lower, with similar activation in all conditions. The biceps femoris's low activation (<21% MVIC) is related to 90° knee flexion in the suspended supine bridge. Ho et al. (2020) found a similar low activation (18% MVIC) of the biceps femoris in the dynamic supine bridge (90° knee flexion). However, the effect of WBV in the static supine bridge, maintaining the 90° of knee flexion, elicited a significant moderate activation (21-40 % MVIC) of the biceps femoris at 30 Hz and 50 Hz, although the non-vibration condition also showed a moderate level of activation (27 % MVIC). The authors supported that 50 Hz vibration was more demanding for the biceps femoris in the static supine bridge (Marín & Cochrane, 2021). Similarly, Hazell et al. (2007) found an increase in biceps femoris activation between 35 Hz and 45 Hz for dynamic and static squats. This suggested that superimposed vibration (25 Hz and 40 Hz) in the dynamic suspended supine bridge is insufficient to significantly stimulate the biceps femoris compared to the non-vibration condition significantly. Likewise, there could be several reasons for the small differences between the biceps femoris and semitendinosus in the suspended supine bridge. One reason is that the suspended exercise produces lateral instability, provoking a lateral rotation of the thighs and, consequently, an increased semitendinosus activity because of its role in counteracting this movement (Tobey & Mike, 2018). Furthermore, the amplitude of the vibrating machine (8 mm, peak to peak) is suggested to provoke more horizontal oscillations and focus on the stabilizing structures that, in the present study, are stabilized by the semitendinosus (D. P. Cook et al., 2011). Another reason is that the necessity to keep the feet stable and maintain the anchor in a plumb line (perpendicular to the ground) of the suspension strap requires the participation of the posterior thigh muscles, similar to the feet-away hip thrust (Collazo García et al., 2020). This semi-stretched position provokes increased muscle tension and enhances the effects of the vibration in the hamstrings muscles (Cardinale & Lim, 2003; Marín & Cochrane, 2021).

Although the barbell hip thrust is a very demanding exercise for gluteus maximus (>60% MVIC) (Neto et al., 2019), the variation of suspended (and unloaded) exercise proposed in this study elicited low activation (<23 % MVIC) with a trivial and small effect among conditions. In this vein, previous studies have reported activation levels ranging from moderate to low (<25 % MVIC) for gluteus maximus in unloaded supine bridge on the floor (Ekstrom et al., 2007; Jang et al., 2013; Kim & Park, 2016). Thus, it appears that the suspended supine bridge (with an additional effect of vibration) is as demanding for the gluteus maximus as the traditional supine bridge exercise and are not sufficiently challenged to reach high and very high activation values (>40% MVIC) in the gluteus maximus, as happens with the single-leg bridge (Ekstrom et al., 2007; Lehecka et al., 2017), the WBV supine bridge (Marín & Cochrane, 2021) or the barbell hip thrust (Andersen et al., 2018; Contreras et al., 2016b; Williams et al., 2021). Therefore, although the gluteus maximus is the prime supine bridge mover, its activation is still low. Moreover, superimposed vibrations were dampened by the more proximal to vibration musculature, and the gluteus maximus were not overstimulated. In addition, the rectus femoris showed the lowest activation (<2.0% MVIC) with a trivial effect in both phases of exercise without significant differences among conditions. Collazo García et al. (2020) showed a significantly (2.4%) lower rectus femoris activation in the feet-away barbell hip thrust (3.4% MVIC) compared to the original hip thrust condition (5.8 % MVIC).

Likewise, Lehecka et al. (2017) found similar rectus femoris activity in the unloaded single-leg bridge with 90° of knee flexion, agreeing with the present study results.

Conversely, as hypothesized, the additional effect of the superimposed vibration did not result in a significantly higher activation in any of the analysed muscles, or the global activity, during the concentric and eccentric phases of the suspended hamstring curl. Moreover, differences among exercise conditions ranged from trivial to small. Even though the muscle excitation sequence was like the suspended supine bridge, the main difference in transmitting the vibration between the suspended supine bridge and the suspended hamstring curl was the suspension strap position. The straps remained in a plumb line in the supine bridge, acting as a pendulum in the suspended hamstring curl. Several studies suggested that vibration transmission via cable in pulley exercises such as biceps curl or one arm pulleying keep the perpendicular between the anchor point, vibration device, and handle to enhance the effects of local vibration (Bosco et al., 1999; Issurin et al., 2012; Issurin & Tenenbaum, 1999). Nevertheless, the pendulum motion in the suspended hamstring curl could attenuate vibration transmission because the vibratory system is designed to transmit the vibration. Moreover, it could be speculated that the pendulum motion could also exert a dampening effect by inhibiting the tonic vibratory reflex (Rittweger, 2010). On the other hand, the pendulum motion and plantar flexion to keep the feet on the cradles could explain the gastrocnemius activity in the suspended hamstring curl conditions. Additionally, Bettendorf (2010) suggested that the intensity variation in a suspended exercise is based on three fundamental principles. Thus, the pendulum principle could justify that the prime mover activations (biceps femoris and semitendinosus) in this study were slightly higher than low activations (<21% MVIC) reported by Árnason et al. (2014) in the suspended hamstring curl without pendulum movement and lower than high and very high activations (>50% MVIC) registered by

Malliaropoulos et al. (2015) in the suspended hamstring curl with alternating knee flexion and pendulum motion.

Regarding OMNI-Res, the finding was that superimposed vibration increased the value of subjective perception of exertion compared to the non-vibration suspended condition around 10% (small increase) for both vibration frequencies in the suspended hamstring curl and from 18% to 32% (small to moderate increase) for the suspended supine bridge. Thus, it seems that the value of OMNI-Res increases progressively while increasing the vibration frequency, being consistent with the significant correlation ($r = 0.95$) between OMNI-Res and a range of vibration frequency (25 Hz to 45 Hz) and amplitudes (1 and 3 mm) found by Marín et al. (2011). Additionally, the validity and reliability of the intensity of exertion using subjective scales in exercises with superimposed vibration have been demonstrated for both vibration frequency and muscle activation (Marín et al., 2012).

There were some limitations in the study. The effect of superimposed vibration on suspended exercises has been assessed in physically active men and women, so the results obtained in the present study cannot be generalized to other populations. The footwear soles were different among participants, and since this area is the most exposed to vibration, this could slightly modify the vibratory stimulus due to the damping effect of the footwear soles. Likewise, the vibration transmitted through the suspension strap could have dissipated the vibration effect. Another limitation was that the erector spinae and vastus (medialis and lateralis) requested in the supine bridge were not evaluated because the electromyography system employed only offers six channels.

CONCLUSIONS AND PRACTICAL APPLICATIONS

Regarding the main objective and the different objectives and hypotheses for each one of the studies presented in this PhD thesis, it is obtained the following conclusions:

- There is a lack of studies that quantify the effect of suspension devices in lower body exercises; however, on the upper body, we conclude that the suspension device increased activation in most of the muscle groups participating in suspension training exercises (push-ups, inverted row, prone bridge, and hamstring curl) compared to traditional. However, specific muscles (middle trapezius, posterior deltoid, and biceps brachii) did not differ concerning activation by exercise type, concretely during suspension inverted row (Study 1).
- Performing a suspended lunge at 80 cm feet distance, with a pace of 70 bpm and under dynamic contraction type enhance the force exerted on the suspension strap (Study 2).
- The assessment of the suspension training load, using a load cell, during a lunge seems to be useful for strength and conditioning coaches to individualize the athletes' load related to lunge position and force production (Study 2).
- Performing a suspended lunge provided no additional benefit than a Bulgarian squat to enhance lower body muscle activity (Study 3).
- A dual condition is needed to increase exercise muscle activity compared with a Bulgarian squat and suspended lunge. Despite this, dual condition decreases the suspension strap load, mainly when the front leg leans on an unstable surface (BOSU) (Study 3).
- VGRF exerted by the front leg in the suspended lunge (compared to its traditional counterparts) is enhanced to overcome the instability generated by the suspension device (Study 3).

- An accelerometer appears to be helpful in assessing the amount of perturbation produced by unstable surfaces under single and dual conditions in the suspended lunge (Study 4).
- An increase in the perturbation, quantified with the accelerometer, is related to an increase in muscle recruitment and a decrease in the force exerted on the suspension strap in the suspended lunge (Study 4).
- The sum of the amplitudes of the BCMA measured with the accelerometer is an adequate proposal to quantify the amount of instability during the unstable half-squat (Study 5).
- The OMNI-Res reflects the different levels of muscle activation obtained in half-squat conditions, but it does not coincide with BCMA levels in BOSU conditions (Study 5).
- In elite athletes, squatting on unstable surfaces, BOSU conditions, increases the activation of the primary half-squat movers (Study 5).
- The additional effect of the superimposed vibration was more challenging for the suspended supine bridge than the suspended hamstring curl. The pendulum movement of the suspended hamstring had a damper effect (Study 6).
- For the global activity, superimposed vibration at 25 Hz and 40 Hz in suspended supine bridge elicited the same effects on muscle activity (Study 6).
- The frequency of 25 Hz vibration provoked on the suspended supine bridge the higher activity of the most proximal muscles to the vibration device (gastrocnemius medialis, lateralis, and semitendinosus), with meaningless effects on the primary movers. Furthermore, the amount of instability provoked by the suspended supine bridge with superimposed vibration increased the stabilizing role of the gastrocnemius and semitendinosus (Study 6).

- OMNI-Res value progressively increased as vibration frequency raised (non-vibration, 25 Hz and 40 Hz vibration) in both suspended supine bridge and suspended hamstring curl (Study 6).

Additionally, a set of practical applications are presented to bring the different findings of this PhD thesis to strength and conditioning coaches, athletes, and fitness enthusiasts. The practical applications are detailed below:

Overall, depending on the conditioning goals and the strength demands of a given sport, a mixture of traditional exercises, suspension exercises and other conditioning methods can maximise performance. Specifically, it should consider the use of push-ups and suspended push-ups to improve the activation of pectoralis major, serratus anterior and anterior deltoid. The use of a pulley suspension device inhibits the muscles mentioned above, making these exercises less demanding. Likewise, the using a suspension device with a pulley is challenging for the core muscles in suspension push-ups, prone bridge, and hamstring curl. Moreover, when the participant tries to strengthen the triceps brachii, a suspension device with pulley is appropriate in suspension push-ups. In addition, the supine grip when performing the suspension inverted row exercise is recommended for higher demands on the biceps brachii. On the other hand, the use of different suspension devices and the body position in the most highly practised exercises in suspension training are crucial factors for exercise prescription in clinics and rehabilitation. The different set-ups in these exercises have an influence on the situations during which a high muscle activation is required. This is the case when suspension training becomes an interesting method in injury prevention and other clinical programs, such as those designed in the different phases of the rehabilitation and return-to-play protocols. Another clinical application of the different variation of suspension training exercises might guide the low back pain prevention programs and other postural pains.

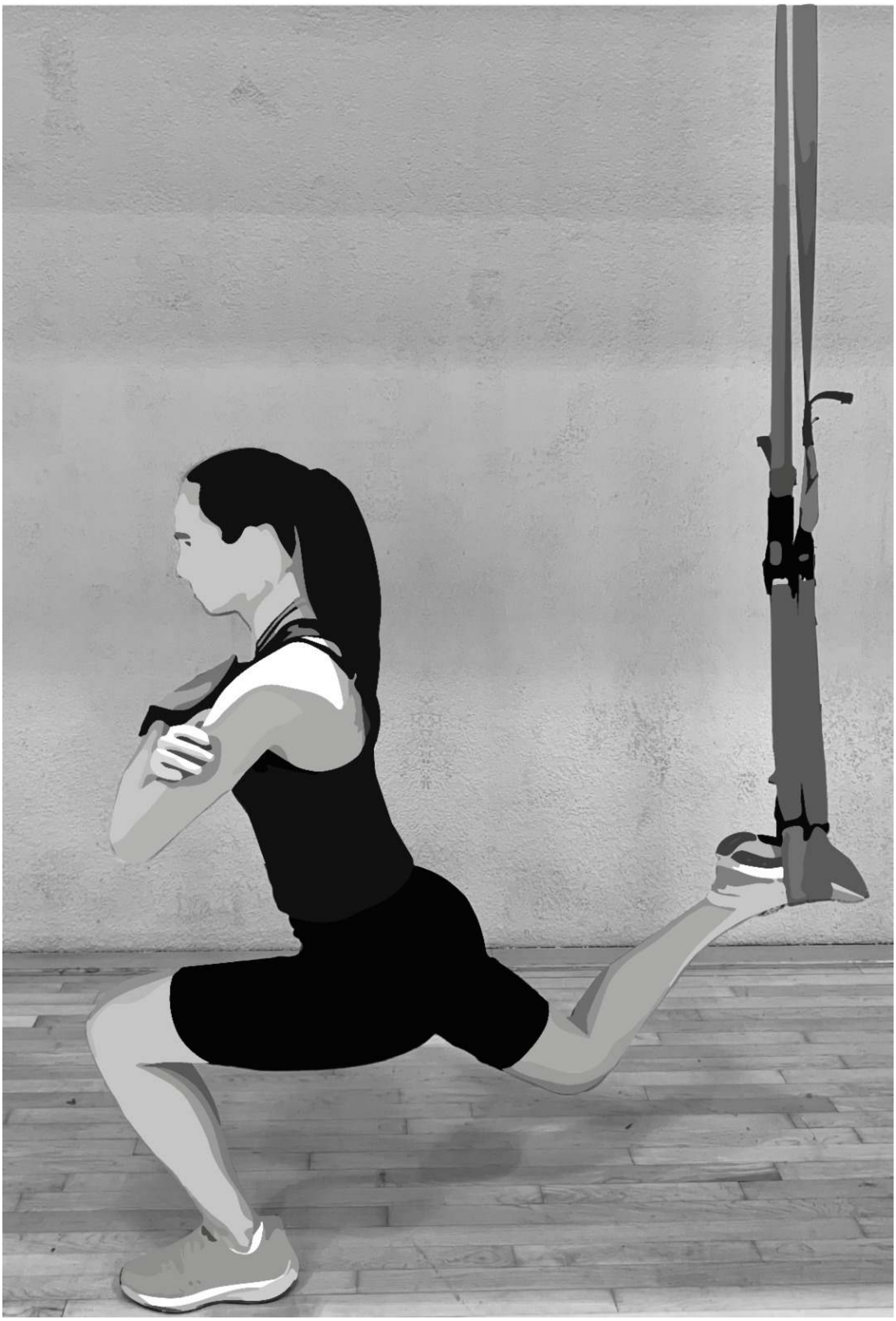
Regarding the quantification of the Bulgarian squat in an unstable environment (unstable surface and suspended) in terms of practical applicability, it is suggested that the variations on suspension lunge positions allow coaches and practitioners to achieve progress through position difficulty. Performing the suspended lunge is a good choice to strengthening the lower limb. The inclusion of this exercise in the strength and conditioning programs could be useful for improving the unilateral sport skills as jumping, COD, sprinting or shooting. Also, leaning the rear leg on the suspension device in the lunge exercise allows a higher demand in the front leg, this increasing the strength, power and balance. Apart of the changes in the body position, contraction type and pace, the coaches and practitioners could increase the muscular and force demands in the suspended lunge adding other sources of instability (in the front leg) or extra weights. In terms of muscular activity, the inclusion of additional methods increasing the instability (BOSU®, stability ball, Pielaster®, rubber mats), vibration with demanding amplitudes and frequencies, and extra weights (weighted vests and belts, barbells, kettlebells) on the front leg is necessary to increase the muscle activation because the simple use of a suspension device is not demanding enough for the Bulgarian squat under suspended conditions. The use of the accelerometer for quantifying the magnitude of the instability can be useful for the choice of the level of difficulty of the exercise (more or less unstable) relative to the objective of strength and conditioning training under unstable conditions, such as suspended lunge or half-squat, focusing on muscle recruitment or force production. Likewise, trainers or practitioners themselves can establish which exercises are more or less challenging based on the magnitude of the perturbation, in order to design a variety of exercise progressions.

Regarding to use the unstable surfaces for performing a loaded free barbell half-squat, it is proposed that unstable environments be used for increasing the activity of the prime movers and for other exercises with similar muscular demands. Furthermore, in strength and conditioning programmes the use of unstable surfaces provides variability which can be a crucial factor for maintaining the chronic adaptations of resistance training or for improving the acute effects of training through the design of tasks that produce instability in lower body exercises. In this sense, it should be considered that the selection of destabilising materials depends on the ability of the individuals to control movement by maintaining a balanced posture. In this manner, the primary muscles can be activated more to improve the effects of the training. For controlling the movement, it is recommended to determine a BCMA limit that could clarify how balanced is the execution of the exercise and the possible acute responses of the neuromuscular system. Additionally, BCMA monitoring would be useful to provide real time information and quantify the degree of instability in strength and conditioning exercises.

The inclusion of superimposed vibration in lower-body exercises investigated in this PhD thesis, such as suspended supine bridge does not elicit an additional effect on prime movers. Thus, the suspended supine bridge with superimposed vibration is as demanding as a traditional exercise for the gluteus maximus. However, the additional effect of the superimposed vibration in the suspended supine bridge provides greater gastrocnemius and hamstrings activity. Plantar flexion in the suspended supine bridge with superimposed vibration is a successful manner for strengthening the gastrocnemius, demanded in sports actions such as COD, jumps, and sprints. Furthermore, this method allows dynamic tasks, changing the planes of the force production and offering a continuous exposition to vibration for the working muscles. Likewise, the increased instability generated through vibration to the suspension straps turns the suspended supine bridge into an exercise that demands the neutralization of the lateral rotation of the thighs, similar to other lateral actions in several sports actions. Moreover, superimposed vibration in a suspension device can complement traditional exercises such as the Nordic hamstring, leg curl, or deadlift to develop the strength and endurance of the hamstrings in strength and conditioning programs. Additionally, injury prevention and rehabilitation can benefit from the outputs of the present PhD thesis to further evaluate the inclusion of superimposed vibration in the prescribed protocols since hamstrings injuries are prevalent in many sports.

Suggestions for future research

Athletes differ functionally and morphologically from non-athletes and unhealthy patients. They have more advanced muscular development, and their muscles are trained to sport-specific tasks. These differences may result in variations in muscle activation patterns during execution of suspension or unstable exercises. Future research should consider this type of population and focus on identifying muscle activations patterns for a greater variety of suspended and unstable lower limb exercises, and examine the muscle activity (specifically the role of the stabilizers muscles, such as gluteus maximus and medius, rectus abdominis, adductors or erector spinae) and force output when performing suspended lunges to compare the muscle recruitment between lower body under suspended and unstable conditions, and traditional resistance training exercises, including bilateral and unilateral suspension training exercises. Likewise, could be interesting to examine the muscle demands under the effects of superimposed vibration on suspension devices in several variations of lower-body exercises performed bilateral, unilateral or with additional load (kettlebell, barbells, weight plates). Also, future studies about superimposed vibration on suspensions straps should standardize the footwear for all participants to minimize the damping effect of the footwear soles. On the other hand, further research should study the effects of performing suspended and unstable lower limb exercises at different velocities and even in explosive exercises to establish the relationship between velocity-based resistance training with the body centre of mass acceleration. In addition, more research is needed to determine the effectiveness of the inclusion of suspension training exercises during the return-to-play phase of rehabilitation. Finally, carry out longitudinal and intervention studies to determine the effects of loaded suspension training (traditional, with pulley or with superimposed vibration) in strength and conditioning programs on the lower body and compare this training method with others (i.e., resistance training, inertial, dual condition, or superimposed vibration).



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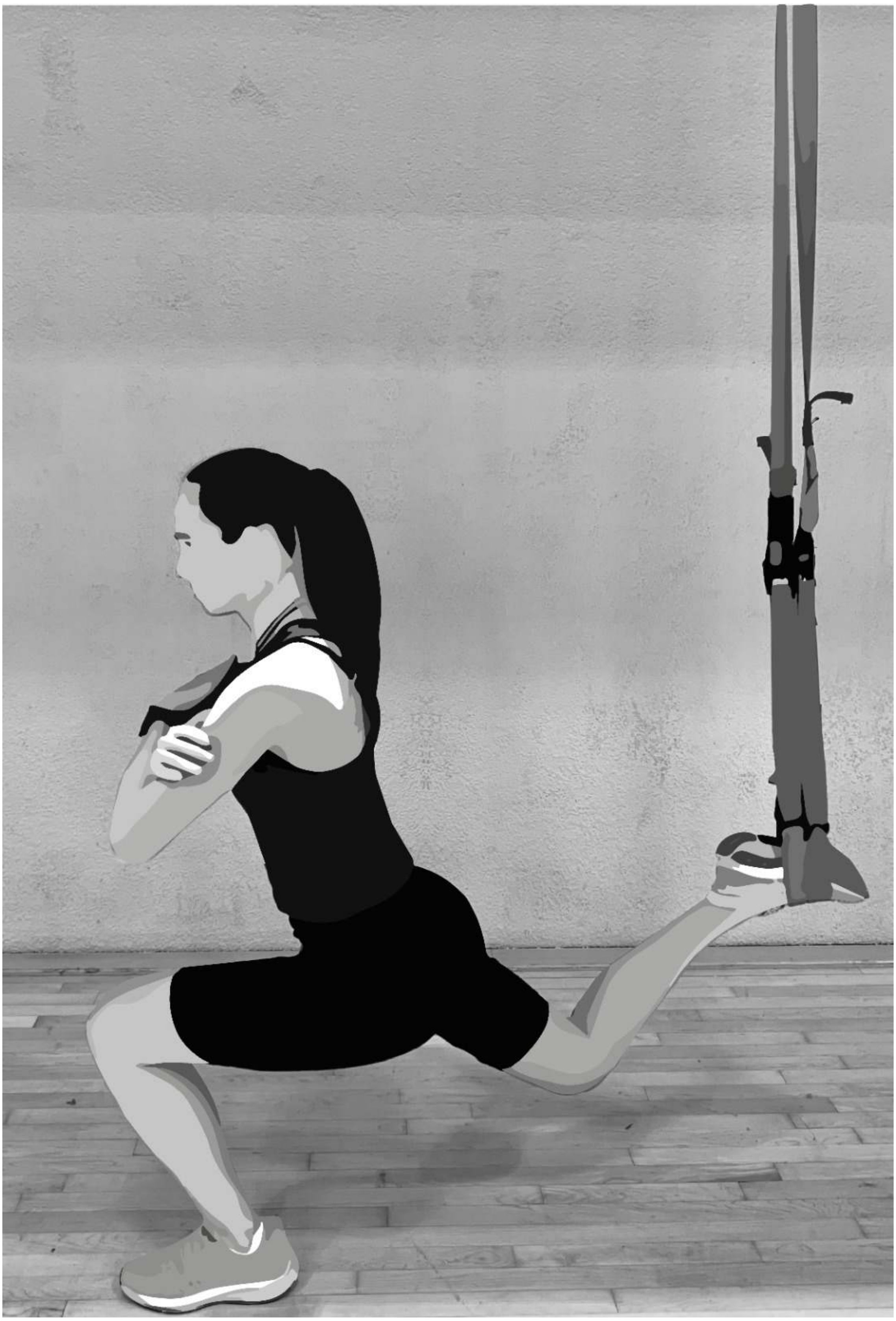
APPENDIX



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MUSCLE ACTIVATION IN SUSPENSION TRAINING: A SYSTEMATIC REVIEW

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Abstract

Suspension training is an adjunct to traditional strength and conditioning. The effect of added instability on muscle activation during traditional exercises is unclear and depends on the exercise and type of instability. The purpose of this review was to compare the activations of different muscles in suspension training exercises and their traditional counterparts. A search of the current literature was performed without language restrictions using the electronic databases PubMed (1969 – January 12, 2017), SPORTDiscus (1969 – January 12, 2017) and Scopus (1969 – January 12, 2017). The inclusion criteria were: (1) descriptive studies; (2) physically active participants; and (3) studies that analysed muscle activation using normalised electromyographic signals during different suspension training exercises. Eighteen studies met the inclusion criteria. For the push-up, inverted row, prone bridge and hamstring curl in suspension, the activation of upper-body and core muscles ranged between moderate (21%-40% maximum voluntary isometric contraction (MVIC)) and very high (>60% MVIC). Muscle activation in these same muscle groups was greater with suspension exercises relative to comparable traditional exercises, except for the inverted row. Muscle activation in the upper extremity and core muscles varied greatly amongst studies.

Keywords:
Electromyography
Instability
Muscular activity
Revision

Introduction

Traditional external load strength training focuses on working specific muscle groups with an emphasis on primary movers. However, advancements in knowledge about sport-specific demands have resulted in the development of training methodologies that involve new functional conditions. Thus, new trends in training have emerged to develop and enhance muscle activation during sport-specific movements (Lawrence & Carlson, 2015), and to improve the strength of accessory muscles by emphasising multiplanar movements. These movements result in improvements to agility, core strength, and posture (DiStefano, DiStefano, Frank, Clark, & Padua, 2013).

In recent years, the addition of instability to traditional exercises has become a popular method for increasing sport-specificity. The ability to maintain balance and the desired posture during sport-specific movements requires activation of core muscles, including the abdominal, back, and hip muscles. Instability resistance training increases the activation of core muscles essential to force production by the large primary movers (eg. hamstrings, quadriceps) (Behm & Colado, 2012). Furthermore, an unstable resistance training environment stresses the neuromuscular system and may promote greater strength gains and increases in cross-sectional area (Behm & Anderson, 2006; Cormie, McGuigan, & Newton, 2011). Unstable training may also increase motor unit recruitment and improve neuromuscular coordination without an increase in the mechanical load when performing push-ups under unstable conditions (Anderson, Gaetz, Holzmann, & Twist, 2013).

Traditionally, tools that create unstable surfaces have been used to progress the difficulty of exercises by stimulating increased motor unit recruitment. These tools include the Swiss ball, Balance Board®, Wobble Board®, and BOSU® exercises (Anderson et al., 2013; Duncan, 2009; Norwood, Anderson, Gaetz, & Twist, 2007; Vera-Garcia, Grenier, & McGill, 2000), and exercises with basketballs (Freeman, Karpowicz, Gray, & McGill, 2006).

A newer method available to increase activation is suspension training. Suspension training uses body weight and force momentum principles to enhance motor unit recruitment. The difficulty of the suspension training exercise and the number of motor units recruited depend on the amount of instability caused by the suspension device and the body position (Dawes & Melrose, 2015; Maté-Muñoz, Monroy, Jodra Jiménez, & Garnacho-Castaño, 2014). Although we cannot measure the amount of instability or the effect of the body position on instability during suspension training, we can measure electromyography (EMG) during exercise. Thus, EMG can be used to quantify the 'load' (Atkins et al., 2015; Borreani et al., 2015a; Snarr & Esco, 2013a).

Typically, muscle activation is presented as a percent of the maximum voluntary isometric contraction (%MVIC). Once MVIC is obtained, the EMG signal can processed several different ways: 1) by using high-pass filtering, 2) by rectifying and smoothing, or 3) by calculating the root mean square of the signal. The peak value registered after the signal processing constitutes the reference value of the normalised EMG signal (Halaki & Ginn, 2012). However, the same EMG signal may vary depending on the technique used to process it. Currently, there is no consensus about which technique should be used to process EMG signals. Studies comparing suspension training to traditional training utilise similar/the same exercises, but the results are difficult to reconcile unless there is an understanding on the differences in EMG signal processing techniques.

While activation of each muscle used to maintain stability has been studied for the most popular suspension training exercises, such as push-ups, inverted rows and prone bridges, to our knowledge, a review of this literature does not currently exist. The results of this review might be used to encourage the use of suspension exercises in place of traditional exercises in sport-specific resistance training. Choosing exercises that better suit an athlete's goals could enhance the effects and the quality of training programs. Therefore, the main purpose of this study was to perform a systematic review of studies that analysed the activation (% MVIC) of stabilising muscles involved in the most studied suspension training exercises in physically active populations. The secondary purpose of this study was to compare the muscle activation of the suspension training exercises with their traditional counterparts.

Methods

A Preferred Reporting Items for Systematic Reviews and Meta-Analysis (PRISMA) statement guidelines provided by Moher, Liberati, Tetzlaff, and Altman (2009) were used to conduct the present systematic review. Additionally, the study quality of all eligible cross-sectional studies was assessed by the first author and was checked by the second and the third co-authors using the Strengthening the Reporting of Observational Studies in Epidemiology (STROBE) criteria (Vandenbroucke et al., 2007). The following scale was used to classify study quality: a) good quality (>14 points, low risk of major or minor bias), b) fair quality (7-4 points, moderate risk of major bias), and c) poor quality (< 7 points, high risk of major bias). The score was obtained through the 22 items of the STROBE checklist.

A search of the current literature was performed without language restrictions using the electronic databases PubMed (1969 – January 12, 2017), SPORTDiscus (1969 – January 12, 2017) and Scopus (1969 – January 12, 2017). The search strategy for each database is listed in Table 1. MeSH terms were not used. The inclusion criteria were: a) studies that had a descriptive design, b) studies that utilised physically active participants, and c) studies that analysed muscle activation using normalised EMG signals during different suspension training exercises. Randomised control trials and clinical trials were excluded if they did not analyse muscle activation. Additionally, articles with insufficient discussion, poor data presentation, and unclear or vague descriptions of the applied protocols were excluded (see the flowchart of the search and study selection in Figure 2).

Table 1 near here

The first author performed the data analysis. First, a pre-reading was conducted to familiarise with terminology, Then, each article was re-read and the following information was extracted: 1) study design, 2) sample size, 3) gender, 4) age, 5) types of intervention (objective, exercise, measuring instruments), 6) EMG activation (expressed as %MVIC from normalised EMG), and 7) differences in EMG activation between traditional and suspension training exercises. Only the exercise type and the EMG activation differences between traditional and suspension training exercises (expressed as %MVIC from normalised EMG) were included in Tables 3 and 4. With regard to EMG signal, all analysed articles reported the MVIC protocol used. These protocols utilised isometric contraction against a matched resistance for each examined muscle. Similarly, all articles included used surface EMG. To facilitate the comparison of the muscle activation in different suspension training exercises, activation (% of MVIC) was categorised into four levels as described in previous studies: >60%, very high; 41-60%, high; 21-40%, moderate and <21%, low (Calatayud et al., 2014b; Escamilla et al., 2010; Mok et al., 2014).

All studies reported the muscle activation of each analysed muscle. Some authors examined the muscle activation of transversus abdominis and internal oblique together (Mok et al., 2014; Fong et al., 2015), while others only reported internal oblique activation (Beach, Howarth, & Callaghan, 2008; McGill, Cannon, & Andersen, 2014a, 2014b). For this reason transversus abdominis and internal oblique, and internal oblique activations were included in the analysis.

The suspension training exercises reported in the included studies were performed using three different suspension devices (Figure 1). The traditional suspension device has a main strap. On the bottom of this strap there are both, a main carabineer and a stabilising loop where another strap is locked, forming a V with handles on the bottom. The TRX® is an example of a traditional suspension device (Calatayud et al., 2014c). The pulley suspension device has a main strap supported by a spring and a V-rope with a pulley in the middle. The pulley's function is to reduce friction and increase unilateral motion (Calatayud et al., 2014c). Finally, Beach and colleagues (2008) used a suspension device with two parallel chains (similar to Olympic rings) and two independent anchors.

Figure 1 near here

Results

Search results

Three independent reviewers identified a total of 218 articles in the initial search. Sixty-eight articles were duplicates, which left 150 unique articles. Following title/abstract and full-text screening, 132 articles were eliminated because they did not meet the inclusion criteria (75 articles without EMG analysis, 5 clinical trials or randomised control trials and 52 articles that did not use a suspension device). A total of 18 articles were selected for final review (Figure 2).

Figure 2 near here

From the 18 studies reviewed, the suspension exercises described were: push-ups (Beach et al., 2008; Borreani et al., 2015a; Borreani et al., 2015b; Calatayud et al., 2014b; Calatayud et al., 2014a; Calatayud et al., 2014c; Fong et al., 2015; McGill et al., 2014a; Mok et al., 2014; Snarr, Esco, Witte, Jenkins & Brannan, 2013; Snarr & Esco, 2013b), inverted row (Fong et al., 2015; McGill et al., 2014b; Mok et al., 2014; Snarr & Esco, 2013a; Snarr, Nickerson & Esco, 2014), prone bridge (Atkins et al., 2015; Byrne et al., 2014; Fong et al., 2015; Mok et al., 2014; Snarr & Esco, 2014), and hamstring curl (Fong et al., 2015; Malliaropoulos et al., 2015; Mok et al., 2014).

Of the 18 included studies, only Fong and colleagues (2015) and Mok and colleagues (2014) were excluded from the secondary analysis. Fong and colleagues (2015) and Mok and colleagues (2014) only compared muscle activation during different suspension training exercises. Of the 16 remaining studies, nine compared muscle activation during suspension training and traditional exercise (Beach et al., 2008; Borreani et al., 2015b; Byrne et al., 2014; Calatayud et al., 2014a; McGill et al., 2014a, 2014b; Snarr & Esco, 2013a, 2013b; Snarr, Nickerson, et al., 2014), five compared muscle activation during suspension training exercise with traditional exercise performed on an unstable surface (eg. prone bridge on a Swiss ball)(Atkins et al., 2015; Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014b; Snarr & Esco, 2014), and two compared muscle activation during suspension training exercise with different traditional exercises (Malliaropoulos et al., 2015; Snarr et al., 2013). The differences in activity (expressed as % of MVIC) of the muscles involved in push-ups, inverted row, prone bridge and hamstring curl exercises are described in the following paragraph. Results are presented below according to primary objective (muscle activation of the suspension training exercises) and secondary objective (muscle activation comparison between suspension training exercises and traditional strength training counterparts). Table 2 shows the descriptive characteristics and the quality of all studies revised.

Table 2 near here

Muscle activation during suspension exercises

Suspension push-up

Muscle activation during suspension push-ups is reported in Figure 3. For suspension push-ups, activation of triceps brachii, serratus anterior, and rectus abdominis were high (41%-60% MVIC). Activations of pectoralis major, anterior deltoid, transversus abdominis and internal oblique, external oblique, and rectus femoris were moderate (21%-40% MVIC). Activations of upper trapezius, posterior deltoid, latissimus dorsi, internal oblique, erector spinae, lumbar multifidus, and gluteus maximus were low (<21% MVIC).

Figure 3 near here

Suspension inverted row, prone bridge and hamstring curl

Muscle activations during suspension inverted rows, suspension prone bridges and suspension hamstring curls are reported in Figures 4, 5 and 6 respectively. For suspension inverted row, activations of latissimus dorsi, middle trapezius, posterior deltoid, and biceps brachii were very high (>60% MVIC). Activations of core muscles (transversus abdominis and internal oblique, rectus abdominis, external oblique, internal oblique, lumbar multifidus) were low (<21% MVIC). In suspension prone bridge, activations of some core muscles (transversus abdominis and internal oblique, rectus abdominis, external oblique) ranged from high to very high (>41% MVIC), while activations of other core muscles (serratus anterior, lumbar multifidus, erector spinae, rectus femoris) ranged from moderate to low (< 40% MVIC). For suspension hamstring curl, activations of biceps femoris and semitendinosus were very high (> 60% MVIC). Activations of some core muscles (transversus abdominis and internal oblique, and lumbar multifidus) were high (41-60% MVIC) while others (external oblique and rectus abdominis) ranged from moderate to low (<40% MVIC).

Figure 4 near here

Figure 5 near here

Figure 6 near here

Muscle activation in suspension exercises compared to traditional exercises

Suspension push-up

Differences in activations between suspension push-ups and push-ups for each muscle are reported in Table 3. Activations of pectoralis major, anterior deltoid, upper trapezius, triceps brachii, latissimus dorsi, serratus anterior, rectus abdominis, external oblique, internal oblique, lumbar multifidus and rectus femoris were significantly greater in suspension push-up compared to push-up (Beach et al., 2008; Borreani et al., 2015a; Borreani et al., 2015b; Calatayud et al., 2014a; McGill et al., 2014a; Snarr & Esco, 2013b). The use of a suspension device with pulley, a type of suspension training device that uses a pulley to further increase instability, caused significant increases in activation of upper trapezius, triceps brachii, posterior deltoid, rectus abdominis, external oblique, erector spinae, rectus femoris and gluteus maximus relative to traditional push-up (Calatayud et al., 2014b; Calatayud et al., 2014a; Calatayud et al., 2014c). The traditional push-up resulted in significantly higher activations of pectoralis major and anterior deltoid compared to suspension push-up with pulley (Borreani et al., 2014a). However, for certain muscles, like pectoralis major (Borreani et al., 2015b), anterior deltoid (Borreani et al., 2015b; Calatayud et al., 2014b; Calatayud et al., 2014c) and serratus anterior (Borreani et al., 2014b; McGill et al., 2014a), significantly greater activation was found in traditional push-up in comparison with suspension push-up in the aforementioned studies.

** Table 3 near here**

Suspension inverted row, prone bridge and hamstring curl

Differences in activations between suspension and traditional inverted row, prone bridge and hamstring curl for each muscle are reported in Table 4. Activations of middle trapezius, posterior deltoid, rectus abdominis, internal oblique, external oblique and erector spinae were greater in suspension inverted row compared to inverted row; however, the increases were not statistically significant (McGill et al., 2014b; Snarr & Esco, 2013a; Snarr, Nickerson et al., 2014). Activation of latissimus dorsi was significantly greater in inverted row compared to suspension inverted row, but biceps brachii activity was significantly higher in suspension inverted row compared to inverted row (Snarr, Nickerson et al., 2014). Activations of core muscles (rectus abdominis, external oblique, erector spinae and rectus femoris) were significantly greater in suspension prone bridge compared to prone bridge (Atkins et al., 2015; Byrne et al., 2014; Snarr & Esco, 2014). Activations of biceps femoris and semitendinosus were significantly greater in suspension hamstring curl compared to traditional exercise with and without destabilising devices (Malliaropoulos et al., 2015).

** Table 4 near here**

Discussion and implications

Muscle activation during suspension exercises

The primary aim of this study was to identify muscle activation during execution of different suspension training exercises. We found both similarities and differences in the activations of the same muscles amongst the exercises. Any differences may be attributed to differences in body position and conditioning parameters (range of motion, suspension height, type of grip, type of suspension training device, etc.). Each type of exercise will be discussed in detail below.

Suspension push-up

Some studies found the pectoralis major activation to be moderate (Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b), but Snarr and Esco (2013b) found that the pectoralis major activation was very high. This difference may be explained by differences in the types of suspension device used. Several studies used a traditional suspension devices (Borreani et al., 2015b; Snarr & Esco, 2013b), while other studies used a pulley suspension devices (Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b). Additionally, there were differences in the height of the suspension device. The activation of the pectoralis major decreased as the trunk-legs inclination, or the angle between the body and the floor, increased and the hip flexion decreased (Borreani et al., 2015b). Similarly, the wide range of the anterior deltoid activation can be attributed to these same two factors (suspension device type and length). These results suggest that the anterior deltoid is inhibited as a synergist of the pectoralis major, thus reducing the anterior deltoid's activation under highly unstable conditions. The stabilising function of core muscles (rectus abdominis and external oblique) using a pulley suspension device is more demanding (Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b), in contrast to the lower activation of the serratus anterior (Calatayud et al., 2014b). The differences in the rectus abdominis and external oblique activations amongst studies could be partially explained by differences in suspension device height and the trunk-legs inclination (Dawes & Melrose, 2015) and by changes caused by increased instability. Despite changes in activation of the other muscles, the activation of the triceps brachii during push-ups did not vary by suspension device type (Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b); however, it did vary by the trunk-legs inclination (Snarr & Esco, 2013b).

With regard to back muscles, the activation of the erector spinae was reduced with the use of a pulley suspension device (Calatayud et al., 2014a; Calatayud et al., 2014c). This is because suspension push-ups increase the activation of the latissimus dorsi to stabilise the shoulder joint. Additionally, the use of a suspension device requires the abdominal muscles and latissimus dorsi to be sufficiently activated to achieve mechanical equilibrium around the lower back (Beach et al., 2008).

The differences between the muscle activity of the pectoralis major, anterior deltoid and upper trapezius as reported by Snarr and Esco (2013b) and the other studies can be explained by the use of a traditional suspension device that did not inhibit its activation and the body inclination having supported the major part of the body mass, in accordance with the vector resistance and stability fundamental principles (Bettendorf, 2010). Again, the differences in the muscle activation of the anterior deltoid, serratus anterior and rectus femoris, between Borreani and colleagues (2015a) and the other studies, could be explained by the use of the traditional suspension device, which produces a degree of instability to increase the activity of the aforementioned musculature. The differences in the activation of the rectus abdominis were provoked by the type of suspension device and for modified the vector resistance and stability fundamental principles. Thus, by reducing the body angle inclination and performing the push-ups exercise with the suspension device with pulley, the participants in the study by Calatayud and colleagues (2014a, 2014c) achieved an increase of muscle demands and very high activations (>60% MVIC) from the rectus abdominis.

Suspension inverted row, prone bridge and hamstring curl

The differences in activations of the latissimus dorsi and middle trapezius between suspension inverted row studies can be explained by handgrip type. Snarr and Esco (2013a) and Snarr, Nickerson, and colleagues (2014) used pronated and supinated handgrips, respectively, while McGill and colleagues (2014b) used a neutral handgrip. The pronated handgrips demand a higher activation of the posterior deltoid, whereas the supinated handgrips demands a higher activation of the biceps brachii (Snarr, Nickerson, et al., 2014). The prone handgrip could be enhanced the extensor role of the posterior chain musculature.

The biceps brachii activity observed by Snarr, Nickerson and colleagues (2014) differed to the other studies. This could be because they used a supine handgrip, which increases the recruitment of the biceps brachii during elbow flexion. Additionally, the trunk-legs inclination and the hip flexion angle influence the latissimus dorsi and middle trapezius activation. Activations of core muscles (rectus abdominis, external oblique and lumbar multifidus) were low in suspension inverted row, as the instability created by traditional suspension devices does not engage the trunk muscles, even though instability increases with the variation of the trunk-legs inclination and the hip flexion angle. The included suspension inverted row studies showed that the variations in the suspension training fundamental principles (Bettendorf, 2010) were insufficient to offer a challenge to the recruitment of the core muscles.

In the suspension prone bridge, the variation in rectus abdominis, external oblique and erector spinae activations may be due to the additions of hip abduction (Fong et al., 2015; Mok et al., 2014), arms extension (Atkins et al., 2015), suspension of feet or arms (Snarr & Esco, 2014), or suspension of feet and arms (Byrne et al., 2014). Additionally, the modification in the fundamental principles of stability and pendulum (Bettendorf, 2010) by Mok and colleagues (2014) were insufficient to increase the degree of instability, especially in the rectus abdominis, which activation was lower compared to the aforementioned studies.

The suspension hamstring curl technique utilised by Fong and colleagues (2015) in chronic back pain patients, which entails positioning the supine trunk with a lumbopelvic retroversion, could explain the differences in the activation of trunk muscles (transversus abdominis and internal oblique, rectus abdominis and external oblique) in comparison with Mok and colleagues (2014).

Muscle activation in suspension exercises compared to traditional exercises

Studies that compared muscle activation during suspension and traditional exercises utilised physically active participants between 15 and 28 years of age. Most studies presented activations as %MVIC; however, these studies used different procedures to obtain the MVIC. For example, Malliaropoulos and colleagues (2015) performed three five-second MVICs with an isokinetic dynamometer whereas other studies performed MVIC trials against a matched resistance (Beach et al., 2008; Byrne et al., 2014; Borreani et al., 2015a), with trial lengths ranging from three to ten seconds (Atkins et al., 2015; Byrne et al., 2014; Mok et al., 2014). The MVIC trials for each analysed muscle differed in accordance with the protocol used (e.g. the protocols described by Konrad (2005) compared to Escamilla and colleagues (2010)).

Suspension push-up

All the reviewed studies showed a greater activation of suspension push-ups compared to traditional push-ups, regardless of the use of stabilising loop. The use of a suspension device without a stabilising loop significantly improved the activation of the pectoralis major (Calatayud et al., 2014c). However, suspension push-ups performed under highly unstable conditions provoked a lower activation of the pectoralis major (Calatayud et al., 2014a) and anterior deltoid (Borreani et al., 2015b). This is because the triceps brachii acts as a stabiliser under unstable conditions, especially when instability is lateral (Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b; Snarr & Esco, 2013b). Furthermore, lateral instability of the pulley suspension device results in greater activation of the trunk muscles (rectus abdominis, internal oblique and external oblique). This is especially true for the rectus abdominis, which acts as a grater stabiliser during suspension push-ups compared to traditional push-ups (Beach et al., 2008; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b; McGill et al., 2014a; Snarr et al., 2013). Thus, the activation of the rectus abdominis during suspension push-ups with a pulley substantially increases (around 20%) in comparison with traditional suspension push-ups. This is an important finding for clinicians looking for a way to more thoroughly engage the abdominals during sports-specific upper extremity movements. In contrast, there was no consensus in the included studies for the serratus anterior on the effects of instability during push-ups; however, variations in the suspension device height suggested that the higher the position of the suspension device, the greater the activation of the serratus anterior. (Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b).

Suspension inverted row, prone bridge and hamstring curl

Variations in the grip type used in studies investigating inverted row exercise make it difficult to compare muscle activations. Along with the grip type, variations in trunk-legs inclination and the hip flexion angle result in variations in the activation of the involved muscles (latissimus dorsi, middle trapezius, posterior deltoid, and biceps brachii), which cause significant differences in activations between suspension inverted row and inverted row (McGill et al., 2014b; Snarr & Esco, 2013a; Snarr, Nickerson, et al., 2014). As such, it is difficult to tell if this difference is due to the addition of instability to the exercise. However, for the suspension prone bridge, compared with the traditional prone bridge, the variations in execution (arms or legs suspended) increases the activation of muscles used to maintain body position (rectus abdominis, external oblique, rectus femoris, serratus anterior and erector spinae) (Byrne et al., 2014; Snarr & Esco, 2014). Furthermore, instability of the arms may increase the difficulty of the exercise more than instability of the legs, which may cause higher rectus abdominis activation (Byrne et al., 2014). In contrast, the activation of trunk muscles reported by Atkins and colleagues (2015) differ from other studies because they studied elite swimmers who have particular neuromuscular characteristics. Finally, although there is little evidence in the literature, the hamstring musculature was more activated during the suspension hamstring curl than other bilateral hamstring exercises (Malliaropoulos et al., 2015).

Methodological considerations

There is no consensus about the protocol used to normalise the EMG signal for calculating MVIC. The procedures used to produce the MVIC, which can be achieved progressively or explosively, were not provided by most authors. Compared to the use of rapid movement, the use of isometric and slow dynamic movements for achieving a maximal contraction results in more reliable and easy-to-compare EMG signals (Alizadehkhaiyat & Frostick, 2015). Only three studies (Atkins et al., 2015; Byrne et al., 2014; Malliaropoulos et al., 2015) followed the European Recommendation for Surface ElectroMyoGraphy (SENIAM) (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000) guidelines, which warn clinicians to be cautious when making comparisons of muscle activations. Regardless, all reviewed studies reported the electrode placement in detail.

The majority of the reviewed studies used the metronome for pace control in push-ups, except for Beach and colleagues (2008), Fong and colleagues (2015) and Snarr and colleagues (2013), and for prone-bridge studies, except for Mok and colleagues (2014). As such, it is possible that the differences in muscle activity amongst studies for suspension push-ups exercises may have been caused by the type of muscle contraction and the execution velocity. This may be true for dynamic suspension push-up protocols (Beach et al., 2008; Borreani et al., 2015a; Borreani et al., 2015b; Calatayud et al., 2014a; Calatayud et al., 2014c; Calatayud et al., 2014b; Snarr & Esco, 2013b; Snarr et al., 2013), dynamic suspension push-up protocols combining isometric contraction while maintaining movement during the eccentric phase (McGill et al., 2014a; Mok et al., 2014), and isometric suspension push-up protocols (Fong et al., 2015). Research has indicated that muscle activation is highest for dynamic suspension push-ups compared to isometric suspension push-ups or protocols combining both types of contractions. Furthermore, studies with dynamic suspension inverted row protocols (Snarr & Esco, 2013a; Snarr, Nickerson, et al., 2014) demonstrated higher muscle activation compared to studies that used a combination of dynamic and isometric contractions during the concentric phase (McGill et al., 2014b). In addition, the protocol conducted to analyse the core musculature (transversus abdominis and internal oblique, rectus abdominis, external oblique, and lumbar multifidus) in the suspension hamstring curl was similar in Mok and colleagues (2014) and Fong and colleagues (2015). These findings suggest that the protocol (pace control and type of contraction) is not responsible for the differences in muscle activation. Instead, differences may be due to variations in the study populations and procedures used to normalise the EMG signals.

Limitations

In this review, most of the references are from 2013 or later, which indicates that the study of muscle activation during suspension training exercises is a new topic in strength and conditioning. In fact, the manufacturer of the most popular suspension devices patented its first device in 2006. There were certain variations in the execution of the exercises, the muscles assessed for activation, the methods used to assess muscle activity and the EMG signal normalisation procedures. As a result, readers should be cautious in interpreting the magnitudes of muscle activations between the same muscles and amongst exercises. Moreover, the participants in a majority of the studies were physically active male non-athletes, which makes it difficult to generalise findings to both females and athletes.

Suggestions for future research

Athletes differ functionally and morphologically from non-athletes and unhealthy patients. They have more advanced muscular development and their muscles are trained to sport-specific tasks. These differences may result in variations in muscle activation patterns during execution of suspension exercises. Future research should consider this type of population and focus on identifying muscle activations patterns for a greater variety of suspension exercises, including exercises involving the lower extremity, and comparing bilateral and unilateral suspension training exercises. In addition, more research is needed to determine the effectiveness of the inclusion of suspension training exercises during the return-to-play phase of rehabilitation. Finally, Genevois and colleagues (2014); Janot and colleagues (2013) and Maté-Muñoz and colleagues (2014) all recommended longitudinal research and interventional research that measures muscle activity using EMG.

Conclusion

After a detailed systematic review of studies analysing muscle activity during different suspension training exercises (push-ups, inverted row, prone bridge and hamstring curl), we can conclude that there are differences in muscle activation between the exercises using suspension devices and their traditional counterparts. The suspension device increased activation in the most of the muscle groups participating in suspension training exercises (push-ups, inverted row, prone bridge and hamstring curl) compared to traditional. However, certain muscles (middle trapezius, posterior deltoid and biceps brachii) did not differ with regard to activation by exercise type, concretely during suspension inverted row. Depending on the conditioning goals and the strength demands for a given sport, a mixture of traditional exercises, suspension exercises and other conditioning methods can be used to maximise performance.

Clinicians and practitioners should consider the use of push-ups and suspended push-ups to improve the activation of pectoralis major, serratus anterior and anterior deltoid. The use of a pulley suspension device inhibits the aforementioned muscles, which make these exercises less demanding. Likewise, the use of a suspension device with pulley is a challenge for the core muscles in suspension push-ups, prone bridge and hamstring curl. Moreover, when the participant tries to strengthen the triceps brachii, the use of a suspension device with pulley is appropriate in suspension push-ups. In addition, the supine grip when performing the suspension inverted row exercise is recommended for higher demands on the biceps brachii.

The use of different suspension devices and the body position in the most highly practised exercises in suspension training are crucial factors for exercise prescription in clinics and rehabilitation. The different set-ups in these exercises have an influence on the situations during which a high muscle activation is required. This is the case when suspension training becomes an interesting method in injury prevention and other clinical programs, such as those designed in the different phases of the rehabilitation and return-to-play protocols. Another clinical application of the different variation of suspension training exercises might guide the low back pain prevention programs and other postural pains.

Disclosure statement

The authors have no conflicts of interest to disclose.

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Table 1. Search strategy, filters, and databases used

Database	Records identified	Search strategy
MEDLINE (PubMed)	n= 54	1 suspen* OR TRX 2 electromyograp* OR ‘muscle activation’ 3 training OR stability OR stable OR instability OR unstable 4 1 AND 2 AND 3
SPORTDiscus	n= 33	1 suspen* OR TRX 2 electromyograp* OR ‘muscle activation’ 3 training OR stability OR stable OR instability OR unstable 4 1 AND 2 AND 3
Scopus	n= 131	1 suspen* OR TRX 2 electromyograp* OR ‘muscle activation’ 3 training OR stability OR stable OR instability OR unstable 4 1 AND 2 AND 3
Total	n= 218	

Filter: References from other articles were consulted

Table 2. Descriptive characteristics and quality assessment of cross-sectional studies describing the effects of suspension training exercises on muscle activity in physically active populations

Author/s (year)	Suspension exercise	Sample Size	Sex	Age (mean)	Resistance training experience	Study quality (STROBE points)
Atkins et al. (2015)	Prone Bridge	18	18 Male	15.9	NA	Good (17)
Beach et al. (2008)	Push-ups	11	18 Male	27.4	NA	Good (16)
Borreani et al. (2015a)	Push-ups	30	30 Male	23	1 year	Fair (14)
Borreani et al. (2015b)	Push-ups	29	29 Male	23.5	1 year	Good (15)
Byrne et al. (2014)	Prone Bridge	21	11 Male, 10 Female	21.9	NA	Fair (14)
Calatayud et al. (2014a)	Push-ups	29	29 Male	23.5	1 year	Good (15)
Calatayud et al. (2014b)	Push-ups	29	29 Male	22..6	1 year	Good (15)
Calatayud et al. (2014c)	Push-ups	29	29 Male	23.5	1 year	Good (15)
Fong et al. (2015)	Push-ups Inverted row Prone bridge Hamstring curl	21	11 Male, 10 Female	21.4	NA	Good (15)
Malliaropoulos et al. (2015)	Hamstring curl	20	20 Female	22.8	NA	Fair (14)
McGill et al. (2014a)	Push-ups	14	14 Male	21.1	NA	Good (15)
McGill et al. (2014b)	Inverted row	14	14 Male	21.1	NA	Good (16)
Mok et al. (2014)	Push-ups Inverted row Prone bridge Hamstring curl	18	8 Male, 10 Female	21.9	NA	Fair (14)
Snarr and Esco (2013a)	Inverted row	15	11 Male, 4 Female	25.4	NA	Fair (14)
Snarr and Esco (2013b)	Push-ups	21	15 Male, 6 Female	25.24	6 months	Fair (14)
Snarr and Esco (2014)	Prone bridge	12	6 Male, 6 Female	23.25	6 months	Fair (14)
Snarr et al. (2013)	Push-ups	15	12 Male, 3 Female	25.27	NA	Fair (14)
Snarr, Nickerson et al. (2014)	Inverted row	20	12 Male, 8 Female	26.6	NA	Fair (14)

NA_Not available

Table 3. Push-ups: Differences in muscle activation between traditional and suspension exercise for each muscle

STUDY	EXERCISE TYPE	RESULTS (% MVIC)
Beach et al. (2008)	SD: Push-ups with chains FL: Push-ups	**Rectus abdominis:** ↑* 25.6% in suspension push-ups vs push-ups **External oblique:** ↑* 8.4% in suspension push-ups vs push-ups **Internal oblique:** ↑* 8.5% in suspension push-ups vs push-ups **Latissimus dorsi**: ↑* 4.2% in suspension push-ups vs push-ups **Erector spinae:** No # in suspension push-ups vs push-ups
Snarr and Esco (2013b)	SD: Push-ups with TRX® FL: Push-ups	**Pectoralis major:** ↑* 5.92% in suspension push-ups vs push-ups **Anterior deltoid:** ↑* 22.22% in suspension push-ups vs push-ups **Triceps brachii:** ↑* 31.51% in suspension push-ups vs push-ups
Snarr et al. (2013)	SD: Push-ups with TRX® FL: Push-ups and abdominal supine crunch	**Rectus abdominis:** ↑* 47% in suspension push-ups vs push-ups No # in suspension push-ups vs supine crunch
McGill et al. (2014a)	SD: Push-ups with TRX® at different angles FL: Push-ups	**Serratus anterior:** ↑61.73% in push-ups vs suspension push-ups ↑19.6% in suspension push-ups scapula vs suspension push-ups angle 2 **Rectus abdominis:** ↑22% in suspension push-ups vs push-ups **External oblique:** ↑8% in suspension push-ups vs push-ups **Internal oblique:** ↑6% in suspension push-ups vs push-ups
Calatayud et al. (2014a)	SD: Push-ups with pulley (AirFit Trainer Pro®) at different heights (10 cm and 65 cm) FL: Push-ups at different heights (10 cm and 65 cm)	**Pectoralis major:** ↑* 3.65% in push-ups vs suspension push-ups with pulley **Anterior deltoid:** ↑* 6.74% in push-ups vs suspension push-ups with pulley **Upper trapezius:** ↑* 10.92% in suspension push-ups with pulley vs push-ups **Triceps brachii:** ↑* 31.79% in suspension push-ups with pulley vs push-ups **Rectus abdominis:** ↑* 79.94% in suspension push-ups with pulley vs push-ups **Rectus femoris:** ↑* 9.57% in suspension push-ups with pulley vs push-ups **Erector spinae:** ↑* 2.08% in suspension push-ups with pulley vs push-ups **Gluteus maximus:** ↑* 1.57% in suspension push-ups with pulley vs push-ups
Calatayud et al. (2014b)	SD: Push-ups with pulley (AirFit Trainer Pro®) and TRX® FL: Push-ups and variations (elastic resistance, bench press and press of cables)	**Pectoralis major:** ↑* 26.78% in press bench (85% 1RM) vs suspension push-ups / No # in suspension push-ups vs push-ups **Anterior deltoid:** ↑* 13.46% in push-ups vs suspension push-ups **Upper trapezius:** ↑* 5.61% in suspension push-ups with pulley vs push-ups **Triceps brachii:** ↑* 27.15% in suspension push-ups with pulley vs push-ups **Rectus abdominis:** ↑* 47.67% in suspension push-ups vs push-ups **External oblique:** ↑* 36.83% in suspension push-ups vs push-ups **Serratus anterior:** ↑* 11.27% in push-ups vs suspension push-ups **Posterior deltoid:** ↑* 7.45% in suspension push-ups with pulley vs push-ups

Calatayud et al. (2014c)	SD: Push-ups with pulley (AirFit Trainer Pro®) and without pulley (TRX®, Flying®, Jungle Gym XT®) FL: Push-ups	**Pectoralis major:** ↑* 12% in suspension push-ups double anchor (Jungle Gym XT®) vs push-ups **Anterior deltoid:** ↑* 7.81% in push-ups vs suspension push-ups No # in push-ups vs suspension push-ups double anchor (Jungle Gym XT®) **Upper Trapezius:** ↑* 14.49% in suspension push-ups with pulley vs push-ups **Triceps brachii:** ↑* 30.68% in suspension push-ups with pulley vs push-ups **Rectus abdominis:** ↑* 81.68% in suspension push-ups with pulley vs push-ups **Rectus femoris:** ↑* 11.78% in suspension push-ups with pulley vs push-ups **Erector spinae:** ↑* 2.29% in suspension push-ups with pulley vs push-ups
Borreani et al. (2015a)	SD: Push-ups with TRX® DM: Push-ups with Wobble board®, Stability disc® and Fitness Dome® FL: Push-ups	**Anterior deltoid:** No # in suspension push-ups vs push-ups, wobble board®, stability disc®, fitness dome® **Serratus anterior:** ↑* 66.76% in wobble board® vs push-ups ↑* 66.05% in fitness dome® vs push-ups ↑* 55.15% in stability disc® vs push-ups ↑* 46.41% in suspension push-ups vs push-ups **Lumbar multifidus:** ↑* 3.38% in suspension push-ups vs push-ups **Rectus femoris:** ↑* 17.31% in suspension push-ups vs push-ups
Borreani et al. (2015b)	SD: Push-ups with TRX® at different heights (10 cm and 65 cm) FL: Push-ups at different heights (10 cm and 65 cm)	**Pectoralis major:** ↑* 4.33% in push-ups (65 cm) vs suspension push-ups (65 cm) No # in suspension push-ups (10 cm) vs push-ups (10 cm) **Anterior deltoid:** ↑* 5.1% in push-ups vs suspension push-ups **Upper trapezius:** ↑* 10.28% in suspension push-ups vs push-ups **Triceps brachii:** ↑* 28.27% in suspension push-ups vs push-ups

SD _ Suspension Device; FL _ Floor; DM _ Destabilizing Material; %MVIC _ percent of maximum voluntary isometric contraction; ↑* _ Significantly increases ($p<0.05$); ↑ _ increases; # _ difference

Table 4. Inverted row, prone bridge and hamstring curl: Differences in muscle activation between traditional and suspension exercise for each muscle

STUDY	EXERCISE TYPE	RESULTS (%MVIC)
Snarr and Esco (2013a)	SD: inverted row with TRX® FL: inverted row	**Latissimus dorsi:** No # in suspension inverted row vs inverted row **Middle trapezius:** No # in suspension inverted row vs inverted row **Posterior deltoid:** No # in suspension inverted row vs inverted row **Biceps brachii:** ↑* 8.8% in inverted row vs suspension inverted row
Snarr, Nickerson et al. (2014)	SD: inverted row with pronated and supinated grip in TRX ® FL: inverted row with pronated and supinated grip	**Latissimus dorsi:** ↑* 8.47% in inverted row (prone grip) vs suspension inverted row (prone grip) / ↑* 11.01% in suspension inverted row (supine grip) vs suspension inverted row (prone grip) / No # in suspension inverted row (supine grip) vs suspension inverted row (neutral grip) **Middle trapezius:** ↓* 29.05% in suspension inverted row (supine grip) vs inverted row (prone grip) / ↓* 28.64% in suspension inverted row (supine grip) vs suspension inverted row (neutral grip) / ↓* 15.1% in suspension inverted row (supine grip) vs suspension inverted row (prone grip) **Posterior deltoid:** ↑* 18.27% in suspension inverted row (neutral grip) vs suspension inverted row (prone grip) / ↑* 23.32% in suspension inverted row (neutral grip) vs suspension inverted row (supine grip) / ↑* 6.17% in suspension inverted row (neutral grip) vs inverted row (prone grip) **Biceps brachii:** ↑* 6.65% in suspension inverted row (supine grip) vs inverted row (prone grip) / ↑* 6.02% in suspension inverted row (supine grip) vs suspension inverted row (prone grip) / ↑* 17.83% in suspension inverted row (supine grip) vs inverted row (supine grip)
McGill et al. (2014b)	SD: inverted row and other pulling exercises with TRX® FL: inverted row and other pulling exercises	**Latissimus dorsi:** No # in suspension inverted row vs inverted row /↑* 16.5% in suspension inverted row vs suspension inverted row (TRX® angle 1 and angle 2) / No # in TRX® ghost vs TRX® one-arm ghost **Rectus abdominis:** No # in suspension inverted row vs inverted row / No # in TRX® ghost vs TRX® one-arm ghost **Internal oblique:** No # in suspension inverted row vs inverted row / No # in TRX® ghost vs TRX® one-arm ghost **External oblique:** No # in suspension inverted row vs inverted row / No # in TRX® ghost vs TRX® one-arm ghost **Erector spinae:** No # in suspension inverted row vs inverted row /↑* 18% in suspension inverted row vs suspension inverted row (TRX® angle 1 and angle 2) / No # in TRX® ghost vs TRX® one-arm ghost

Byrne et al. (2014)	SD: Three conditions of prone bridge in TRX® (arms in suspension, legs in suspension and, arms and legs in suspension FL: prone bridge	**Rectus abdominis:** ↑* 31.6% in suspension prone bridge (legs) vs prone bridge / ↑* 20% in suspension prone bridge (arms) vs suspension prone bridge (legs) / ↑* 20% in suspension prone bridge (arms and legs) vs suspension prone bridge (legs) / No # in suspension prone bridge (arms) vs suspension prone bridge (arms and legs) **External oblique:** ↑* 14% in suspension prone bridge (legs) vs prone bridge / ↑* 15% in suspension prone bridge (arms) vs suspension prone bridge (legs) /↑* 10% in suspension prone bridge (arms and legs) vs suspension prone bridge (legs) / No # in suspension prone bridge (arms) vs suspension prone bridge (arms and legs) **Rectus femoris:** ↑* 8% in suspension prone bridge (arms) vs suspension prone bridge (legs) / ↑* 10% in suspension prone bridge (arms) vs prone bridge **Serratus anterior:** ↑* 12% in suspension prone bridge (legs) vs suspension prone bridge (arms) / ↑* 11% in suspension prone bridge (legs) vs suspension prone bridge (arms and legs)
Snarr and Esco (2014)	SD: Two conditions of prone bridge in TRX® (arms in suspension; legs in suspension) DM: two conditions of prone bridge in swiss ball (arms on swiss ball; legs on swiss ball) FL: Prone bridge	**Rectus abdominis:** ↑* 34.2% in suspension prone bridge (arms) vs prone bridge, and swiss ball prone bridge (legs) **External oblique:** ↑* 42.5% in swiss ball prone bridge (legs) vs prone bridge / ↑* 25.4% in swiss ball prone bridge (legs) vs swiss ball prone bridge (arms) / ↑* 21% in swiss ball prone bridge (legs) vs suspension prone bridge (legs) / No # in swiss ball prone bridge (legs) vs suspension prone bridge (arms) **Erector spinae:** ↑* 4.8% in swiss ball prone bridge (arms) vs suspension prone bridge (legs) / ↑* 8.6% in suspension prone bridge (arms) vs suspension prone bridge (legs) / ↑* 6.5% in swiss ball prone bridge (legs) vs prone bridge / ↑* 2.7% in suspension prone bridge (legs) vs prone bridge /↑* 11.3% in suspension prone bridge (arms) vs prone bridge
Atkins et al. (2015)	SD: Prone bridge (arms in suspension) DM: prone bridge with arms on swiss ball FL: Prone bridge	**Rectus abdominis:** ↑* 34.56% in suspension prone bridge vs swiss ball prone bridge **External oblique:** ↑* 12.03% in prone bridge vs swiss ball prone bridge, and suspension prone bridge **Erector spinae:** No # in prone bridge conditions (SD, DM, and FL)
Malliaropouls et al. (2015)	SD: Hamstring curl with TRX® DM: curl with elastic band and fitball flexion FL: lunge, deadlift, kettlebell swing, bridge, hamstring bridge, hamstring curl, nordic and side leg	**Biceps femoris:** ↑* 69% in suspension hamstring curl vs lunge / ↑* 62% in suspension hamstring curl vs deadlift / ↑* 60% in suspension hamstring curl vs kettlebell swing / ↑* 44% in suspension hamstring curl vs bridge / ↑* 67.37% in fitball flexion vs hamstring curl (SD and FL)/ ↑* 99.37% in side leg vs hamstring curl (SD and FL) / ↑* 28% in suspension hamstring curl vs hamstring bridge /↑* 34% in suspension hamstring curl vs curl and nordic **Semitendinosus:** ↑* 55% in suspension hamstring curl vs lunge / ↑* 45% in suspension hamstring curl vs deadlift /↑* 35% in suspension hamstring curl vs kettlebell swing /↑* 30% in suspension hamstring curl vs bridge / ↑* 41.38% in fitball flexion vs hamstring curl (SD and FL)/ ↑* 91.38% in side leg vs hamstring curl (SD and FL) / ↑* 15% in suspension hamstring curl vs hamstring bridge and nordic / ↑* 16% in suspension hamstring curl vs curl

SD _ Suspension Device; FL _ Floor; DM _ Destabilizing Material; %MVIC _ percent of maximum voluntary isometric contraction; ↑* _ Significantly increases ($p<0.05$); ↓*_ significantly reduces ($p < 0.05$); # _ difference

Figure captions

Figure 1. Suspension training devices and their main features: **a)** traditional suspension device, **b)** pulley suspension device and **c)** suspension device with two parallel chains

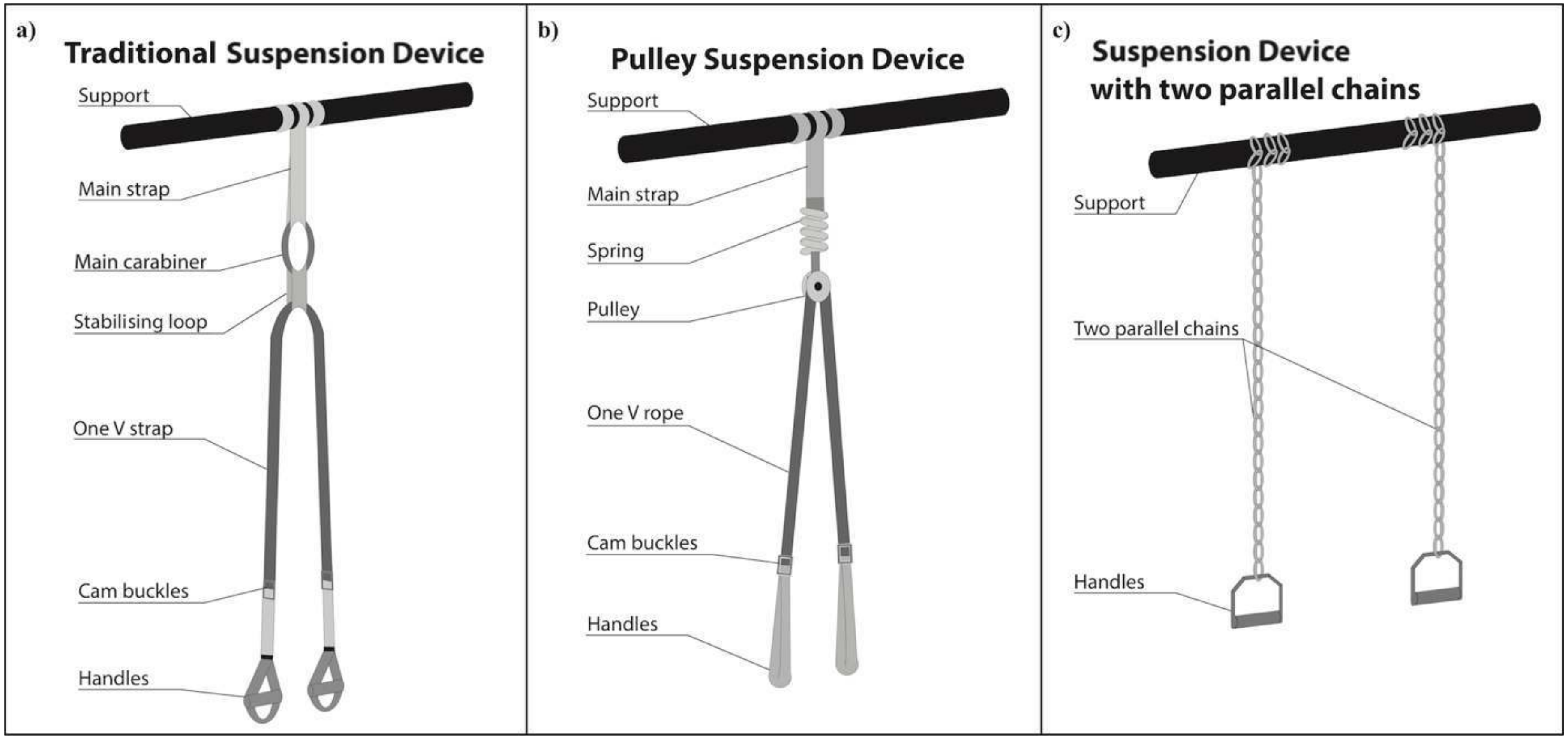
Figure 2. PRISMA flowchart of the search and study selection

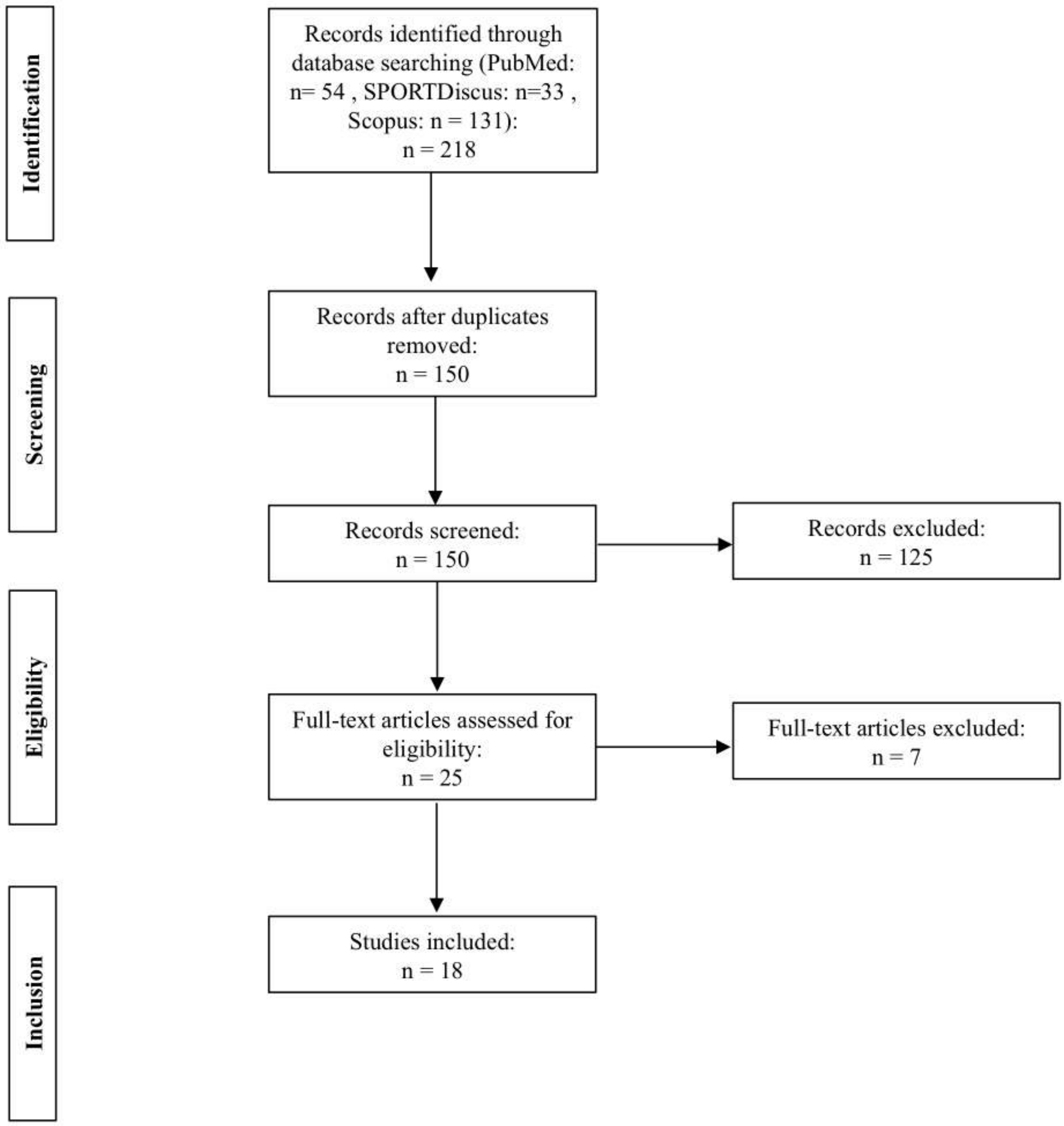
Figure 3. Percentage of maximum voluntary isometric contraction achieved for each muscle in suspension push-ups studies. REF_1: Beach et al. (2008); REF_2: Snarr and Esco (2013b); REF_3: Snarr et al. (2013); REF_4: McGill et al. (2014a); REF_5: Calatayud et al. (2014a); REF_6: Calatayud et al. (2014b); REF_7: Calatayud et al. (2014c); REF_8: Mok et al. (2014); REF_9: Borreani et al. (2015a); REF_10: Borreani et al. (2015b); REF_11: Fong et al. (2015)

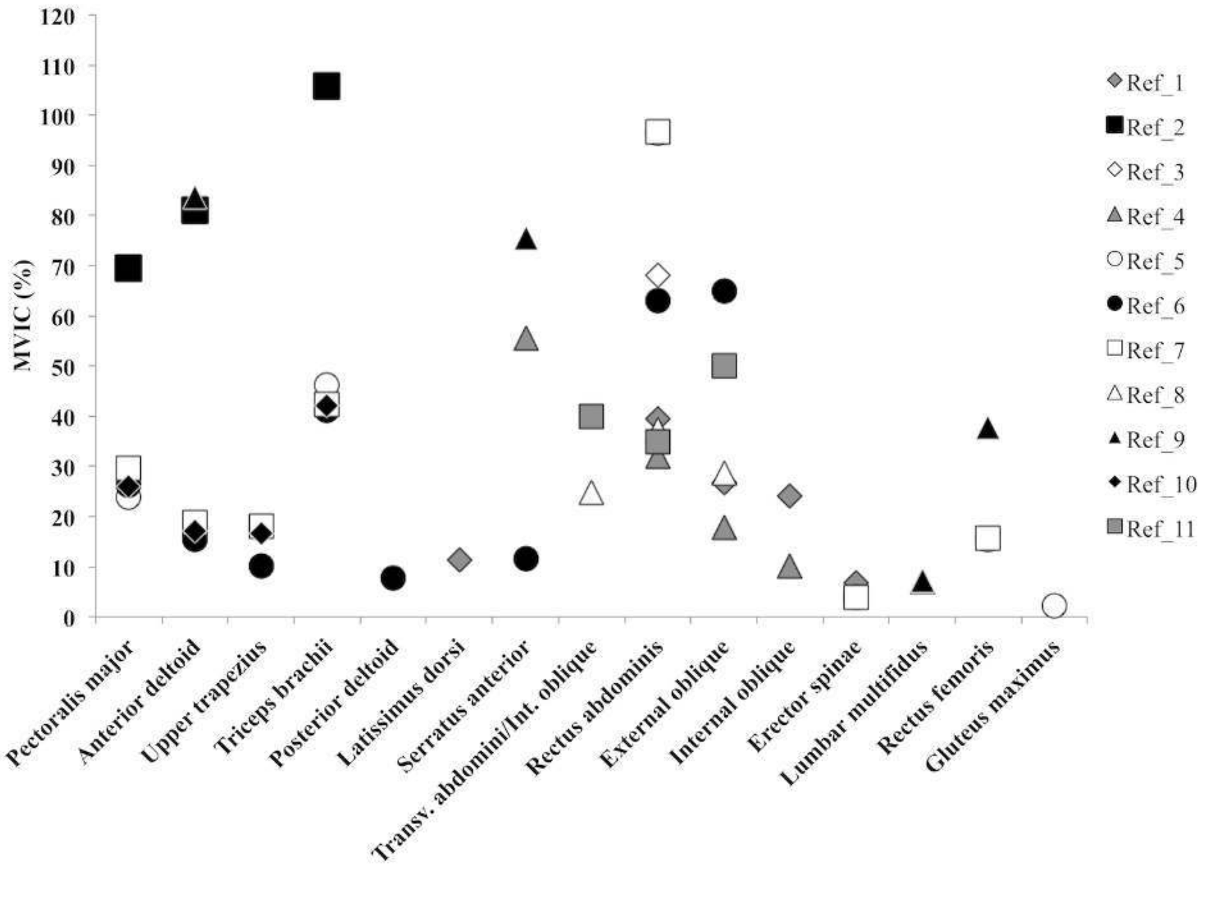
Figure 4. Percentage of maximum voluntary isometric contraction achieved for each muscle in suspension inverted row studies. REF_1: Snarr and Esco (2013a); REF_2: Snarr, Nickerson et al. (2014); REF_3: McGill et al. (2014b); REF_4: Mok et al. (2014); REF_5: Fong et al. (2015)

Figure 5. Percentage of maximum voluntary isometric contraction achieved for each muscle in suspension prone bridge studies. REF_1: Byrne et al. (2014); REF_2: Mok et al. (2014); REF_3: Snarr and Esco (2014); REF_4: Atkins et al. (2015); REF_5: Fong et al. (2015)

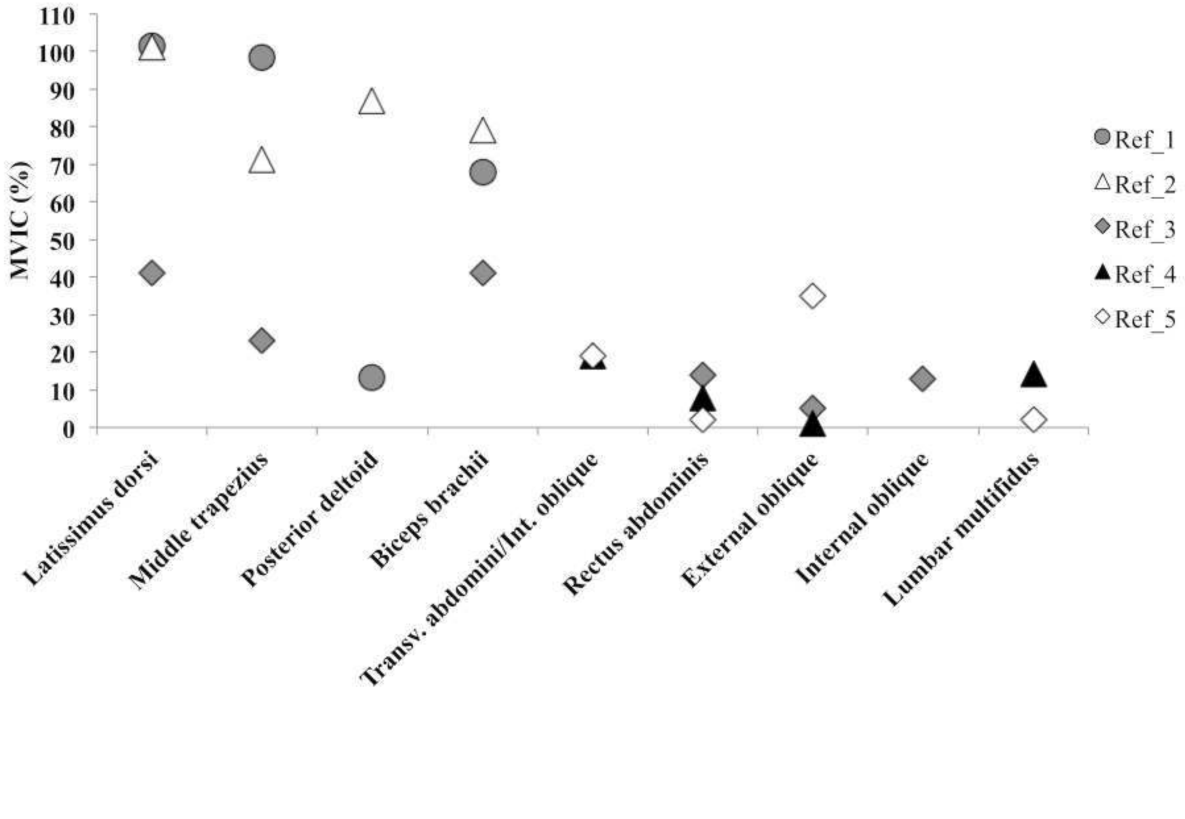
Figure 6. Percentage of maximum voluntary isometric contraction achieved for each muscle in suspension hamstring curl studies. REF_1: Mok et al. (2014); REF_2: Fong et al. (2014); REF_3: Malliaropoulos et al. (2015)


a)
Traditional Suspension Device
Support
Main strap
Main carabiner
Stabilising loop
One V strap
Cam buckles
Handles
b)
Pulley Suspension Device
Support
Main strap
Spring
Pulley
One V rope
Cam buckles
Handles
c)
Suspension Device
with two parallel chains
Support
Two parallel chains
Handles

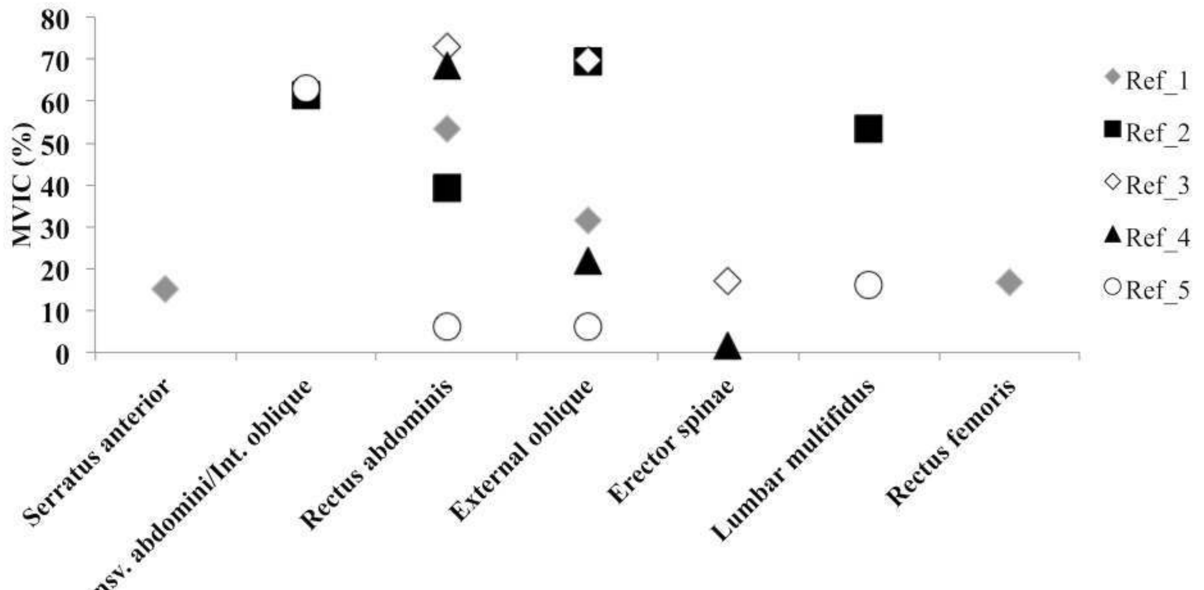

Identification
Records identified through database searching (PubMed: n= 54 , SPORTDiscus: n=33 , Scopus: n = 131): n = 218
Screening
Records after duplicates removed: n = 150
Records screened: n = 150
Records excluded: n = 125
Eligibility
Full-text articles assessed for eligibility: n = 25
Full-text articles excluded: n = 7
Inclusion
Studies included: n = 18

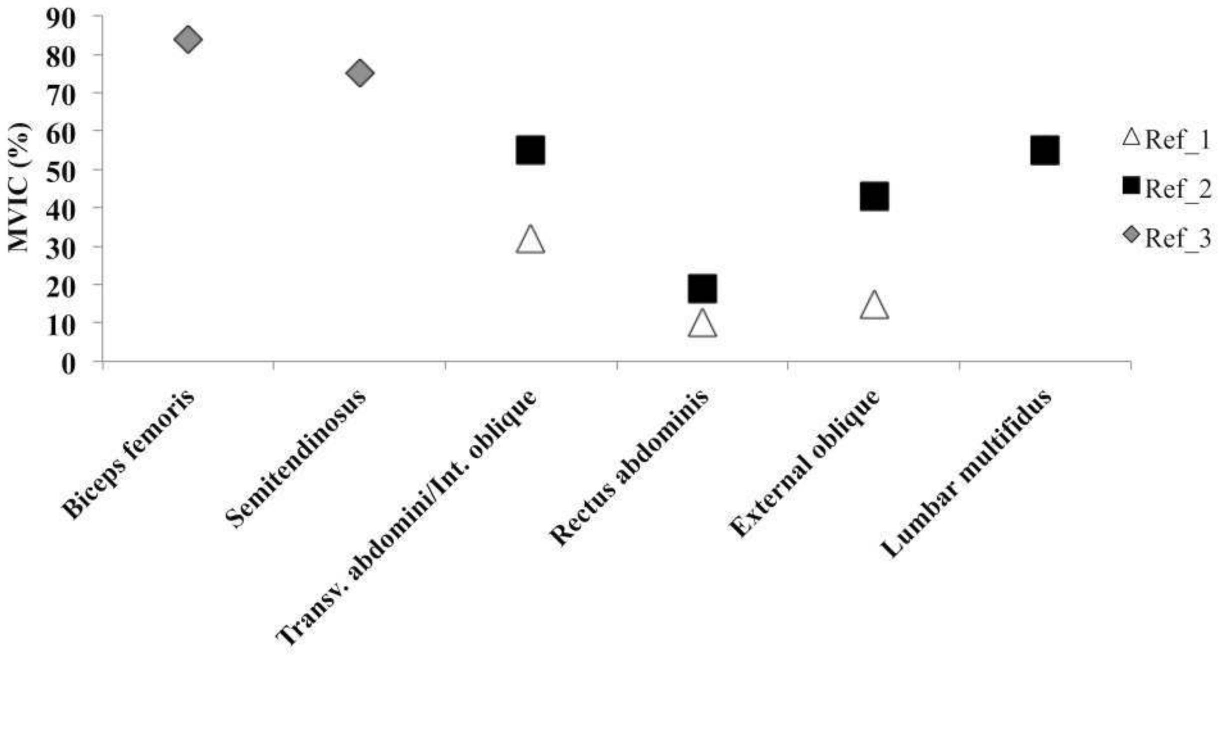


MVIC (%)
120
110
100
90
80
70
60
50
40
30
20
10
0
Pectoralis major
Anterior deltoid
Upper trapezius
Triceps brachii
Posterior deltoid
Latissimus dorsi
Serratus anterior
Transv. abdomini/Int. oblique
Rectus abdominis
External oblique
Internal oblique
Erector spinae
Lumbar multifidus
Rectus femoris
Gluteus maximus
Ref_1
Ref_2
Ref_3
Ref_4
Ref_5
Ref_6
Ref_7
Ref_8
Ref_9
Ref_10
Ref_11



110
100
90
80
70
60
50
40
30
20
10
0
MVIC (%)
Ref_1
Ref_2
Ref_3
Ref_4
Ref_5
Latissimus dorsi
Middle trapezius
Posterior deltoid
Biceps brachii
Transv. abdomini/Int. oblique
Rectus abdominis
External oblique
Internal oblique
Lumbar multifidus


MVIC (%)
80
70
60
50
40
30
20
10
0
Serratus anterior
Transv. abdomini/Int. oblique
Rectus abdominis
External oblique
Erector spinae
Lumbar multifidus
Rectus femoris
Ref_1
Ref_2
Ref_3
Ref_4
Ref_5


MVIC (%)
90
80
70
60
50
40
30
20
10
0
Biceps femoris
Semitendinosus
Transv. abdomini/Int. oblique
Rectus abdominis
External oblique
Lumbar multifidus
Ref_1
Ref_2
Ref_3

Study 2

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Title: Suspended lunge exercise: assessment of forces in different positions and paces

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Suspended lunge exercise: assessment of forces in different positions and paces

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Suspended lunge exercise: assessment of forces in different positions and paces

Summary. *The forces exerted on a suspension device have been examined in the upper body exercises such as push-ups or inverted row. However, there is a lack of evidence with regard to the effects of a suspension device on force production in lower limb exercises. For this reason, this aim of this study was to determine the effects of body position, contraction patterns and pace on force production by the lower limb during the execution of suspended lunge exercises. Ten physically active male university students (n = 10, age = 23.70±2.83 years old) performed sixteen suspended lunges in four different positions and at three different paces (60, 70, and 80 beats per minute). A load cell was used to assess the forces exerted on the suspension device. Force data were analysed with factorial repeated measurements (ANOVA). Significant main effects for position in concentric force (p= .000), average force (p= .002), and for frequency in peak force (p= .004) were found. Peak force was significantly higher in all positions for dynamic contraction type than for isometric suspended lunge. In conclusion, a greater distance of the feet, frequencies around 70 beats per minute and the dynamic contraction type all contributed to enhancing the forces exerted on the suspension strap in the performance of the lunge exercise.*

Key words: *Suspension training; lower-body strength; strain gauge*

L'exercici de lunge en suspensió: valoració de les forces en diferents posicions i ritmes

Resum. *Les forces exercides sobre un dispositiu de suspensió han estat examinades en exercicis com les flexions de braços o el rem invertit. No obstant això, hi ha una manca d'evidències investigant l'efecte dels dispositius de suspensió sobre la producció de força a l'extremitat inferior. Per aquesta raó, l'objectiu de l'estudi va ser determinar els efectes de la posició corporal, els règims de contracció i la velocitat d'execució sobre la producció de força de l'extremitat inferior durant l'exercici del lunge en suspensió. Es van reclutar joves universitaris físicament actius (n= 10, edat = 23.70±2.83 anys) per fer setze lunges en suspensió en quatre posicions i tres ritmes diferents (60, 70, i 80 batecs per minut). Es va utilitzar una cèl·lula de força per valorar les forces exercides sobre el dispositiu de suspensió. Les dades de força es van analitzar amb l'ANOVA factorial de mesures repetides. Es va obtenir un efecte principal per la posició en la força concèntrica (p= .000), força mitjana (p= .002), i per la freqüència en el pic de força (p= .004). El pic de força va ser significativament més alt durant la contracció dinàmic en comparació amb la isomètrica en totes les posicions. Les posicions amb amplituds més grans entre cames, freqüències al voltant dels 70 batecs per minut i el règim de contracció dinàmic milloren les forces exercides sobre el dispositiu de suspensió en l'exercici de lunge.*

Paraules clau: *Entrenament en suspensió; força tren inferior; galga de força*

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Introduction

In recent years, the use of instability resistance training in the field of sports training and fitness activities has been gradually gaining prominence over more traditional resistance training. The main reasons for the method's increased popularity include its simplicity, (as it requires nothing more than body weight in terms of load), its specificity, and its high degree of transferability to actions in sporting competition. The use of unstable surfaces has been shown to be effective in the past for use in untrained populations, satisfactorily replacing the use of external loads to achieve strength benefits (Sparkes & Behm, 2010; Tomljanovic, Spasic, Gabrilo, & Uljevic, 2011). Strength gains in bench press and back squat exercises have been shown to be associated with increases in one-repetition maximums when performed under unstable conditions, after six to eight weeks of training (Marinkovic, Bratic, Ignjatovic, & Radovanic, 2012; Maté-Muñoz, Monroy, Jodra Jiménez, & Garnacho-Castaño, 2014). Nevertheless, when comparing force outputs for exercises under unstable conditions to data for exercises done on even surfaces, the values are lower for instability exercises like the squat or the deadlift (Chulvi-Medrano et al., 2010; Saeterbakken & Fimland, 2013). With regard to muscle activation, unstable surfaces demand a higher degree of activity in most of the muscle groups participating in exercise than is the case on stable surfaces (Anderson, Gaetz, Holzmann, & Twist, 2013; Escamilla et al., 2010). This increased muscle activation, caused by increased instability, can be achieved by reducing the contact area and contact points on the surface, thus leading to greater reliance on the core muscles when there are larger demands for stabilization and balance (Snarr, Hallmark, Nickerson, & Esco, 2016). For this reason, both in the field of strength and conditioning and in rehabilitation, the use of suspension devices (SD) has become more commonplace, due to the ability of these devices to help boost muscle activity and increase exercise demands.

Suspension training (ST) generates instability by using an SD consisting of two straps joined by a single-point anchor with two handles at its ends similar to rings. The degree of instability created by this SD and its effects on muscle activation have been analysed in exercises such as push-ups (Calatayud et al., 2014; McGill, Cannon, & Andersen, 2014a), inverted rows (McGill, Cannon, & Andersen, 2014b; Melrose & Dawes, 2015) and the plank (Atkins et al., 2015; Byrne et al., 2014), all of which feature a great deal of involvement of the upper body muscles. However, there is less data when it comes to the exercises used to strengthen the lower extremities, among which lunges (the traditional exercise done without instability) and variations on them are some of the most widespread training activities (McCurdy, Langford, Cline, Doscher, & Hoff, 2004). A lunge is a unilateral, closed kinetic chain exercise. It is functional and involves the use of multiple joints (Boudreau et al., 2009; Jönhagen, Halvorsen, & Benoit, 2009). Lunges involve the constant activation of the gluteus, quadriceps, hamstrings and the triceps surae during the different contraction regimes (Boudreau et al., 2009). Several studies have recorded muscular activity during the execution of lunges and their variations in order to quantify the load supported by the forward leg during these exercises (Ekstrom, Donatelli, & Carp, 2007; Jönhagen, Ackermann, & Saartok, 2009). These studies, however, did not analyze the implications of differences in positioning or of variations in pace for the execution of a single exercise. The literature provides evidence that contraction types and execution speeds are parameters that should be taken into account when modifying lunge characteristics. Jakobsen, Sundstrup, Andersen, Aagaard, and Andersen (2013) examined lunges to test the effects of differences in execution speed (low vs. high) on the levels of activation of the muscles involved. Prior research on ST has also examined the effects of positioning, contraction regimes and speed of execution in different exercises. Borreani et al. (2015) and Calatayud et al. (2014) analysed the activation of the upper extremity and core muscles during the execution of suspended push-ups at different heights (10 cm and 65 cm), observing that the level of muscular activation was higher when push-ups were performed in suspension at 10 cm from the floor than when they were done at a height of 65 cm. Similarly, McGill et al. (2014a, 2014b) found that as they modified the angle of inclination of push-ups and inverted rows by increasing the strap length, the degree of activation of the muscles analysed increased significantly.

In ST research, the effects brought about by changing body positions and by varying contraction regimes and execution speeds are traditionally assessed using measurements of force and muscular activity. To evaluate muscle activation, electromyography (EMG) is commonly used, often in the form of Maximum Voluntary Isometric Contraction (MVIC) tests, but the use of dynamometers or strain gauges to measure the strain forces constitute a more affordable and practical way to evaluate force production, because ST exercise use body weight and the initial moment of inertia to generate muscle demands (Gulmez, 2017; Melrose & Dawes, 2015). Therefore, the magnitude of the forces generated depends on the degree of instability caused by the SD and body position (Maté-Muñoz et al., 2014). Melrose and Dawes (2015) used a dynamometer to quantify load when performing an isometric suspended inverted row. To modify body position, they used four different inclination angles. The researchers observed that as the angle of inclination between the subject and the ground increased (from 30º to 75º), body mass resistance also augmented from 37.4% to 79.4%. Furthermore, Gulmez (2017) used a strain gauge or load cell to gather data on body mass resistance at different angles of inclination during participants' execution of a number of SD push-ups . Body mass resistance was found to increase (from 36.8% to 75.3% in elbow flexion and from 11.9% to 50.4% in elbow extension) when the angle of the TRX straps was modified (from 45º to 0º).

To the best of our knowledge, quantification of body mass resistance has only been calculated in a few upper-body ST exercises, such as push-ups and inverted row exercises (Gulmez, 2017; Melrose & Dawes, 2015). For this reason, it seemed worth examining the quantification of loads in different lower body ST exercises. Therefore, this research was undertaken with the main aim of determining the lower limb force production during suspended lunge (SUL) exercises, and secondarily to compare force production achieved with different body positions (thanks to alterations in the height of the suspension device and in the distance between the suspended lower limb and the support lower limb), different contraction types (isometric and dynamic) and different paces of exercise (60, 70 and 80 beats per minute (bpm)). The first hypothesis was that the strength production of the lower limb in suspension would increase along with as the height and distance between the suspension device and the lower limb in contact with the ground. Secondly, it was hypothesised that increasing execution speeds would also be associated with increases in the force produced during dynamic actions (60, 70 and 80 bpm) and that these dynamic contractions require a greater production of force than isometric movements. The final hypothesis was that body mass resistance would significantly increase as the distance between the participants' feet and the height of the suspended foot increase.

Methods

Design

A repeated measures design was used to compare force production under 16 different conditions of the SUL exercises and thus to determine if the resulting force increased when the position, the contraction regime, and the execution speed were modified. The SUL exercise was carried out using the TRX Suspension trainer™ device. An S-Type Load Cell force sensor was used to obtain strength extension values for the suspended lower extremity. To determine the effect of SD on force production, four body positions were used (TRX_height with respect to the floor$_{(cm)}$-distance between lower support limb and in suspension $_{(cm)}$): 1) TRX_40-60, 2) TRX_40-80, 3) TRX_60-60, and 4) TRX_60-80. In each of these positions, one isometric and 3 dynamic SULs were performed at 60, 70 and 80 bpm.

Participants

Ten healthy and physically active male subjects (mean age = 23.70 ± 2.83 years, height = 1.83 ± .043 m, weight = 79.30 ± 10.85 kg, body mass index = 23.58 ± 2.42 kg·m^{-2}, thigh length = 40.90 ± 2.02 cm, leg length = 53.40 ± 2.46 cm) were voluntarily recruited for the study. All the participants not presenting a height between 1.77 m and 1.87 m, as well as those who presented diseases and/or pain related to the cardiovascular, musculoskeletal and neuromuscular apparatuses, were excluded. All the participants received clear information on the research protocol and signed the informed consent, having previously read the information document. The ethics committee of the Ramon Llull University of Barcelona approved the development of this study, which was conducted in accordance with the latest version of the declaration of Helsinki (revised in Fortaleza, Brazil, 2013).

Procedures

The research was carried out in two sessions. The familiarisation and the experimental sessions were held the same day of the week. All the exercise conditions and the order of the participants were randomly assigned. During the familiarisation session, the anthropometric data on the different subjects were collected, and each participant was also asked to identify his dominant leg, defined here as the participant's preferred kicking leg (Meylan, Nosaka, Green, & Cronin, 2010). The length of the thigh and the leg were also measured, in accordance with the ISAK (2001) protocol. Additionally, in this session participants were given training in the correct mechanics of SUL exercises to be done under all the analysis conditions.

Participants were asked not to perform any intense physical activity during the 12-hour span prior to the experimental sessions. Participants performed the SUL with the TRX Suspension trainer (Fitness Anywhere, San Francisco, CA), with a distance of 40 cm and 60 cm between the ground and the device's strap. The TRX® anchor point was located on the ceiling at a height of 2.95 m from the ground. Participants performed the 16 variations of the movement in two blocks (first block TRX_40 cm and second block TRX_60 cm). During each block, the distance between the lower limbs, contraction types (isometric and dynamic) and the execution speeds (60, 70 and 80 bpm) were modified. Participants performed a set of three repetitions of the isometric SUL (three seconds at the top position, ten seconds at the bottom position) and a set of five repetitions of the dynamic SUL with 90 seconds of rest between each SUL condition. The execution speed in the dynamic SUL was established using a metronome (the app *Pro Metronome,* version 3.13.2; EUMLab-Xannin Technology Gmbh., Hangzhou, China).

Standarisation of the SUL exercises was ensured by asking the participants to cross their arms over their chests and to keep their trunks in a neutral position. The heel of the lower extremity in contact with the ground had to be placed in front of the marks indicating the different distances (60 cm and 80 cm). The sole of this foot had to be completely flat on the ground. The foot of the non-dominant lower extremity was placed inside the device handle, with a slight plantar flexion of the ankle joint (Figure 1). To obtain better control of the SUL movement range, a WSB 16k-200

Figura 1. Standardised position during suspended lunge.

position encoder was used (ASM Inc., Moosinning, DEU), and all the SUL exercises were recorded at 30 fps with an iPhone 6 Plus (Apple Inc., Cupertino, CA). To control the movement of the suspended lower limb and the participants' trunk movements and to prevent the knee from going past the toes during the flexion movement, five plastic poles were used as reference marks. They were placed at a distance of 40 cm from one another and aligned with the camera's line of vision (Payton, 2008).

If a participant's execution of any of the SUL was incorrect, researchers asked him to stop the exercise, and then (respecting the 90 seconds of rest between sets) to perform the SUL again. However, only those repetitions of SULs that met the standard criteria for the movement were analysed.

Measures

The forces exerted on the suspension strap during SULs were assessed with a 200 HZ S-Type Load Cell strain gauge CZL301C (Phidgets Inc., Alberta, CAN), which was placed between the anchorage point and the TRX. The load cell was calibrated following the manufacturer's recommendations. According to Tiainen et al. (2004) and Vivodtzev et al. (2006), force sensors are a valid and reliable tool to measure muscle strength. The data gathering process was performed using the DA 100C transducer force sensor (BIOPAC System, INC., Goleta, CA) connected to the BIOPAC MP-150 (BIOPAC System, INC., Goleta, CA). The information was transferred to AcqKnowledge software (Version 4.2 for Windows 7; BIOPAC System, INC., Goleta, CA), where a force/time curve was displayed. The average recorded force was calculated for each of the repetitions in the concentric phase, the eccentric phase and the concentric/eccentric phases. The variables of concentric force, eccentric force and average force were expressed as the mean total of the average force of each of the repetitions in the concentric phase, the eccentric phase and the concentric/eccentric phase, respectively. The peak force variable was calculated for both the dynamic SUL and the isometric SUL and was expressed as the average value of the maximum force production in each of the repetitions (five reps for dynamic suspended and three reps for isometric suspended). Additionally, the peak force of the isometric conditions was normalised for each participant using the following equation: load norm (%) = load /body weight x 100 (Gulmez, 2017). The normalised values were expressed as a percentage of the total load.

Statistical Analysis

Descriptive statistical analysis and frequencies were used to describe the sample. Descriptive statistical methods were used to calculate the mean and the standard deviations. To test the normality of the sample, the Shapiro-Wilk hypothesis test for samples of below 50 subjects was performed. The number of subjects recruited was based on effect size 0.4 SD with an α level of .05 and power at .95, calculated with the G Power Software (University of Dusseldorf, Germany). The analysis of the factorial variance (Position [height_distance_TRX] X Frequency [60, 70 and 80 bpm] and Position [height_distance_TRX] X Contraction type [dynamic and isometric]) of repeated measurements (ANOVA) was used. The Greenhouse-Geisser correction was applied when sphericity was violated (Mauchly's Test). A One-way ANOVA test was used to compare the body mass resistance (kg) recorded in the different positions under isometric conditions. In both analyses, Post-hoc analyses with Bonferroni corrections were carried out when significant effects were assumed. The effect of the size was indicated with partial eta squared (η_p^2), with cut values of .01, .06, and .14 for a small, medium and large effect, respectively (Cohen, 1988). The significance level was established at $p<.05$. The results were expressed using mean ± standard deviation. Statistical analysis were carried out using SPSS version 20.0 for Mac (SPSS Inc., Chicago, IL, USA).

Results

Table 1 shows results of the mean (±SD) SUL dynamic force production in each position (TRX_40-60, TRX_40-80, TRX_60-60 and TRX_60-80) and frequency (60, 70 and 80 bpm) in the different variables of the study: concentric force, eccentric force, average force and peak force.

Concentric force

A significant main effect was found for position [$F_{(3, 27)}$ = 8.284, p = .000, η_p^2 = .47], but no such effect was found for frequency [$F_{(1.28, 11.50)}$ = .854, p = .442, η_p^2 = .08], nor for interaction [$F_{(6, 54)}$ = .663, p = .681, η_p^2 = .06]. Pairwise comparisons showed significant differences between TRX_40-60 and TRX_60-80 (p = .008)

Table 1. Force production (*N*) during dynamic suspended lunge at four different positions and three different frequencies. Values showed in mean ± SD

		Dynamic frequency				
		60 bpm	70 bpm	80 bpm	Interaction effect	
	Position	Mean ±SD	Mean ±SD	Mean ±SD	*p* (*p*<.05)	η_p^2
Concentric Force	TRX_40-60	116.21±37.15	115.81± 32.61*	112.97±39.06	.681	.06
	TRX_40-80	120.23±33.02	122.98±38.40	122.99±43.88		
	TRX_60-60	114.74±32.41	117.40±36.83†	119.48±41.34		
	TRX_60-80	123.42±36.87	131.04±36.84* †	128.41±35.22		
Eccentric Force	TRX_40-60	159.47±43.69	160.18±41.64	156.53±49.67	.467	.08
	TRX_40-80	160.89±47.19	166.90±45.78	163.52±50.31		
	TRX_60-60	156.82±38.38	167.05±47.04	164.49±52.96		
	TRX_60-80	162.57±46.37	178.31±48.68	166.87±47.04		
Average Force	TRX_40-60	130.56±39.94	130.03±36.11*	126.62±42.91 ¶	.575	.08
	TRX_40-80	133.26±37.77	136.50±39.56	136.07±45.94 ¶		
	TRX_60-60	128.57±34.06	132.45±40.45 †	133.96±45.63		
	TRX_60-80	136.48±40.84	146.43±42.06* †	141.21±39.54		
Peak Force	TRX_40-60	205.85±63.40	215.85±64.12	221.94±83.44	.607	.06
	TRX_40-80	207.84±61.82 §	223.14±78.70	233.14±78.70 §		
	TRX_60-60	199.14±51.15	221.50±67.47	226.45±81.88		
	TRX_60-80	210.63±61.60 ✝	233.24±68.04 ✝	229.65±72.97		

Notes:
(N) = Newton; bpm = Beats per minute
* Significant differences between TRX_40-60 and TRX_60-80
†Significant differences between TRX_60-60 and TRX_60-80
¶ Significant differences between TRX_40-60 and TRX_40-80
✝ Significant differences between frequency 60 bpm and 70 bpm
§ Significant differences between frequency 60 bpm and 80 bpm

and between TRX_60-60 and TRX_60-80 (p = .021) at the frequency of 70 bpm (Table 1).

Eccentric force

No significant main effects were found for position [$F_{(3, 27)} = 2.562$, $p = .076$, $\eta_p^2 = .22$], frequency [$F_{(2, 18)} = 3.466$, $p = .053$, $\eta_p^2 = .27$] or interaction [$F_{(2.41, 21.70)} = .834$, $p = .467$, $\eta_p^2 = .08$].

Average force

A significant main effect was found for position [$F_{(3, 27)} = 6.565$, $p = .002$, $\eta_p^2 = .42$], but no such effect was found for frequency [$F_{(2, 18)} = 1.174$, $p = .332$, $\eta_p^2 = .11$], nor for interaction [$F_{(6, 54)} = .799$, $p = .575$, $\eta_p^2 = .08$]. Pairwise comparisons showed significant differences between TRX_40-60 and TRX_60-80 (p = .007), between TRX_60-60 and TRX_60-80 (p = .020) at the frequency of 70 bpm (Table 1). Furthermore, significant differences were found between TRX_40-60 and TRX_40-80 (p = .036) at the frequency of 80 bpm (Table 1).

Peak force

A significant main effect was found for frequency [$F_{(1.22, 11.04)} = 7.776$, $p = .004$, $\eta_p^2 = .46$] but no such effect was found for position [$F_{(3, 27)} = 1.946$, $p = .146$, $\eta_p^2 = .17$], nor for interaction [$F_{(2.68, 24.18)} = .594$, $p = .607$, $\eta_p^2 = .06$]. Pairwise comparisons showed significant differences ($p < .05$) between 60 bpm and 80 bpm at TRX_40-80 (p = .035), and between 60 bpm and 70 bpm at TRX_60-80 (p = .006) (Table 1).

Figure 2 shows the comparison of force production in SULs by the type of contraction (isometric or dynamic) and by body position (TRX_40-60, TRX_40-80, TRX_60-60 or TRX_60-80), in peak force. A significant main effect was found for contraction type [$F_{(1, 36)} = 52.346$, $p = .000$, $\eta_p^2 = .59$], but not for the interaction effects [$F_{(3,36)} = .862$, $p = .469$, $\eta_p^2 = .07$]. Pairwise comparisons showed significantly greater peak force in the dynamic SUL than the isometric exercise at TRX_40-60 (p = .003), TRX_40-80 (p = .000), TRX_60-60 (p = .001) and TRX_60-80 (p = .009).

A significant effect was found for position in body mass resistance in isometric SUL [$F_{(3, 36)} = 21.103$, $p = .000$, $\eta_p^2 = .64$]. Pairwise comparison showed significantly higher percentages of body mass resistance in isometric SUL for the position TRX_40-60 (20.00% ± 6.25) than for TRX_40-80 (10.21% ± 1.21, p = .000), TRX_60-60 (10.07% ± 1.21, p = .000) and TRX_60-80 (11.02% ± 1.27, p = .000).

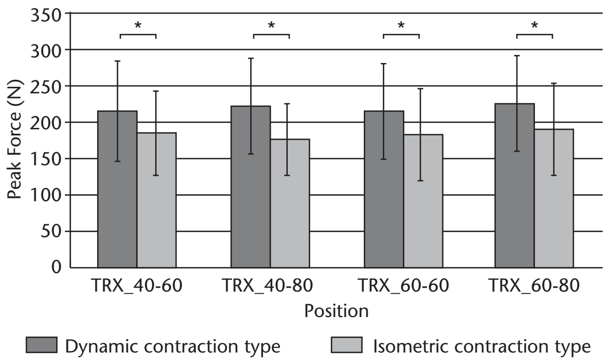

Figura 2. Peak force comparison between dynamic and isometric suspended lunge at four different positions. Each bar represents the mean, and the error bar the standard deviation (SD). *Note:* *Significant differences (*p<.05*) between dynamic and isometric contraction type

Discussion

The main finding of the present study was that position significantly affected average and concentric force production. Force production increased as the distance of the feet and the height of the suspended foot increased. Moreover, the pace of repetitions of a dynamic lunge exercise affected the peak force, but not the other variables analysed. The peak force was significantly higher in the dynamic SUL than the isometric exercise. In addition, body mass resistance was significantly greater in the closest body position in the isometric SUL (TRX_40-60).

Our findings showed that TRX_60-80 position elicited greater force production than TRX_40-60 for all the analysed variables (concentric, eccentric, average and peak force variables). This finding is in accordance with those of authors who have reported greater body mass resistance from position 1 to position 4 in suspended inverted row exercises (Melrose & Dawes, 2015) and suspended push-ups (Gulmez, 2017). This tendency may be explained by the very body weight and force momentum principles upon which suspended training are based, because the difficulty of ST exercises and the number of motor units recruited depend on the amount of instability caused by the device and the body position (Maté-Muñoz et al., 2014). With regard to strap length, there is a lack of previously published research into force production with a load cell and with strap length variations during ST exercises. However, some evidence suggests there is higher muscle activity when suspension push-ups are performed with a 10 cm strap length from the floor in than is the case in push-ups at 65 cm (Borreani et al., 2015; Calatayud et al., 2014). In contrast, the present study showed that a variation in strap length (40 cm to 60 cm) was not associated with significant differences in force production in any of the variables. This finding suggests that a variation in the strap length of 20 cm probably does not create a sufficient degree of instability to lead to changes in force production. On the other hand, a variation in feet distance (60 cm to 80 cm) did lead to greater force production for all the analysed variables. Furthermore, force production was significantly higher at TRX_40-80 (average force) and TRX_60-80 (concentric and average force) than at TRX_40-60 and TRX_60-60, respectively. Another study reported similar force production patterns when the distance between the TRX anchorage point and the foot fulcrum was increased (in 3 to 6 increments of 30.5 cm) during a suspended inverted row exercise (Melrose & Dawes, 2015). We might speculate that increasing the distance of the feet also increases force production when performing a SUL. Nevertheless, variations in feet distance were smaller than those found by Melrose and Dawes (2015). However, a greater distance of the feet would lead to more instability, which would probably lead to inappropriate technique in the performance of SUL.

The results of this study showed that the pace did not significantly alter force variables (concentric, eccentric, and average force). However, when peak force was analysed, significantly higher force production was achieved between 80 bpm and 60 bpm at TRX_40-80. Likewise, a significantly higher force production was found between 70 bpm and 60 bpm at TRX_60-80. These differences may be explained by the need to apply more force on the SD. Stability needs increase along with the frequency of movement. In fact, the peak force values obtained at 70 bpm and 80 bpm are very similar using TRX_60-80. This finding is in accordance with result obtained by other authors, who have recorded greater muscle activity in those performing lunges under ballistic conditions than those doing the exercises using slow, controlled contractions (Jakobsen et al., 2013). However, the values found among lunge positions are smaller than those reported by LaChance and Hortobagyi (1994) for push-ups and pull-ups. With regard to eccentric force, statistically non-significant results were found. This fact could be explained by the need for controlled braking to keep moving as stable as possible. Nonetheless, our frequency outcomes suggested a trend towards improvement in force production at 70 bpm over the force achieved at 60 bpm. There is no corresponding improvement, though, between 70 bpm and 80 bpm. Although more evidence is necessary, it seems that 70 bpm could be an optimal frequency to stimulate force production in SUL.

The results of the present study showed significant differences in peak force produced using different contraction types and body positions. Peak force was significantly higher when the participants employed the dynamic contraction type, regardless of body position. This finding is largely consistent with those of Jakobsen et al. (2013) and Jönhagen et al. (2009), who conducted studies of a dynamic lunges done at different paces. These authors reported that high velocity is associated with a higher degree of activity of the muscles analysed in the dynamic lunge. In contrast, as Ekstrom et al. (2007) stated that an isometric lunge elicits lower muscle activity. It appears that dynamic contraction leads to greater force production than isometric contraction, likely because an increase in pace leads to greater recruitment of the motor unit, thus increasing muscle activity. However, there is still a lack of research available investigating the effects of contraction type during suspended exercises. Meanwhile, there is some evidence in the literature that the percentage of body mass resistance increases as a result of the position change during isometric suspended push-ups (Gulmez, 2017) and isometric suspended inverted row (Melrose & Dawes, 2015). These authors reported that position 4 (body angle closer to the floor) is associated with a greater percentage of body mass resistance from the TRX strap than the other positions (1,2,3), where the body angle is farther from the floor. The results of the present study stand in contrast to those of Melrose and Dawes (2015) and Gulmez (2017), because our findings show that TRX_40-60 (the closest position) was associated with a greater percentage of body mass resistance than the other positions (TRX_40-80, TRX_60-60, TRX_60-80) during isometric SUL. We expected that body mass resistance would significantly increase as the distance between the feet and the height of the suspended foot increased. Nevertheless, the results suggested that TRX_40-60 provides greater support on the suspension device handles than the other positions. A 9.93% body mass resistance increase between TRX_40-60 and other positions may explain this outcome. Finally, we could speculate that other positions require the application of more force on the forward foot than on the suspended foot, probably due to increases in strap height and feet distance.

There were some limitations associated with this study. Firstly, only the strap length variations were established to modify the degree of instability from the SD. No comparison between different strap angles was conducted. Another limitation of our data was the lack of quantification of force on the forward leg to compare it to the rear (suspended) leg. In future studies the assessment of ground reaction forces with a force platform may be worth of attention. Finally, another limitation may be the lack of a normalised distance in the forward step during lunge execution, as Boudreau et al. (2009) recommended. However, the thigh and leg length were measured following the ISAK (2001) protocol, thus ensuring the homogeneity of the participants.

In conclusion, the results demonstrate that force production is enhanced when an SUL is performed with a distance of 80 cm between the feet. Furthermore, there is also evidence from our results to suggest that greater force production during SUL is associated with the choice of pace (70 bpm) and with the use of a dynamic contraction type. Likewise, the assessment of the suspension training load during a lunge seems to be useful for strength and conditioning coaches wishing to individualise the athletes' load related to lunge position and force production. The variations on suspension lunge positions also allow coaches and practitioners to achieve progress through position difficulty. Performing suspended lunges is a good choice for those seeking to strengthen their lower limbs. The inclusion of this exercise in strength and conditioning programs could be useful for those trying to improve their unilateral sport skills such as jumping, changes of direction, sprinting and shooting. Also, leaning the rear leg on the SD in the lunge exercise allows for the creation of higher demands on the FL, thus increasing strength, power and balance. Apart from the changes in the body position, contraction type and pace, coaches and practitioners could increase the muscular and force demands in the suspended lunge by adding other sources of instability (on the front leg) or extra weights.

Conflict of interest statement

The authors declare that there is no conflict of interest relevant to this study.

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RESEARCH ARTICLE

Muscle activity of Bulgarian squat. Effects of additional vibration, suspension and unstable surface

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Abstract

Practitioners of strength and conditioning are increasingly using vibration and unstable environments to enhance training effects. However, little evidence has been found comparing the use of suspension devices and vibratory platforms used in the Bulgarian squat. The purpose of this cross-sectional study was to examine the effect of suspension devices (TRX®), unstable surfaces (BOSU®), and vibration plates on muscle activity and force during the Bulgarian squat. Twenty physically active male students (age = 24.40 ± 3.63 years) performed a set of five repetitions of Bulgarian squats, suspended lunges, suspended lunges-BOSU, suspended lunges-Vibro30, and suspended lunges-Vibro40 (vibration 30 Hz or 40 Hz and 4 mm of amplitude). A randomized within-subject design was used to compare leg muscle activity, vertical ground reaction forces, and force exerted on the strap across the five exercises. Results showed no significant differences in muscle activity between the Bulgarian squat and suspended lunge ($p = 0.109$, $d = 2.84$). However, the suspended lunge significantly decreased muscle activation compared to the suspended lunge-BOSU ($p = 0.012$, $d = 0.47$), suspended lunge-Vibro30 ($p = 0.001$, $d = 1.26$), and suspended lunge-Vibro40 ($p = 0.000$, $d = 1.51$). Likewise, the Bulgarian squat achieved lower activity than the suspended lunge-Vibro40 ($p = 0.010$, $d = 0.96$). The force on the strap significantly decreased in the suspended lunge-BOSU compared to the suspended lunge-Vibro30 ($p = 0.009$, $d = 0.56$). The suspended lunge achieved higher front leg force production than the Bulgarian squat ($p = 0.006$, $d = 0.48$). In conclusion, leaning the rear leg on a suspension device does not provoke an increase in the activation of the front leg during the Bulgarian squat but increases the vertical ground reaction forces. Thus, the use of unstable surfaces or vibration plates for the front leg increased muscular activity when performing a suspended lunge.

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Introduction

In strength and conditioning, recent trends support the use of functional exercises to improve the efficacy of multidirectional sports skills, enhancing the quality of resistance training. These skills include locomotor, manipulative, and stability actions while maintaining control of the kinetic chain [1]. Most of these actions involve unilateral actions of multidirectional jumping, change of direction, and sprinting using different techniques, with a significant anteroposterior, lateral rotational force-vector application [2–4]. Thus, Bulgarian [5] and single-leg squats [6] or side-steps and backward lunges [2] have been a part of effective sport-specific training programs. Nuñez et al. [7] found significant improvements in a 90˚ change of direction in a unilateral resistance training group compared that in a bilateral training group in team sports. Moreover, Bogdanis et al. [8] showed some evidence supporting the benefits of unilateral resistance training in jumping and rate of force development in physical education students. In the same direction, Gonzalo-Skok et al. [2] demonstrated higher improvements in functional tests (180˚ change of direction, lateral jump, and one-leg horizontal jump) for a unilateral resistance training group in team sports. The same leading author also found a between limb imbalance reduction following this training paradigm in basketball players [9]. Therefore, due to their specificity and transferability to sports skills, the step-up, standard lunge (two feet on the floor), or Bulgarian squat (rear foot elevated) are among the most widely used exercises to enhance lower body strength [10].

Instability

Coaches, athletes, and fitness enthusiasts are continuously searching for new challenges to increase training demands through the complexity of the exercises, for instance, by modifying the amount of instability or intensity [11]. Thus, the use of devices that create instability has become popular (i.e., BOSU® Ball, Wobble Board®). Primarily, unstable devices are used to increase the load of traditional exercises by providing higher muscular demands through superior motor unit recruitment. Such devices also improve neuromuscular coordination to maintain balance during training exercises [12]. As Behm et al. [11] stated, strength training on unstable surfaces or unstable implements provides an augmented degree of instability compared to stable surfaces. Hence, destabilizing environments provide more varied and effective training stimuli, enhancing neuromuscular adaptations [13]. Likewise, some evidence supports the idea that instability training elicits higher activity of several upper body and trunk muscles than traditional exercises such as push-ups, sit-ups, and back extensions. Anderson et al. [14] recruited highly trained individuals to examine triceps brachii, erector spinae, rectus abdominis, internal oblique and soleus activation while performing traditional and unstable push-ups in the single (hands or feet on the unstable surface) or dual (both hands and feet on the unstable surface) condition. The authors found that the dual condition provoked the highest percentage of change (>150%) for all the analyzed muscles than the other conditions. Besides, a significant linear effect was found between the amount of instability provided and level of muscle activity in all muscles and exercise conditions. Cosio-Lima et al. 's study [15] showed that after 5 weeks of sit-up and back extension unstable training (Swiss ball) in untrained college women, muscle activity of rectus abdominis and erector spinae significantly increased compared to that of a control group. Furthermore, some evidence of this has been found in lower body exercises such as standard lunges [16] and Bulgarian squats [17]. Concretely, performing standard lunges and Bulgarian squats involves the activation of the gluteus maximus and medius, vastus medialis, vastus lateralis, rectus femoris, biceps femoris, semitendinosus, and gastrocnemius [18,19]. In order to assess muscle activity during a standard lunge, Boudreau et al. [18] used surface electromyography to measure the activity of rectus femoris, gluteus medius, and gluteus maximus in healthy individuals and demonstrated that the activation of gluteus medius, gluteus maximus, and rectus femoris ranged from low to moderate (from <21% to 40%) maximum voluntary isometric contraction (MVIC). Others authors [19] have examined the effect of performing a Bulgarian squat (loaded) on the activity of gluteus maximus, biceps femoris, semitendinosus, rectus femoris, vastus lateralis, vastus medialis, and gastrocnemius and reported that Bulgarian squats provoked higher muscular recruitment (>638 mV) in the quadriceps muscles (rectus femoris, vastus medialis, and lateralis) than in the hamstrings (biceps femoris and semitendinosus), gluteus maximus, and gastrocnemius (all of them <396 mV). DeForest et al. [19] reported that all analyzed muscles achieved higher activation during the concentric phase than in the eccentric phase.

Regarding the effects of unstable conditions in the lower body, only Andersen et al. [17] examined the effect of performing a standardized Bulgarian squat (6-RM loaded) under stable (front leg on the floor) and unstable (front leg on a foam cushion) conditions on the hip and thigh muscles of healthy trained participants. Bulgarian squats significantly increased the activation of biceps femoris under stable conditions compared to those under unstable conditions (stable vs. unstable: 215.5 ± 106.7% MVIC vs. 193.3 101.5% MVIC, $p = 0.030$), and there were no significant differences for rectus femoris, vastus medialis, vastus lateralis, and gastrocnemius, and all of them achieved a high activation (>60% MVIC) under both exercise conditions. In contrast, Youdas et al. [16] found that surface (stable vs. unstable) and sex have a significant effect on the activations of rectus femoris (women vs. men in stable surface: 33.9% MVIC vs. 20.1% MVIC, respectively; $p = 0.04$) and hamstring (men vs. women in unstable surface: 37.9% MVIC vs. 19.9% MVIC, respectively; $p = 0.04$) during the extension of a standard lunge in healthy recreational athletes. Thus, evidence that the use of unstable surfaces increases muscular demands during Bulgarian squat and standard lunge exercises is weak.

Whole-body vibrations

Other devices such as whole-body vibration (WBV) platforms are commonly used to increase neuromuscular performance in strength training. These platforms modify workloads through vibration (side-alternating vibration or synchronous vibration), frequency (in Hz), and amplitude (peak to peak displacement, in mm) and, as a consequence, the magnitude of acceleration following the muscle tuning paradigm [20,21]. WBV is applied to the muscle or tendon to elicit tonic vibration reflex [22], and the beneficial effects of WBV have been demonstrated in lower limb exercises (squat, half-squat, Bulgarian squat, or lunge) in different cohorts such as untrained, recreationally active, and older adults [23,24]. As for muscle activation, vastus lateralis recruitment increases when performing 60 s of static half-squat with 100˚ of knee flexion at three different WBV frequencies (30, 40, and 50 Hz) with 10 mm of amplitude [25]. Likewise, Di Giminiani [26] reported that performing 20 s of static half-squat in four different positions (knee flexion angle ranging from 90˚ to 120˚) with WBV (45–55 Hz and 1 mm of amplitude) increased the activation of vastus lateralis compared to a half-squat with no vibration applied in male sport sciences students. Moreover, Ritzmann et al. [27] found that a progressive increase in WBV frequencies (from 5 to 30 Hz) and amplitudes (from 2 to 4 mm) causes a progressive increase in the activation of vastus medialis, rectus femoris, and biceps femoris while performing 10 s of static half-squat. Thus, frequencies ranging from 30 to 55 Hz and amplitudes from 2 to 5 mm elicited the highest response in the muscles mentioned above [23,27,28]. Although WBV increases the activation of thigh muscles during lower body exercises, such as the squat, Bulgarian squat, or lunge, there is a rising interest in enhancing muscular activity through the use of different suspension devices. Furthermore, the use of a combination of different methods to increase muscular activation has been investigated

[29–31]. Vibratory platforms, flywheels, rubber bands, or pulley machines have been used together with other devices such as Pielaster®, Swiss Balls, Freeman plates, and BOSU® to create instability. Moras et al. [32] recently compared the variability in force production of a stable and unstable bilateral squat using a flywheel machine and found no significant differences between both conditions in terms of sample entropy in healthy trained men. Nevertheless, combinations of suspension devices with other training methods are still unexplored.

Suspension devices

In suspension training, a suspension device is required to create an unstable condition. This method utilizes a system of straps with handles on the bottom and attached to a single anchor point [33]. This device acts as a pendulum by rotating around the singular anchor point. The suspension device uses body weight and fundamental principles (vector resistance, stability, and pendulum) to enhance motor unit recruitment [34]. The effects of using a suspension device on lower body muscle activity have been investigated while performing a hamstring curl. Specifically, Malliaropoulos et al. [35] examined the effect of ten hamstring loading exercises (standard lunge, single-leg Romanian deadlift T-drop, kettlebell swing, bridge, suspended hamstring curl, hamstring bridge, curl, Nordic exercise, Swiss ball flexion and slide leg exercise) on biceps femoris and semitendinosus recruitment in elite female track and field athletes and reported that the biceps femoris and semitendinosus achieved a very high activation (>60% MVIC) in the suspended hamstring curls compared to the high-to-low activity (<60% MVIC) for the standard lunge, single-leg Romanian deadlift T-drop, kettlebell swing, bridge, hamstring bridge, curl, and Nordic exercise. However, the suspended hamstring curl was less demanding for the biceps femoris (84% MVIC) and semitendinosus, (75% MVIC) than the Swiss ball flexion and the slide leg exercise, both with muscle activity >90% MVIC. Recently, Krause et al. [36] assessed the activation of hip and thigh muscles during a suspended lunge (rear leg leaning on the suspension device cradles) and its counterpart. The suspended lunge exercise achieved significantly higher activation in the hamstring, gluteus maximus, gluteus medius, and adductor longus than the standard lunge. Despite this, the authors did not find significant differences in the rectus femoris between the exercise conditions.

Forces in suspension training

Apart from muscular activation, force production is also useful in assessing the load involved in strength exercises. Several studies have examined the forces exerted in different lower limb exercises. Comfort et al. [37] reported that single-leg squat achieved greater peak vertical ground reaction forces (VGRF) and higher ankle-joint moment, but a lower hip-moment, compared to the joint kinetics and kinematics analyses of forward and reverse lunges. Other studies have assessed the load on the suspension strap and VGRF in upper body exercises. Melrose and Dawes [38] measured the force exerted on the suspension strap while performing an isometric suspended inverted row in college students. These authors found that the percentage of body mass resistance on the suspension strap increases from 37.4% to 79.4% when the trunk-leg inclination is closer to the floor (from 30˚ to 75˚). Likewise, Gulmez [39] recruited male sport sciences students to examine the force on the suspension strap and VGRF while performing isometric suspended push-ups under two conditions (elbow flexion and elbow extension). The study found that when trunk-leg inclination is modified (from 45˚ to 0˚), the percentage of body mass resistance increases (elbow flexion: 36.8% to 75.3%; elbow extension: 11.9% to 50.4%), while VGRF decreases (elbow flexion: 80.7% to 32.2%; elbow extension: 97.5% to 46.6%). However, the effect of load on the suspension strap while performing lower body exercises such as squats, standard lunges, Bulgarian squats, or hamstring curls has apparently not been assessed yet. Conversely, the effects of other sources of instability on force production have been examined for lower body exercises. Previous studies have shown that an unstable environment leads to decreased force output [40,41]. Saeterbakken & Fimland [42] examined squat exercise on four different unstable surfaces and the BOSU® condition, obtaining the lowest force output value compared to a stable squat condition. Likewise, another investigation reported that BOSU® and T-Bow® deadlift conditions significantly decreased force production in deadlift on the floor [43]. Although the literature review suggests that unstable surfaces reduce force production, the dual condition (two destabilizing materials or WBV with an unstable surface) might increase muscle activation [29,44]. However, Byrne et al. [45] reported no significant difference when studying the dual condition on the suspended plank.

To the best of our knowledge, there is insufficient evidence of muscle activity and force production when a suspended lower body exercise is performed. Therefore, our primary purpose is to study the effect of suspension devices on muscle activity during a Bulgarian squat. Second, we aim to determine the effect of adding an unstable surface and WBV on muscle activity in the suspended lunge. Regarding force production, the objective was to quantify the effect of adding an unstable surface and WBV on the forces exerted on the suspension strap by the rear leg. We also compared the VGRF produced by the front leg between the Bulgarian squat and suspended lunge. Therefore, it was hypothesized that 1) a suspended lunge results in greater muscle activation than a Bulgarian squat, 2) muscle activation under Bulgarian squat and suspended lunge conditions (suspended, suspended-BOSU, suspended-vibration 30 Hz, and suspended-vibration 40 Hz) significantly differs in all analyzed muscles (rectus femoris, biceps femoris, gluteus medius, vastus lateralis, vastus medialis, and rectus femoris of the rear leg), 3) the force exerted on the suspension strap is significantly lower in suspended lunge-BOSU than under the other suspended lunge conditions, and 4) the suspended lunge condition elicits a higher VGRF load on the front leg than the Bulgarian squat.

Materials and methods

Design

A repeated measures design was used to compare electromyographic activity and force output (force exerted on the suspension strap and VGRF) during the Bulgarian squat and under four suspended lunge conditions. Twenty participants were recruited to perform the Bulgarian squat and suspended lunges. Bulgarian squats were performed with the front foot on the floor and the rear foot leaning on a bench. Suspended lunge conditions were a) suspended lunge (front foot on the floor and the rear foot leaning within the suspension device cradle), b) suspended lunge-BOSU (same as the previous exercise with front foot on BOSU®), c) suspended lunge-Vibro30 (front foot on the WBV platform at 30 Hz and 4 mm of amplitude), and d) suspended lunge-Vibro40 (same as the previous exercise with 40 Hz and 4 mm of amplitude). All suspended lunge conditions were executed using a TRX Suspension Trainer™ device. An S-Type Load Cell was used to measure the force exerted on the suspension strap by the suspended lower limb. The load cell was displayed on the suspension device. A force plate was utilized to register VGRF from the front leg in both the Bulgarian squat and suspended lunge. Surface electromyography (sEMG) was used to measure muscle activity in the dominant leg (front leg). The following muscles were analyzed: 1) rectus femoris, 2) biceps femoris, 3) gluteus medius, 4) vastus medialis, and 5) vastus lateralis. Additionally, activity in the rectus femoris of the rear leg was registered across the five exercises.

Participants

Twenty healthy and physically active male university students (mean age = 24.40 ± 3.63 years, range: 20–31 years, height = 1.79 ± 0.06 m, body mass = 78.06 ± 1.70 kg, body mass index = 24.35 ± 1.58 $kg \cdot m^{-2}$) were voluntarily recruited for this study. Subjects had been physically active with at least three sessions per week with a minimum duration of 30 min. Additionally, eight of the included subjects played soccer, six played basketball, three played handball, and three played tennis. Subjects were excluded if they presented any injuries and/or pain related to cardiovascular, musculoskeletal, or neurological disorders. All subjects were asked to come to the experimental session after refraining from high intensity physical activity for 24 h before the testing, and they consumed no food, drinks, or stimulants (i.e., caffeine) 3–4 h before testing. During the familiarization session, all subjects signed the written informed consent after receiving a clear explanation of the experimental procedures, exercise protocol, benefits, and possible risks associated with their participation. The Ethics and Research Committee Board at Blanquerna Faculty of Psychology and Educational and Sports Sciences of Ramon Llull University of Barcelona approved this study with reference number 1819005D. All protocols conducted in this research complied with the requirements specified in the Declaration of Helsinki (revised in Fortaleza, Brazil, 2013). In accordance with the PLOS consent guidelines, participants gave their written informed consent for their images to be reproduced in this manuscript.

Procedures

The study was conducted in two sessions: familiarization and experimental. They were performed at the same time in the morning, separated by a week. During the familiarization session, researchers recorded the age, weight, and height of each subject, and measured leg length, which was defined as the distance from the anterior superior iliac spine to the medial malleolus of the tibia [18]. Leg dominance was determined by asking subjects which leg they would use to kick a ball [46]. The dominant leg was used as the front leg in the Bulgarian squat and under suspended lunge conditions. To verify adherence to pre-test instructions, all subjects completed a standardized questionnaire. Subjects were familiarized with the exercise procedures by performing two sets of five repetitions under each exercise condition (Bulgarian squat, suspended lunge, suspended lunge-BOSU, suspended lunge-Vibro30, and suspended lunge-Vibro40), to achieve proper technique before data collection. A 1-min resting period between repetitions and a 2-min resting period between exercises were allowed to avoid fatigue.

During the experimental session, subjects were outfitted with surface electrodes and completed a MVIC test. Before the MVIC test, subjects performed a standardized warm-up, which consisted of 5 minutes of cycling with 100 W of cadence maintaining 60 revolutions per minute. After the MVIC test protocol, each subject performed a set of five consecutive repetitions of the Bulgarian squat and the suspended lunge exercises. The objective was to perform the different tasks at a controlled pace, maintaining posture as consistently as possible. The suspended lunge was performed under 4 conditions: 1) suspended lunge, 2) suspended lunge-BOSU, 3) suspended lunge-Vibro30 (WBV at 30 Hz and 4 mm of amplitude), and 4) suspended lunge-Vibro40 (WBV at 40 Hz and 4 mm of amplitude). In the suspended lunge-Vibro30 and -Vibro40, the WBV plate was set at 30 and 40 Hz, respectively. These frequencies show the highest demands for the knee thigh muscles in similar tasks [23,27,28]. The strength and conditioning methods used in the study procedures, including suspension, unstable surfaces, and WBV, are frequently used in several sports where the inclusion of additional weight is less common (i.e., soccer, field hockey, tennis, paddle tennis, and badminton).

The Bulgarian squat and suspended lunge exercise orders were randomized between subjects and 90 seconds of rest between exercises was allowed to prevent fatigue. Pace was standardized using a metronome (*Pro Metronome* application, version 3.13.2; EUM Lab-Xannin Technology Gmbh., Hangzhou, CHN) set at 70 beats per minute (bpm), and the tether of a positional encoder (WSB 16k-200; ASM Inc., Moosinning, DEU) was attached to the hip and used to measure its vertical displacement during all exercises. Trials were discarded and repeated if subjects were unable to perform the exercises with the correct technique.

Surface electromyography signal

All sEMG values were recorded using a BIOPAC MP-150 at a sampling rate of 1.0 kHz. Data were analyzed using the AcqKnowledge 4.2 software (BIOPAC System, INC., Goleta, CA). sEMG signals were bandpass filtered at 50–500 Hz while utilizing a 4th order Butterworth filter. Root mean square sEMG signals were recorded throughout each exercise. The mean root mean square data were then normalized to the maximal voluntary isometric contraction and reported as % MVIC.

Bipolar sEMG electrodes (Biopac EL504 disposable Ag-AgCl) with an inter-electrode distance of 2 cm were used. Surface electrodes were placed on the dominant leg (front leg) on the rectus femoris, biceps femoris, gluteus medius, vastus medialis, and vastus lateralis. An additional electrode was placed on the rectus femoris of the rear leg. Before affixing the electrodes, the subject's skin sites were prepared for application through shaving, exfoliation, and alcohol cleansing in order to reduce impedance from dead surface tissue and oils [47]. After that, the electrodes were placed following the SENIAM Project recommendations [47]. Electrodes for the rectus femoris (front and rear leg) were placed at 50% on the line running from the anterior spine iliac superior to the superior part of the patella, those for the biceps femoris were placed at 50% on the line between the ischial tuberosity and lateral epicondyle of the tibia, those for the gluteus medius were placed at 50% on the line from the crista iliac to the trochanter, those for the vastus medialis were placed at 80% on the line between the anterior spine iliac superior and joint space in front of the anterior border of the medial ligament, and those for the vastus lateralis were placed at 2/3 on the line from the anterior superior spine iliac to the lateral side of the patella. A ground surface electrode was placed directly over the right anterior superior iliac spine.

Force measurements

VGRF was measured using a force plate (Kistler 9260AA, Winterthur, Switzerland) equipped with a data acquisition system (Kistler 5695b, Winterthur, Switzerland). Raw data were acquired (sampling rate 1,000 Hz) using the MARS software (Kistler, Winterthur, Switzerland). Calibration of the system was performed according to the MARS software recommendations. While the Bulgarian squat and suspended lunge were performed, subjects centered their forward foot at a fixed position on the force plate.

To record the load on the suspension device, an S-Type Load Cell (model CZL301C; Phidgets Inc., Alberta, CAN) was displayed between the anchor point (2.95 m from the ground) and suspension device straps. Data were collected (sampling rate 200 Hz) using BIOPAC MP-150 (BIOPAC System, INC., Goleta, CA) and its original software (AcqKnowledge 4.2; BIOPAC System, INC., Goleta, CA). The system was calibrated according to the manufacturer's recommendations in the manual.

Maximum voluntary isometric contraction (MVIC)

Prior to the exercise trials described below, subjects performed three 5-s MVICs for each muscle, and the trial with the higher sEMG signal was selected in accordance with Jakobsen et al.

[48]. Subjects were instructed to increase muscle contraction force gradually towards maximum for a period longer than 2 s, sustain the MVIC for 3 s, and release the force again slowly. Three minutes of rest was allowed between each MVIC, and standardized verbal encouragement was provided to motivate all subjects to achieve maximal muscle activation. Positions during the MVICs were based on the Konrad [49] protocol for the dominant leg (front leg) muscles: rectus femoris, vastus medialis, vastus lateralis, biceps femoris, gluteus medius; and for rectus femoris of the rear leg. To obtain the MVIC of the rectus femoris, vastus medialis, and vastus lateralis, subjects performed an isometric 90˚ single leg knee extension in a seated position against matched resistance (i.e., resistance forceful enough to elicit an isometric contraction from the subject). The resistance was matched using an ankle bracelet attached to a cable that was anchored to a stretcher, thereby guaranteeing a fixed position. To obtain the MVIC of the biceps femoris, subjects performed an isometric 20–30˚ single-knee flexion in a prone-lying position against a matched resistance. Lastly, the MVIC for the gluteus medius was performed with subjects in a fixed side-lying position. An isometric hip abduction was then performed against a matched resistance. The exercise trials were performed once all MVICs were collected.

Exercise trials

To normalize the height and stepped distance under all the Bulgarian squat and suspended lunge conditions, the height of both the Bulgarian squat bench and suspension device straps was normalized to 60% of the subject's leg length; this length added the height of the force plate, BOSU®, and WBV platform (i.e., total height strap = 60% of subject's leg length + BOSU®'s height). The distance that the subjects stepped in all the Bulgarian squat and suspended lunge conditions was normalized to 80% of their leg length, measured as the distance from the anterior superior iliac spine to the medial malleolus of the tibia, in accordance with Boudreau et al. [18]. Regarding the exercise load, all subjects used their bodyweight as a load in the Bulgarian squat and under the suspended lunge conditions. The proper techniques for the exercises were as follows:

- Bulgarian squat: Subjects were instructed to stand upright with one foot in front and the other behind the body. Subjects held their arms crossed on their chest, and their upper body was maintained upright with a lower back natural sway throughout the exercise. Subjects lowered the body (eccentric phase) until the forward knee flexed to 90˚, and subsequently returned the body to the starting position with a full knee extension of the forward leg (concentric phase), maintaining an erect trunk position, as required for subjects. The forward foot was placed at a fixed position with the heel contact on a force plate. The rear foot (instep) was leaned on a horizontal press bench. To adjust the height of the rear leg, EVA foam play mat pieces were used and fixed with a cinch strap (Fig 1). The contact point between the horizontal press bench and foot was controlled so that it was identical in all repetitions.
- Suspended lunge: Prior to performing this exercise, a TRX Suspension Trainer (Fitness Anywhere, San Francisco, CA) was secured in the anchor point. Subjects were instructed to assume a lunge position with the rear foot placed within the suspension device cradle with a slight plantar flexion (Fig 1). The forward foot was placed on a force plate. Then, subjects performed the lunge as previously described.
- Suspended lunge-BOSU: A BOSU® ball (BOSU®, Ashland, OH) was used to perform this exercise. Subjects assumed the above-stated position but with the forward foot placed upon the BOSU®, dome side up (Fig 1).

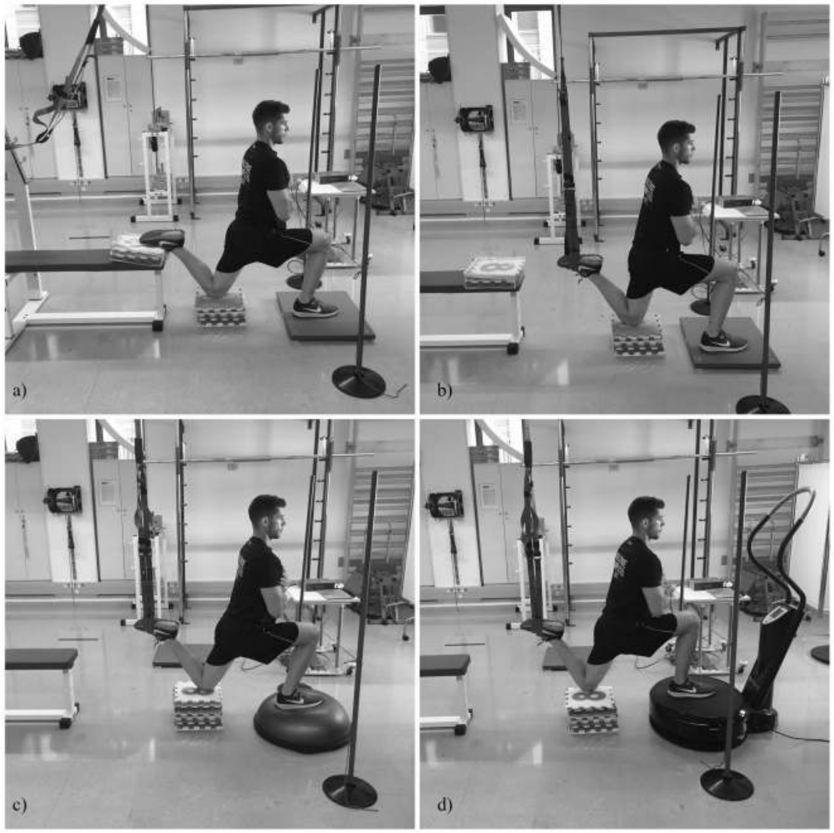

Fig 1. Bulgarian squat (a), suspended lunge (b), suspended lunge-BOSU (c), and suspended lunge-Vibro30 and Vibro-40 (d).

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- Suspended lunge-Vibro30: A WBV platform (Compex® Winplate; DJO UK Ltd., Guildford, GBR) was used to perform this exercise. Subjects were instructed to place the forward foot and maintain the heel in contact upon the Compex Winplate. The WBV platform setting was 30 Hz of frequency and 4 mm of amplitude (high) (Fig 1). Subjects then performed the lunge as previously described.
- Suspended lunge-Vibro40: Subjects performed the lunge with a WBV platform set at 40 Hz of frequency and 4 mm of amplitude (high). They placed the rear foot in the suspension straps using the same techniques as previously described (Fig 1).

Data analysis

All sEMG signal analyses were performed using the AcqKnowledge 4.2 (BIOPAC System, INC., Goleta, CA). The sEMG signals related to isometric exercises were analyzed by using the three middle seconds of the 5-s isometric contraction. The sEMG signals of the Bulgarian squat and suspended lunge conditions were analyzed by taking the average of the three middle repetitions. The first and fifth repetitions were excluded from data analysis. The sEMG amplitude in the domain was quantified using the root mean square. The mean root mean square values were selected for every trail and normalized to the maximum EMG (%MVIC). The global mean of all muscles (i.e., rectus femoris, biceps femoris, gluteus medius, vastus medialis, vastus lateralis, and rectus femoris of the rear leg) was also calculated (arithmetic mean) and analyzed. To facilitate comparison of muscle activation between conditions, activation was categorized into four levels: >60%, very high; 41–60%, high; 21–40%, moderate; and <21%, low [50].

Recorded load data from the force plate and load cell were analyzed using the entire lunge phase (eccentric-concentric repetition). Maximum force values reached in the entire phase were used during the Bulgarian squat and suspended lunge conditions. The first and fifth repetitions were excluded from data analysis.

To normalize the load, an equation was calculated for each subject based on load and body weight (load_norm = load / body weight x 100) in accordance with Gulmez [39]. The normalized values were expressed as a percentage of the total load.

Statistical analysis

Statistical analysis was accomplished using SPSS (Version 20 for Mac; SPSS Inc., Chicago, IL, USA). The sEMG signal of each muscle analyzed through all the Bulgarian and suspended lunge conditions, forces exerted on the suspension strap, VGRF, and MVICs assessment were measured. The intra-rater reliability of all the dependent variables was assessed using an intra-class correlation coefficient (ICC), and their 95% confidence intervals based on mean-rating (k = 3), absolute-agreement, two-way mixed effects model. The ICC was interpreted using the recommendations of Koo & Li [51] such as poor (<0.5), moderate (0.5–0.75), good (0.75–0.90), or excellent (>0.90) reliability. The number of subjects chosen was based on effect size 0.30 SD with an α level of 0.05 and power at 0.95 using G Power Software (University of Dusseldorf, Germany). The Shapiro-Wilk test was used to confirm that data were normally distributed to approve the use of parametric techniques. The results are reported as mean ± standard deviation. One-way repeated-measures analysis of variance (ANOVA) was employed to examine the effect of exercise condition on mean muscle activation and the forces exerted on the suspension straps. A paired t-test was conducted to compare VGRF produced by the front leg on the force plate in Bulgarian squat and suspended lunge. The Greenhouse-Greisser correction was used when the assumption of sphericity (Mauchly's test) was violated. Post hoc analysis with Bonferroni correction was used in case of significant main effects. Effect sizes are reported as partial eta-squared (η_p^2), with cut-off values of 0.01–0.05, 0.06–0.13, and >0.14 for small, medium, and large effects, respectively. For pairwise comparison, the Cohen's *d* effect size was calculated [52], and the magnitude of the effect size was interpreted as <0.2 = trivial; 0.2–0.6 = small; 0.6–1.2 = moderate; 1.2–2.0 = large; >2.0 = very large [53]. Significance was accepted when *p* value was <0.05.

Results

The ICC demonstrated good to excellent reliability under all exercise conditions for the rectus femoris, biceps femoris, gluteus medius, vastus medialis, vastus lateralis, and rectus femoris of the rear leg (Table 1). The MVIC assessment demonstrated an excellent reliability for the rectus femoris (0.955; 95% CI: 0.90–0.98), rectus femoris of the rear leg (0.973; 95% CI: 0.94–0.98), vastus medialis (0.945; 95% CI: 0.88–0.97), vastus lateralis (0.956; 95% CI: 0.90–0.98), biceps femoris (0.956; 95% CI: 0.90–0.98), and gluteus medius (0.987; 95% CI: 0.97–0.99). The ICC for the forces exerted on the suspension straps for the suspended lunge (0.982; 95% CI: 0.95–0.99), suspended lunge-BOSU (0.956; 95% CI: 0.90–0.98), suspended lunge-Vibro30 (0.978; 95% CI: 0.95–0.99), and suspended lunge-Vibro40 (0.973; 95% CI: 0.94–0.98) demonstrated an excellent reliability. The ICC showed an excellent reliability for VGRF under the Bulgarian squat (0.996; 95% CI: 0.99–0.99) and suspended lunge (0. 995; 95% CI: 0.98–0.99).

Table 1. Reliability values for each muscle analyzed under the Bulgarian squat and suspended lunge conditions. Intra-rater reliability is expressed as ICC (95% CI).

	Exercise condition	ICC (95% CI)
Rectus Femoris	Bulgarian squat	0.943 (0.88–0.97)
	Suspended lunge	0.882 (0.75–0.95)
	Suspended lunge-BOSU	0.888 (0.76–0.95)
	Suspended lunge-Vibro30	0.899 (0.78–0.95)
	Suspended lunge-Vibro40	0.945 (0.88–0.97)
Biceps Femoris	Bulgarian squat	0.919 (0.83–0.96)
	Suspended lunge	0.871 (0.73–0.94)
	Suspended lunge-BOSU	0.878 (0.74–0.94)
	Suspended lunge-Vibro30	0.795 (0.57–0.91)
	Suspended lunge-Vibro40	0.990 (0.97–0.99)
Gluteus Medius	Bulgarian squat	0.895 (0.78–0.95)
	Suspended lunge	0.894 (0.77–0.95)
	Suspended lunge-BOSU	0.946 (0.88–0.97)
	Suspended lunge-Vibro30	0.941 (0.87–0.97)
	Suspended lunge-Vibro40	0.925 (0.84–0.96)
Vastus Medialis	Bulgarian squat	0.947 (0.88–0.97)
	Suspended lunge	0.914 (0.82–0.96)
	Suspended lunge-BOSU	0.935 (0.86–0.97)
	Suspended lunge-Vibro30	0.904 (0.79–0.95)
	Suspended lunge-Vibro40	0.918 (0.82–0.96)
Vastus Lateralis	Bulgarian squat	0.880 (0.74–0.94)
	Suspended lunge	0.916 (0.82–0.96)
	Suspended lunge-BOSU	0.926 (0.84–0.96)
	Suspended lunge-Vibro30	0.758 (0.49–0.89)
	Suspended lunge-Vibro40	0.922 (0.83–0.96)
Rectus Femoris_RL	Bulgarian squat	0.887 (0.76–0.95)
	Suspended lunge	0.855 (0.69–0.93)
	Suspended lunge-BOSU	0.856 (0.70–0.93)
	Suspended lunge-Vibro30	0.911 (0.78–0.96)
	Suspended lunge-Vibro40	0.959 (0.91–0.98)

RL = Rear leg; CI = Confidence interval

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The main effects of exercise condition were identified for mean muscle activation of the rectus femoris [$F_{(2.57,48.79)} = 8.557$ $p = 0.000$, $\eta_p^2 = 0.31$], biceps femoris [$F_{(4,76)} = 3.495$ $p = 0.011$, $\eta_p^2 = 0.15$], gluteus medius [$F_{(4,76)} = 17.467$ $p = 0.000$, $\eta_p^2 = 0.47$], vastus medialis [$F_{(4,76)} = 5.578$ $p = 0.001$, $\eta_p^2 = 0.23$], vastus lateralis [$F_{(4,76)} = 6.074$ $p = 0.003$, $\eta_p^2 = 0.24$], rectus femoris of the rear leg [$F_{(4,76)} = 5.501$ $p = 0.001$, $\eta_p^2 = 0.23$]; mean muscle activation of the front leg muscles (Global_FL) [$F_{(4,76)} = 18.611$ $p = 0.000$, $\eta_p^2 = 0.49$]; and mean muscle activation of all muscles (Global) [$F_{(4,76)} = 10.524$ $p = 0.000$, $\eta_p^2 = 0.36$]. The suspended lunge provided lower but non-significant activations than the Bulgarian squat for the biceps femoris ($p = 0.392$, $d = 1.33$), gluteus medius ($p = 1.000$, $d = 0.27$), vastus medialis ($p = 1.000$, $d = 0.63$), vastus lateralis ($p = 0.647$, $d = 1.66$), Global_FL ($p = 1.000$, $d = 1.78$), and Global ($p = 0.109$, $d = 2.84$). Furthermore, the suspended lunge showed significantly lower activations than the suspended lunge-BOSU, suspended lunge-Vibro30, and suspended lunge-Vibro40 in the muscles above (Table 2). Pairwise comparisons details between exercise conditions and all muscle

Table 2. Normalized electromyographic activation for each lower body muscle under different lunge conditions as a percentage of maximum voluntary isometric contraction (%MVIC). Values are expressed as mean ± standard error of the mean (SEM).

	Bulgarian Squat (a)	Suspended Lunge (b)	Suspended Lunge-BOSU (c)	Suspended Lunge-Vibro30 (d)	Suspended Lunge-Vibro40 (e)	P-value (effect size *d*)					
						a-c	b-c	d-c	d-e		
RF_FL	32.72 ± 3.48†	33.50 ± 3.45†	45.30 ± 4.28	35.16 ± 3.96†§	44.90 ± 5.72	0.010 (0.72)	0.002 (0.68)	0.001 (0.55)	0.012 (0.44)		
						b-d	b-e				
BF	24.50 ± 2.40	21.48 ± 2.14‡§	27.21 ± 2.21	28.07 ± 2.30	26.92 ± 2.38	0.044 (0.66)	0.014 (0.54)				
						a-c	a-e	b-c	b-d	b-e	
Gmed	46.53 ± 4.18†§	45.54 ± 3.15†‡§	65.67 ± 4.85	55.73 ± 4.67	65.59 ± 4.98	0.000 (0.95)	0.001 (0.93)	0.000 (1.10)	0.022 (0.57)	0.000 (1.08)	
						a-e	b-e				
VM	64.58 ± 3.75§	62.18 ± 3.90§	67.61 ± 2.87	69.05 ± 4.45	76.23 ± 4.57	0.014 (0.62)	0.006 (0.74)				
						b-d	b-e				
VL	72.34 ± 4.81	64.92 ± 4.13‡§	76.79 ± 3.80	81.13 ± 6.31	87.63 ± 5.49	0.038 (0.68)	0.03 (1.05)				
						c-a					
RF_RL	33.51 ± 3.76	24.69 ± 3.87	23.61 ± 2.56*	26.31 ± 3.09	28.60 ± 3.00	0.019 (0.69)					
						a-c	a-e	b-c	b-d	b-e	d-e
GL_FL	47.94 ± 1.40†§	45.52 ± 1.31†‡§	56.31 ± 1.96	53.83 ± 1.89§	60.26 ± 2.32	0.005 (1.10)	0.000 (1.44)	0.000 (1.44)	0.001 (1.14)	0.000 (1.75)	0.043 (0.68)
						a-e	b-c	b-d	b-e		
GL	46.75 ± 1.48§	42.76 ± 1.33†‡§	50.64 ± 2.20	50.53 ± 1.46	54.37 ± 2.03	0.010 (0.96)	0.012 (0.97)	0.001 (1.26)	0.000 (1.51)		

RF_FL = Rectus femoris front leg; BF = Biceps femoris; Gmed = Gluteus medius; VM = Vastus medialis; VL = Vastus lateralis; RF_RL = Rectus femoris rear leg; GL_FL = Global mean of the five front leg muscles; GL = Global mean of the six muscles

* = Significantly lower than Bulgarian squat;

† = Significantly lower than Suspension lunge-BOSU

‡ = Significantly lower than Suspension lunge-Vibro30;

§ = Significantly lower than Suspension lunge-Vibro40

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activation data are presented in Table 2. The percentage of electromyographic activations for all suspended lunges related to the Bulgarian squat conditions is shown in Fig 2.

Fig 3 shows the forces exerted on the suspension straps by the rear leg for each suspended lunge condition and VGRF produced by the front leg in the Bulgarian and suspended lunge exercises. An exercise condition main effect was found for the forces exerted by the rear leg on the suspension strap [$F_{(3,57)} = 5.106$ $p = 0.003$, $\eta_p^2 = 0.21$]. The force exerted on the suspension strap was significantly lower during the suspended lunge-BOSU than during the suspended lunge-Vibro30 ($p = 0.009$, $d = 0.56$) (Fig 3a). Furthermore, the front leg force production was significantly higher during the suspended lunge than during the Bulgarian squat ($t_{(19)} = -3.106$, $p = 0.006$, $d = 0.48$) (Fig 3b).

Discussion

The main findings of the study were that the effect of the suspension strap does not provoke an increase of the muscle activity in the front leg in the suspended lunge and the lack of a

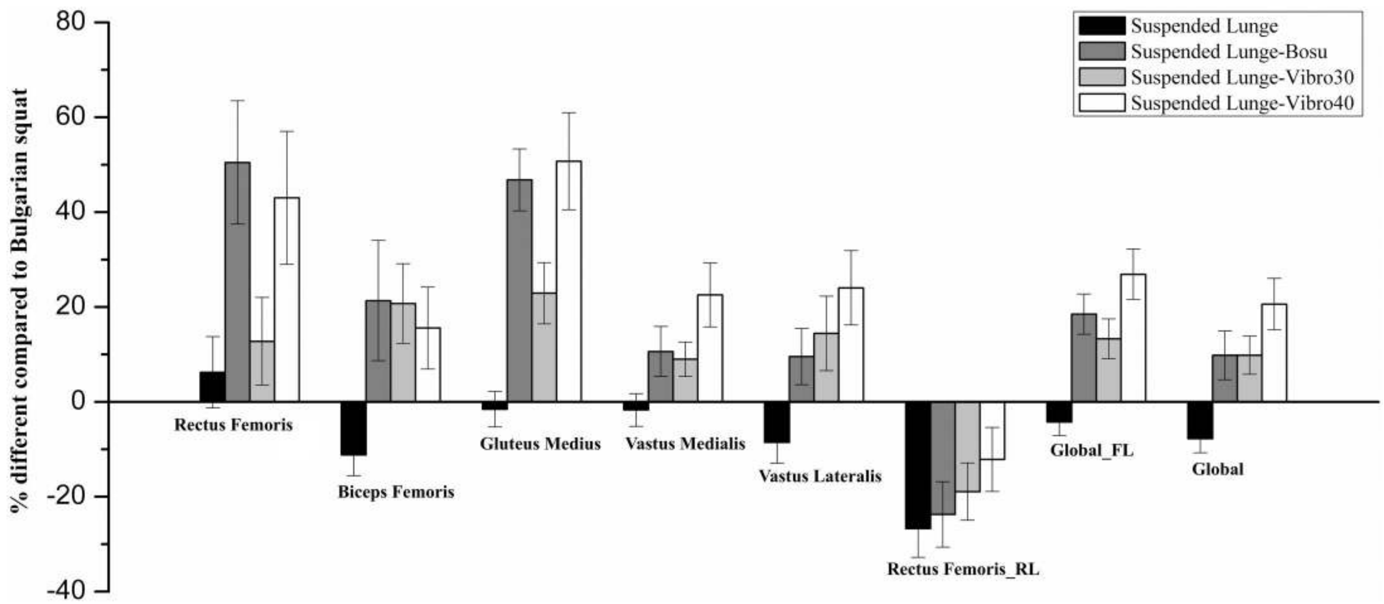

Fig 2. Electromyographic activations for all conditions relative to the Bulgarian squat. Each bar represents the mean, and the error bar represents the standard error of the mean (SEM). FL = Front leg; RL = Rear leg.

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consistent support point was equally demanding for the analyzed muscles. Thus, similar muscle activation of suspended lunges as that of Bulgarian squats ranged from moderate (rectus femoris and biceps femoris) to high (gluteus medius) and very high (vastus medialis and lateralis), which reinforces this argument. All the suspended lunge conditions, except the suspended lunge-BOSU, showed a higher but non-significant activation of the rectus femoris compared to the Bulgarian squat. The suspended lunge-BOSU achieved a significantly higher activation of the rectus femoris compared to the moderate activity in the Bulgarian squat (p = 0.010, d = 0.72). The same recruitment patterns for the rectus femoris were found by Krause et al. [36] who reported non-significant differences in the activation of the rectus

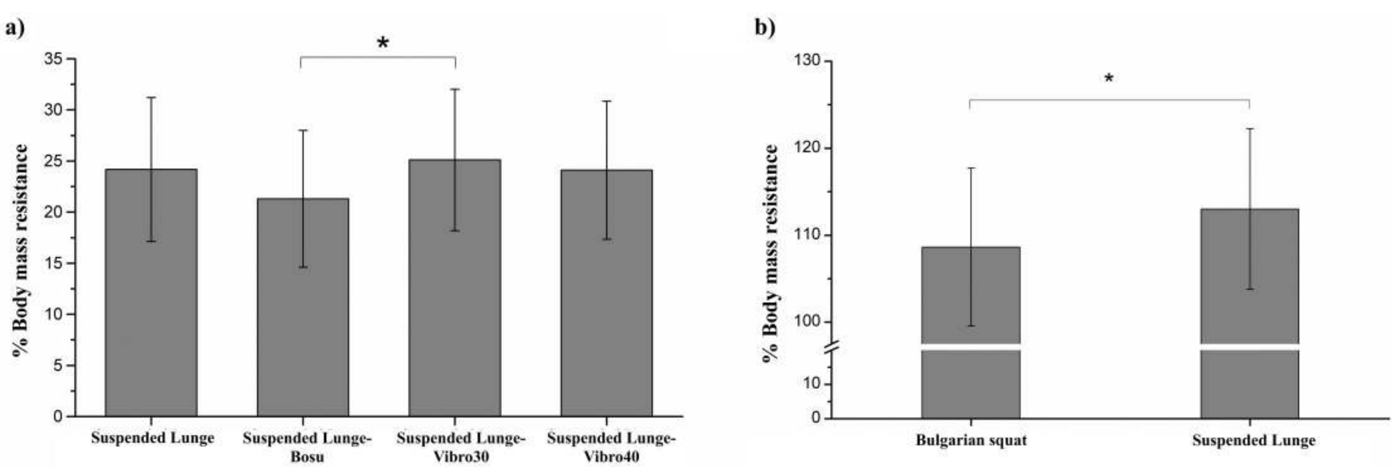

Fig 3. Force values during the Bulgarian squat and suspended lunge conditions: a) Comparison between forces exerted by rear leg on the suspension strap and exercise condition, b) Front leg force production comparison between Bulgarian squat and suspended lunge. Each bar represents the mean, and the error bar represents the standard deviation (SD). * Significant difference (p<0.05).

https://doi.org/10.1371/journal.pone.0221710.g003 femoris in the standard lunge compared to that in the suspended lunge (22.1 ± 22.2%MVIC vs. 24.5 ± 22.0%MVIC, $p = 0.434$). Furthermore, Andersen et al. [17] did not find significant differences in the activation of the rectus femoris while performing a 6-RM Bulgarian squat under stable and unstable conditions (stable vs. unstable: 70.7 ± 18.3%MVIC vs. 68.9 ± 16.1% MVIC). On the other hand, it seems that performing a unilateral lower limb exercise with a suspension strap on the rear leg or an unstable surface on the front leg causes higher demands for the rectus femoris. This is because the primary role of the rectus femoris in the Bulgarian squat and suspended lunge could be the control of the hip flexion and knee extension movements, instead of stabilizing the abduction, adduction, and rotational movements of the hip and pelvis [36].

Regarding the remaining front leg muscles, the Bulgarian squat showed a slightly greater but non-significant muscle recruitment compared to the suspended lunge. For the biceps femoris, the activation was moderate; in the gluteus medius, the activation was high; and in the vastus medialis and vastus lateralis, the activation was very high among the conditions. As reported in previous studies, the vastus medialis and lateralis achieved a higher, but non-significant, very-high activation during a 6-RM Bulgarian squat compared to the unstable Bulgarian squat [17]. The study conducted by Mausehund et al. [54], in healthy and moderate strength-trained students, indicated that the activation of the vastus lateralis was higher, but not significant, for the 6-RM Bulgarian squat than for the 6-RM split squat and single-leg squat, even though both exercises registered a very high level of activity. These authors also showed non-significant differences for the gluteus medius while performing the Bulgarian squat and split squat, even though these two exercises provided a moderate activity of the gluteus medius. The Bulgarian squat was more gluteus medius demanding. Likewise, DeForest et al. [19] reported that during the concentric phase of a loaded Bulgarian squat, the activation of the biceps femoris (around 390 mV) and vastus medialis (around 640 mV) and lateralis (around 670 mV) was higher than that of a bilateral and split squat. In contrast, Krause et al. [36] reported that the suspended lunge increases significantly the muscle recruitment for the hamstring and gluteus medius (13.1 ± 20.1% MVIC; 24.1 ± 15.1%MVIC, respectively) compared to a standard lunge (hamstring: 8.7 ± 13.2%MVIC, $p = 0.01$; gluteus medius: 15.3 ± 11.4% MVIC, $p = 0.01$). Exercise technique may explain the differences in muscle activity because previous studies showed that when performing a standard lunge, in healthy subjects, the muscle activity of the biceps femoris was low [55,56], that of the gluteus medius ranged from low to moderate [18,55], and that of the vastus medialis and lateralis ranged from high to very high [55,56]. Differently, the Bulgarian squat is more demanding than the standard lunge. Previous studies showed that the activity of the biceps femoris and vastus (medialis and lateralis) was very high [17,54] and that of the gluteus medius was moderate [54]. Thus, performing a Bulgarian squat with the front leg on the floor demands a higher hip and thigh muscle recruitment than a standard lunge, and therefore, the difference in the muscle activation between the traditional and suspended exercises is higher in case of a standard lunge than the Bulgarian squat. Furthermore, leaning the rear leg on the suspension strap appears to produce a decrease in the recruitment of these muscles.

Another finding was the need for a dual condition to elicit higher muscle activation, in the front leg (suspended lunge-BOSU, suspended lunge-Vibro30, and suspended lunge-Vibro40) but not in the rear leg. The two conditions eliciting higher activation of the rectus femoris and gluteus medius in the front leg were suspended lunge-BOSU (45.30 ± 4.28%MVIC; 65.67 ± 4.85%MVIC, respectively) and suspended lunge-Vibro40 (44.90 ± 5.72%MVIC; 65.59 ± 4.98%MVIC, respectively). For these muscles, the stimulus provoked by the BOSU® conditions could be equivalent, in terms of muscle activation, with those offered by the WBV platform at 40 Hz-high, but not at 30 Hz-high. Pollock et al. [57] found in healthy participants standing on a WBV platform at 30 Hz of frequency and 5.5 mm of amplitude that the rectus femoris recruitment was significantly higher than when WBV was set at 5 Hz of frequency and the same amplitude. These authors indicated that muscle recruitment for the rectus femoris depends on the frequency and amplitude of vibration. This finding suggests that dual conditions with WBV and compliant environments compromised the postural stability, leading to increased muscle tuning mechanisms and muscle contraction [29,58]. Furthermore, gluteus medius was solicited to stabilize the body during the dynamic flexo-extension of the front leg, which characterizes lunges under a suspended-BOSU condition, but also to absorb the vibration offered by the vibration plate. Moreover, the activation found in the antagonist (biceps femoris) and vastus (medialis and lateralis) was similar and not significantly different in the three dual conditions, being higher in the Vibro40 condition. The equivalences of the effects between BOSU® and vibratory conditions might be caused by the contribution of multiple neural pathways with distinct functional roles to rapid motor control response to a perturbation [59]. Thus, the neuromuscular response for maintaining the posture on a BOSU® may be more intelligent than merely a voluntary or a reflex mechanism [60] integrating the modulation of the long-latency stretch reflexes. Sensitivity increases of these reflexes were reported when subjects interacted with compliant environments, and this suggests its significant role in maintaining the limb stability in such conditions [59]. According to this, the reflex motor response during the BOSU® condition and the vibratory tonic reflex on the WBV platforms might induce similar activation in the involved muscles. This finding, also reflected in the global activation (the mean of all analyzed muscles), might be explained by the particular requirements of absorbing the vibration or maintaining the stability on a BOSU®. Hence, performing dynamic tasks on a BOSU®, subjects experiment a muscular trembling (micro amplitude changes), provoked by body mass variations projected on the forward leg, leaned on a compliant surface like this during the whole range of movement. These micro amplitude changes are described as one of the muscle tuning mechanisms for vibration training [20]. Additionally, WBV has been proven as beneficial improving the coordination of the synergistic muscles and increasing the inhibition of the antagonists, together with increases in hormonal responses of testosterone and growth hormone [61], besides the beneficial effects on bone mineral density [62], muscle blood volume [63] or balance control, and muscle endurance [64].

In terms of global activation, the use of WBV platforms, together with devices such as BOSU®, enhances muscle activity in the suspended lunge in physically active young adults. Thus, the simple use of a suspension device is not demanding enough for the studied exercise and needs to be complemented with other loading sources. So, inclusion of additional methods increasing the instability (BOSU®, Swiss ball, Pielaster®, rubber mats), vibration with demanding amplitudes and frequencies, and extra weights (weighted vests and belts, barbells, kettlebells) is necessary to increase the muscle activation of the involved muscles and the force produced.

The third finding of this study was that the force produced on the suspension straps was significantly lower for suspended lunge-BOSU than for suspended lunge-Vibro30 (21.3% ± 6.7 vs. 25.1% ± 6.93, $p = 0.009$), and this force was lower, but not significant, than the suspended lunge and suspended lunge-Vibro40. Thus, the present study shows that the percentage of body mass resistance exerted by the rear leg on the suspension strap could not be influenced by the front leg lean (on the floor or the WBV platform). However, to perform the suspended lunge under dual condition with a device such as BOSU® provokes an increase in the amount of instability, and thus, the load exerted by the rear leg on the suspension strap decreases in accordance with Behm et al. [40] and their hierarchy of force outputs proposal, which states that the degree of stability or instability affects limb force production directly. This finding is according to Saeterbakken & Fimland [42] who reported that in healthy subjects, the isometric force output achieved while performing a squat on BOSU® (603 ± 208 N) was significantly lower than the force produced under a stable squat on the floor (749 ± 222 N) or less unstable surfaces as squats on the power board (694 ± 220 N).

The VGRF exerted by the front leg on the force plate was significantly higher during a suspended lunge than during the Bulgarian squat (113.01% ± 9.24 vs. 108.65% ± 9.05, $p = 0.006$). This finding suggests that leaning the rear leg on a suspension strap provokes a transfer of a certain amount of body mass resistance towards the front leg, maintaining the trunk position, which exerts a force on the ground to attempt to keep the posture. Also, the increase of VGRF in the suspended lunge may be due to the low activation of rectus femoris of the rear leg. Consequently, maintaining the rear leg on a suspension device could inhibit the role of rectus femoris as a hip flexor and contribute to the increase of the VGRF in the front leg.

There were some limitations associated with this study. Results of the present study may be influenced by subjects' experiences with similar exercises to those performed in the present investigation. Each individual has a different level of motor control for the same task, and this might be taken into account when assessing muscle electrical signals. Therefore, participants' characteristics might constitute a limitation to infer the results of the present study. This study did not use functional tests to determine participants' laterality, together with their neuromuscular and performance level. Moreover, the lack of quantification about the amount of instability produced by the device should be considered. Another limitation may be that a goniometer did not control the knee flexion angle. However, the displacement during each repetition of the Bulgarian squat and suspended lunge conditions was measured with a positional encoder. Further research should examine the muscle activity and force output when performing suspended lunges to compare the muscle recruitment between lower body suspension and traditional resistance training exercises. Furthermore, the assessment of the perturbation related to the use of unstable surfaces with an accelerometer would be interesting.

In conclusion, the results of this study demonstrated that suspended lunges provide no additional benefit than Bulgarian squats to enhance lower body muscle activity. Performing a lunge at dual conditions increases exercise muscle activity compared with a Bulgarian squat and suspended lunge. However, dual conditions decrease the load on the suspension strap when the front leg leans on an unstable surface (i.e., BOSU®), and the VGRF exerted by the front leg in the suspended lunge (compared to its traditional counterparts) is enhanced to overcome the instability generated by the suspension device.

Supporting information

S1 File. STROBE checklist of the study.
(DOCX)

S2 File. Clinical studies checklist.
(DOCX)

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Study 4

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Title: Correlational data concerning body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions in high-standard track and field athletes

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Data Article

Correlational data concerning body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions in high-standard track and field athletes

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ABSTRACT

This article reports data concerning the body centre of mass acceleration, muscle activity, and forces exerted during a suspended lunge under different stability conditions. Ten high-standard track and field athletes were recruited to perform one set of 5 repetitions of the following exercises: suspended lunge, suspended lunge-Foam (front leg on a foam balance-pad and the rear leg on the suspension cradles), a suspended lunge-BOSU up (dome side up), and a suspended lunge-BOSU down (dome side down). For each exercise trial, the acceleration of the body centre of mass (tri-axial accelerometer BIOPAC), the muscle activity of the front leg (surface electromyography BIOPAC) and the force exerted on the suspension strap (load cell Phidgets) were measured. The data revealed that the intra-reliability of the data range from good (ICC: 0.821) to excellent (ICC: 0.970) in all dependent variables and exercise conditions. Besides, the Pearson correlation between muscle activity and the body centre of mass acceleration showed a significant positive correlation for all the exercises and analysed muscles (range from $r = 0.393$ to $r = 0.826$; $p < 0.05$) with

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moderate to very large effect, except for the rectus and biceps femoris. Moreover, the force exerted on the suspension strap significantly correlated with the body centre of mass acceleration in all the exercises (range from $r = -0.595$ to $r = -0.797$, $p < 0.05$) with a very large effect, except for the suspension lunge that registered a large effect.

Specifications Table

Subject	Sport sciences
Specific subject area	Strength and conditioning
Type of data	Table Image Figure
How data were acquired	Six channels of sEMG (Biopac), tri-axial accelerometer (Biopac) and s-type load cell (Phidgets) acquired using Biopac System MP-150 at a sampling rate of 1.0 kHz.
Data format	Raw Filtered Analysed
Parameters for data collection	Participants (high-standard athletes) were excluded if they presented any injuries or pain related to cardiovascular, musculoskeletal, or neurological disorders. All subjects were instructed to refrain from high-intensity physical activity or neuromuscular stimulation for the 24h before the experimental sessions, and they consumed no food, drinks, or stimulants (i.e., caffeine) 4h before testing.
Description of data collection	The experiment was conducted in 2 sessions: familiarisation and experimental. They were performed at the same time in the morning, separated by a week. All suspended lunge conditions were executed using a TRX Suspension Trainer™ device. An S-Type Load Cell was used to measure the force exerted on the suspension strap by the suspended lower limb in random order (90-s rest). The load cell was displayed on the suspension device. Surface electromyography (sEMG) was used to measure muscle activity in the dominant leg (6 most recruited muscles), which was established as the front leg. The tri-axial accelerometer was placed in the waist to measure the body centre of mass acceleration.
Data source location	Barcelona (Catalonia) Spain
Data accessibility	Repository name: Mendeley Data Direct URL to data: https://doi.org/10.17632/8wj8gpgwmr.3

Value of the Data

- The presented data might improve the understanding of the acceleration contribution to muscle involvement, and the forces exerted in a lower limb suspended exercise commonly used in specific strength and conditioning programs.
- Strength and conditioning coaches and practitioners could use the data to select different variations of a suspended unilateral lower limb exercise.
- The different correlations associating muscle activity and forces exerted in different exercise conditions could be used to analyse the ability of a subject to stabilizing a unilateral lower-limb action.
- Additionally, data might help sports facilities to select the best equipment for creating unstable strength and conditioning environments.

1. Data

The present article contains data concerning body centre of mass acceleration, muscle activity and forces exerted during the execution of a suspended lunge exercise under different conditions

Table 1
Participants' characteristics including athletic background.

Participant	Age	Height (m)	Weight (kg)	Training age	Athletic level	Athletic discipline	Hours of training		
							Weekly	Training specifications	
Sub1	22	1.69	57	16	Int.	Endurance (800 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub2	19	1.79	71	13	Int.	Endurance (800 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub3	21	1.76	63	15	Int.	Sprint (400 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub4	21	1.70	64	15	Int.	Sprint (400 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub5	18	1.70	58	12	Int.	Sprint (400 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub6	20	1.71	63	15	Int.	Sprint (400 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub7	18	1.68	60.5	13	Int.	Sprint (100 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub8	18	1.65	49	12	Int.	Sprint (400 m)	10	S: 3 Sp: 3	E: 3 T: 1
Sub9	21	1.67	51	15	Int.	Endurance (800 m)	10	S: 3	E: 3 T: 1
Sub10	20	1.67	55	15	Int.	Sprint (400 m)	10	S: 3	E: 3 T: 1
Mean	19.80	1.70	59.15	14.10					
SD	1.48	0.04	6.57	1.45					

Sub: Subject; Int: International; S: Strength; Sp: Speed; E: Endurance; T: Technique.

of instability in high-standard athletes (athletes enrolled in a sports talent program, national finalists and training 10 hours weekly, see Table 1). Different variables were measured by using surface electromyography (sEMG), a Tri-axial accelerometer and a load cell simultaneously recorded by the BIOPAC MP-150 at a sampling rate of 1.0 kHz (BIOPAC System, INC., Goleta, CA). Reliability of the data is reported in Table 2. The correlation between the sEMG signals for all analysed muscles and acceleration are reported in Table 3. Correlations among the forces exerted on the suspended strap and acceleration are reported in Table 4. The smallest worthwhile change (SWC) and the coefficient of variation of the dependent variables for each condition are reported in Table 5. Regression point plots expressing the relationship between the acceleration and muscle activity of the rectus femoris, vastus medialis, vastus lateralis, gluteus maximus, gluteus medius and biceps femoris are shown in Fig. 1, Fig. 2, Fig. 3, Fig. 4, Fig. 5 and Fig. 6, respectively. Fig. 7 shows the regression point plots between the acceleration and force exerted on the suspension strap.

2. Experimental design, materials, and methods

A repeated measures design was used to establish the relationship between the body centre of mass acceleration, muscle activity and the force exerted on the suspension strap during different suspended lunge conditions. Ten high-standard track and field athletes (mean ± standard deviation

Table 2
Reliability values for each muscle analysed, acceleration and force under suspended lunge conditions.

	Exercise Condition	ICCs (level of reliability)	95% CI	SEM
Rectus femoris	SL	0.876 (Good)	0.65–0.97	0.06
	SL_Foam	0.873 (Good)	0.62–0.97	0.06
	SL_BU	0.844 (Good)	0.67–0.97	0.07
	SL_BD	0.963 (Excellent)	0.89–0.99	0.04
Vastus medialis	SL	0.879 (Good)	0.64–0.97	0.04
	SL_Foam	0.923 (Excellent)	0.78–0.98	0.04
	SL_BU	0.920 (Excellent)	0.77–0.98	0.05
	SL_BD	0.844 (Good)	0.56–0.96	0.06
Vastus lateralis	SL	0.821 (Good)	0.46–0.95	0.05
	SL_Foam	0.888 (Good)	0.68–0.97	0.04
	SL_BU	0.903 (Excellent)	0.73–0.97	0.05
	SL_BD	0.857 (Good)	0.57–0.96	0.05
Gluteus maximus	SL	0.940 (Excellent)	0.83–0.98	0.04
	SL_Foam	0.945 (Excellent)	0.83–0.99	0.03
	SL_BU	0.960 (Excellent)	0.89–0.99	0.05
	SL_BD	0.939 (Excellent)	0.83–0.98	0.06
Gluteus medius	SL	0.846 (Good)	0.53–0.96	0.07
	SL_Foam	0.912 (Excellent)	0.75–0.98	0.06
	SL_BU	0.916 (Excellent)	0.76–0.98	0.09
	SL_BD	0.896 (Good)	0.69–0.97	0.09
Biceps femoris	SL	0.844 (Good)	0.54–0.96	0.04
	SL_Foam	0.964 (Excellent)	0.90–0.99	0.01
	SL_BU	0.936 (Excellent)	0.82–0.98	0.03
	SL_BD	0.905 (Excellent)	0.72–0.97	0.04
Acceleration	SL	0.990 (Excellent)	0.96–1	0.01
	SL_Foam	0.994 (Excellent)	0.98–1	0.01
	SL_BU	0.996 (Excellent)	0.99–1	0.01
	SL_BD	0.996 (Excellent)	0.99–1	0.01
Force	SL	0.964 (Excellent)	0.90–0.99	1.06
	SL_Foam	0.969 (Excellent)	0.91–0.99	1.02
	SL_BU	0.961 (Excellent)	0.89–0.99	1.16
	SL_BD	0.970 (Excellent)	0.92–0.99	1.08

CI: Confidence interval; ICCs: Interclass correlation coefficients; SEM: Standard error of measurement; SL: Suspended lunge; SL_Foam: Suspended lunge-Foam; SL_BU: Suspended lunge-BOSU up; SL_BD: Suspended lunge-BOSU down.

Table 3
Pearson's correlation between muscle activity values for each muscle analysed and acceleration under suspended lunge conditions.

	Suspended lunge	Suspended lunge-Foam	Suspended lunge-BOSU up	Suspended lunge-BOSU down
Rectus femoris	−0.050	0.192	0.283	−0.087
p-value	0.794	0.310	0.130	0.649
LC	Trivial	Small	Small	Trivial
Vastus medialis	0.699*	0.632*	0.650*	0.588*
p-value	0.000	0.000	0.000	0.001
LC	Large	Large	Large	Large
Vastus lateralis	0.393*	0.689*	0.629*	0.506*
p-value	0.031	0.000	0.000	0.004
LC	Moderate	Large	Large	Large
Gluteus maximus	0.477*	0.553*	0.611*	0.558*
p-value	0.008	0.002	0.000	0.001
LC	Moderate	Large	Large	Large
Gluteus medius	0.526*	0.749*	0.826*	0.646*
p-value	0.003	0.000	0.000	0.000
LC	Large	Very large	Very large	Large
Biceps femoris	0.468*	−0.216	0.250	−0.158
p-value	0.009	0.251	0.183	0.403
LC	Moderate	Small	Small	Small

LC: Level of correlation; *Statistical significance at $p < 0.05$.

Table 4
Pearson's correlation (r) between forces exerted on the suspension strap and acceleration under suspended lunge conditions.

	Suspended lunge	Suspended lunge-Foam	Suspended lunge-BOSU up	Suspended lunge-BOSU down
r	−0.595*	−0.797*	−0.776*	−0.741*
p-value	0.001	0.000	0.000	0.000
LC	Large	Very large	Very large	Very large

LC: Level of correlation; *Statistical significance at $p < 0.05$.

Table 5
Smallest worthwhile change and coefficient of variation values for each muscle analysed, acceleration and force under suspended lunge conditions.

	Exercise Condition	SWC	CV
Rectus femoris	SL	0.03	0.002
	SL_Foam	0.03	0.002
	SL_BU	0.04	0.002
	SL_BD	0.04	0.002
Vastus medialis	SL	0.02	0.001
	SL_Foam	0.03	0.001
	SL_BU	0.03	0.002
	SL_BD	0.03	0.001
Vastus lateralis	SL	0.02	0.001
	SL_Foam	0.02	0.001
	SL_BU	0.03	0.002
	SL_BD	0.03	0.001
Gluteus maximus	SL	0.04	0.002
	SL_Foam	0.03	0.001
	SL_BU	0.05	0.003
	SL_BD	0.05	0.002
Gluteus medius	SL	0.03	0.002
	SL_Foam	0.04	0.002
	SL_BU	0.06	0.003
	SL_BD	0.06	0.003
Biceps femoris	SL	0.02	0.001
	SL_Foam	0.01	0.001
	SL_BU	0.02	0.001
	SL_BD	0.03	0.001
Acceleration	SL	0.02	0.001
	SL_Foam	0.02	0.001
	SL_BU	0.03	0.001
	SL_BD	0.03	0.001
Force	SL	1.11	0.056
	SL_Foam	1.15	0.058
	SL_BU	1.18	0.059
	SL_BD	1.25	0.062

SWC: Smallest worthwhile change; CV: Coefficient of variation; SL: Suspended lunge; SL_Foam: Suspended lunge-Foam; SL_BU: Suspended lunge-Bosu up; SL_BD: Suspended lunge-Bosu down.

(SD): age, 19.8 ± 1.48 years; height, 1.70 ± 0.04 m; body mass, 59.15 ± 6.67 Kg) were recruited to perform a suspended lunge in 4 conditions: a) suspended lunge (front leg on the floor and the rear leg leaning within the suspension device cradle (TRX® Suspension training system, patent No.: US 7,044,896 B2; Fitness Anywhere, San Francisco, CA), b) suspended foam (same as the previous exercise with the front leg on a balance-pad (AIREX®, Sins, CH), c) suspended BOSU up (front leg on the BOSU (BOSU®, Ashland, OH) with the dome side up), and d) suspended BOSU down (same as the previous exercise) with the dome side down. Participants assumed a lunge position with their arms crossed on their chest, and their upper body upright with a lower back natural sway. For the lower body, the subjects lowered the body (eccentric phases) until the forward knee flexed to 90° and

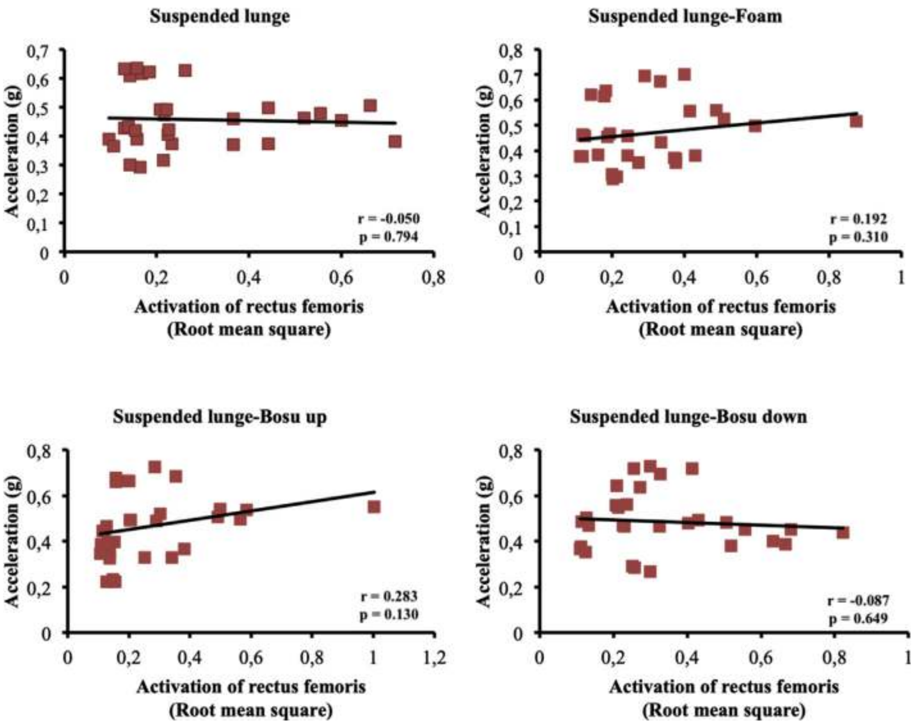

Fig. 1. Correlation between rectus femoris activation and acceleration values under suspended lunge conditions.

then, returned to the starting position with a full knee extension of the forward leg (concentric phase) [1]. The vertical displacement during all exercises was measured with a positional encoder (WSB 16K-200; ASM Inc., Moosinning, DE) and the tether of the positional encoder was attached to the hip. The forward food was placed on different surfaces (floor, balance pad, BOSU dome side up and down) with the heel contact on the floor, balance pad or BOSU. The forward leg was chosen as the dominant leg, which was determined by asking participants which leg they would use to kick a ball [2]. The rear foot was placed within the suspension device cradle with slight plantar flexion in all the exercise conditions (supplementary material). Besides, the height and stepped distance, and 90° of knee flexion were normalized. The height of the suspension straps was established as 60% of the subject's leg length, and the subjects stepped distance was normalized to 80% of their leg length [3]. The 90° of knee flexion were established by measuring with a manual goniometer the knee flexion in the lower position. Once the 90° were identified, customized stoppers (similar to hurdles) were used to fix this position. Feedback on how much they had to go down, and when to start the counter-movement was also provided to the participants (see Supplemental material). Before the exercise trials, a standardized warm-up was carried out, consisting of 5 minutes of cycling with 100 W of cadence maintaining 60 revolutions per minute. Then, each participant performed a set of 5 consecutive repetitions of each suspended lunge exercise. The objective was to perform the different tasks at a controlled pace, maintaining the posture as consistently as possible. During the exercise trials, all subjects performed one set of 5 repetitions of each condition with a standardized pace of 70 beats per minute in a randomized order. Participants were provided with a 90-s rest between exercises to avoid fatigue.

During the trials muscle activity, forces exerted on the suspension strap and body centre of mass acceleration were measured. To record muscle activity, 12 bipolar surface electromyography electrodes were placed on the front leg (dominant leg) on the rectus femoris, vastus lateralis, vastus

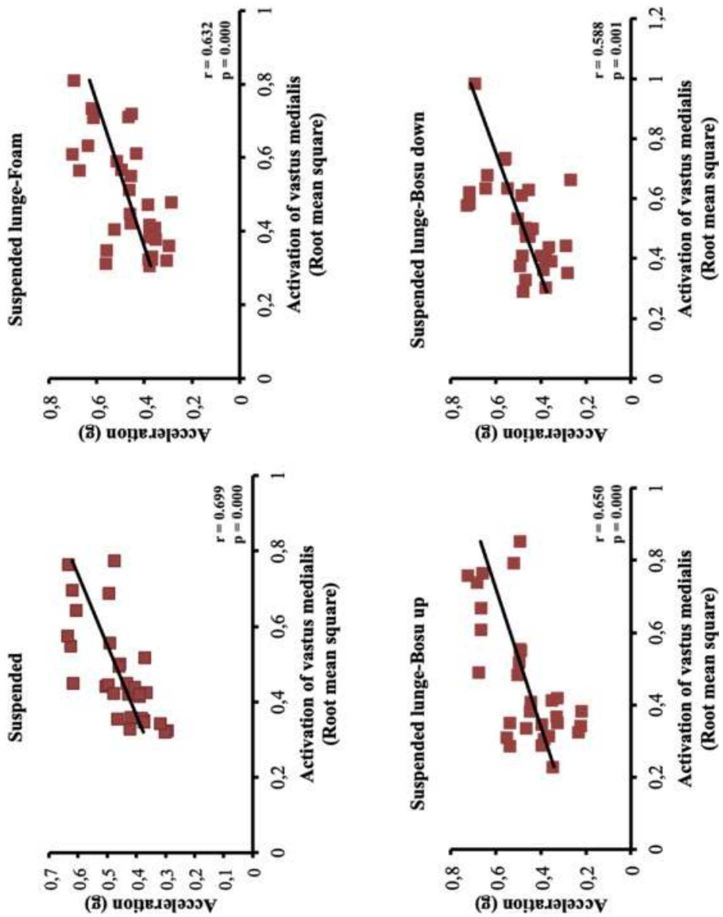

Fig. 2. Correlation between vastus medialis activation and acceleration values under suspended lunge conditions.

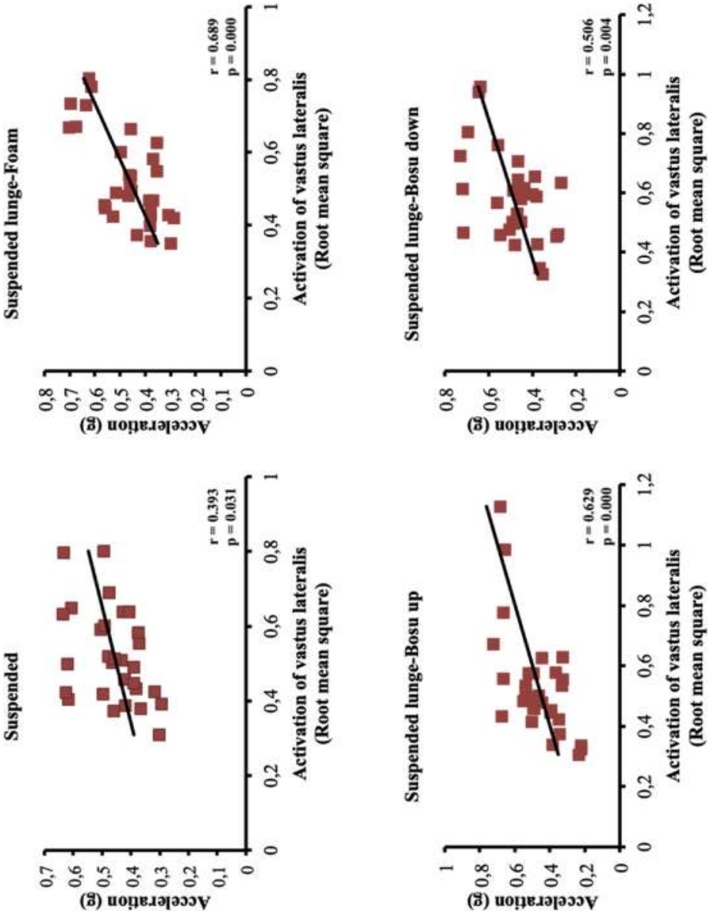

Fig. 3. Correlation between vastus lateralis activation and acceleration values under suspended lunge conditions.

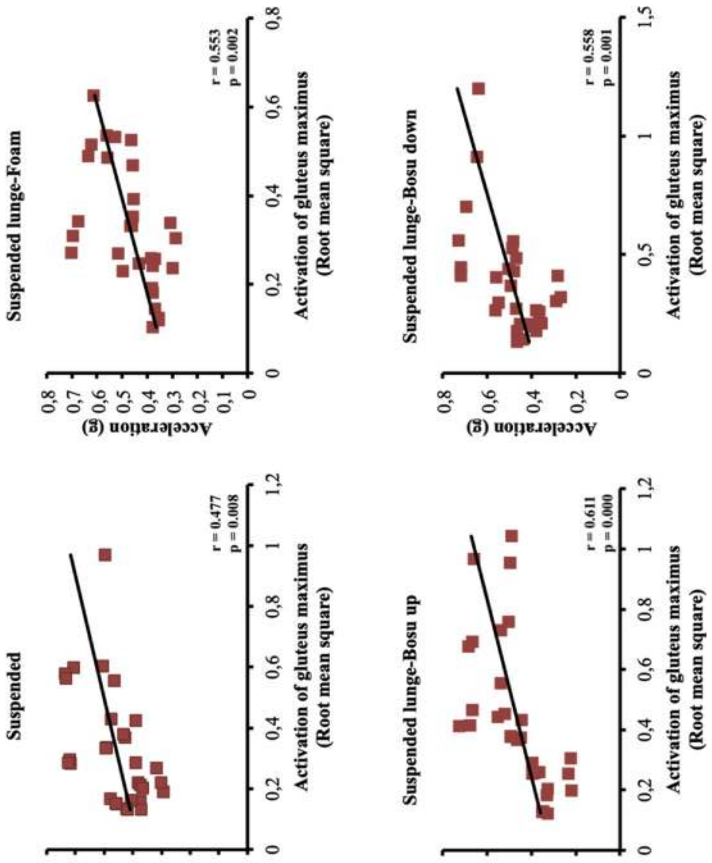

Fig. 4. Correlation between gluteus maximus activation and acceleration values under suspended lunge conditions.

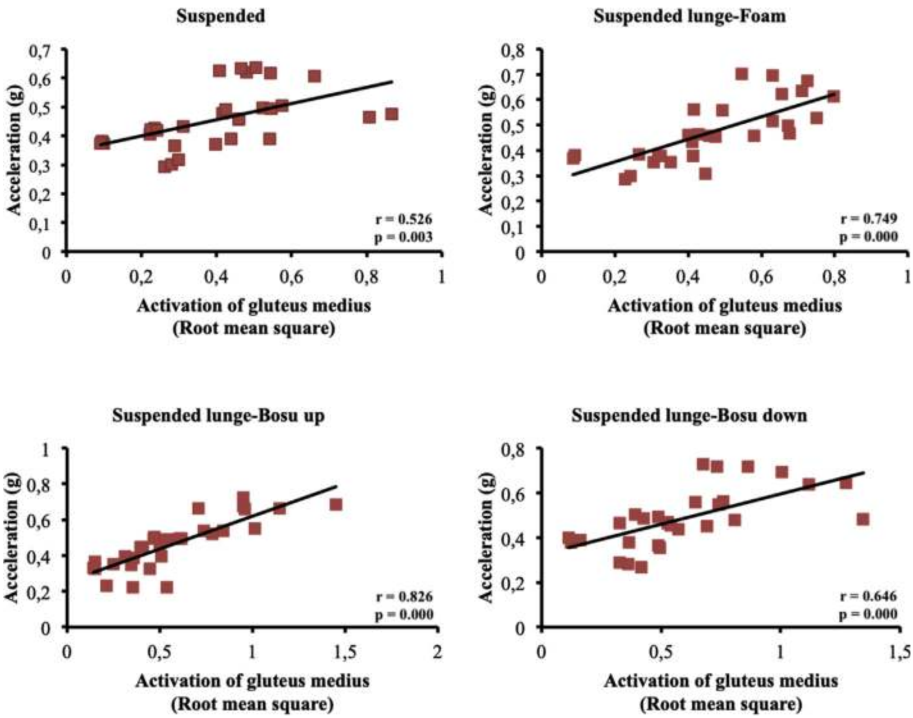

Fig. 5. Correlation between gluteus medius activation and acceleration values under suspended lunge conditions.

medialis, gluteus maximus, gluteus medius and biceps femoris following the SENIAM Project recommendations [4]. An additional electrode was placed directly over the right anterior iliac spine as a ground surface electrode. The surface electromyographic values (root mean square) were registered with a BIOPAC MP-150 at a sampling rate of 1.0 kHz. The signal was bandpass filtered at 50–500 Hz while utilizing a 4th Butterworth filter and then analysed using the AcqKnowledge 4.2 software (BIOPAC System, INC., Goleta, CA). The forces exerted on the suspension strap were recorded using an S-Type Load Cell (model CZL301C; Phidgets Inc., Alberta, CAN) with a sample rate of 200 Hz. The load cell was placed between the anchor point (2.95 m from the ground) and the suspension straps. Moreover, a tri-axial accelerometer (model TSD109F, BIOPAC System, INC., Goleta, CA) was placed in the waist to measure the body centre of mass accelerations with a sample rate of 2.0 kHz, a sensitivity of 40 mV/g, and a range of ±50g. The force and body centre of mass acceleration were recorded using a BIOPAC MP-150 and its original software.

Surface electromyography, force and body centre of mass acceleration signals for each exercise condition were analysed by taking the average of the three middle repetitions, excluding the first and fifth repetitions from data analysis. To normalize the force exerted on the suspension straps, an equation was used for each participant based on load and body mass (%_body mass resistance = load/bodyweight x 100) [5]. The number of participants recruited was established using an α level of 0.05 and setting power at 0.50 using G Power Software (University of Dusseldorf). The Shapiro-Wilk test was carried out to confirm that data were normally distributed to approve the use of parametric techniques. The intra-rater reliability of all the dependent variables was assessed using an intraclass correlation coefficient (ICC), and their 95% confidence interval based on mean-rating (K = 3), absolute-agreement, two-way mixed-effects model. Pearson's correlation (r) was employed to determine the relationship between the following dependent variables a) muscle activity and body centre of mass acceleration, and b) force exerted on the suspension straps and body centre of mass acceleration. The ICC was interpreted such as poor

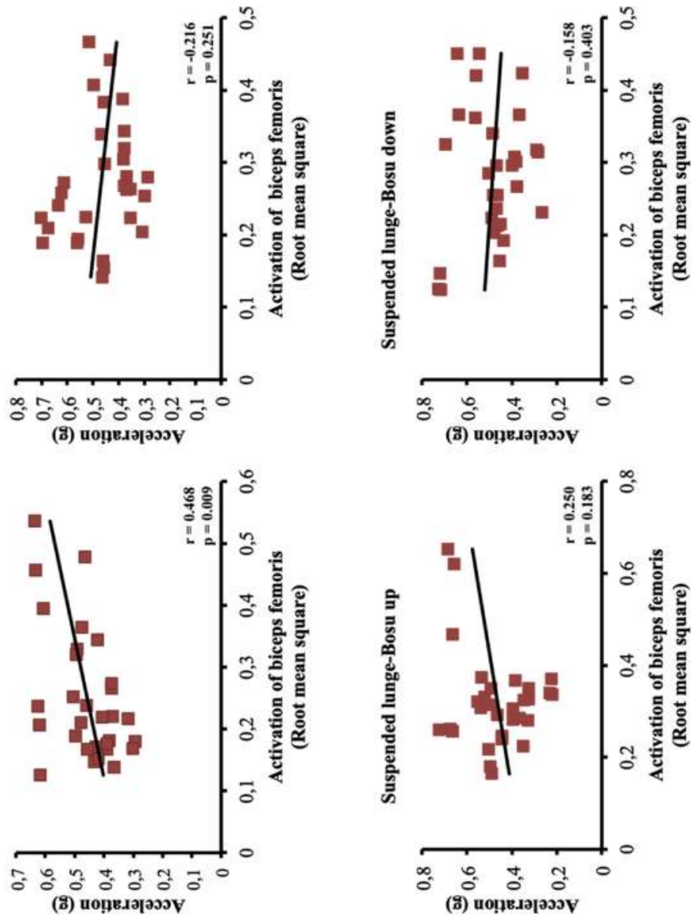

Fig. 6. Correlation between biceps femoris activation and acceleration values under suspended lunge conditions.

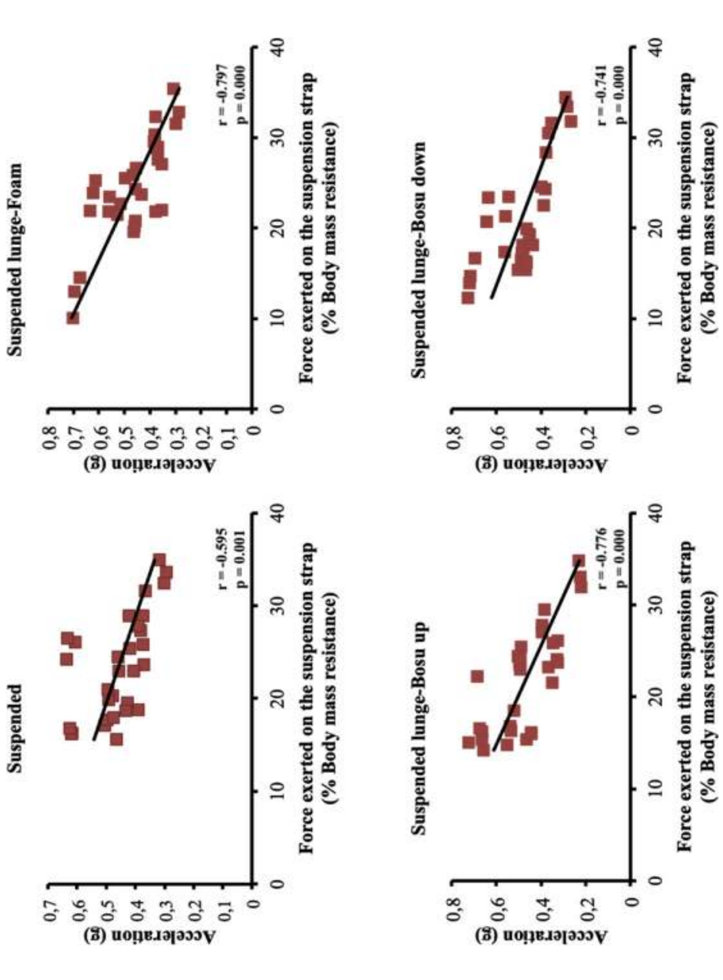

Fig. 7. Correlation between forces exerted on the suspension strap and acceleration values under suspended lunge conditions.

(<0.5), moderate (0.5–0.75), good (0.75–0.90), or excellent (>0.90) reliability [6]. The coefficient of variation was also estimated, and the small-standardized effect based on Cohen's effect size principle (SWC) was calculated as 0.2 x between-subject standard deviation (SD).

Additionally, the magnitude of the Pearson's correlation values were interpreted as <0.2 = trivial; 0.2–0.6 = small; 0.6–1.2 = moderate; 1.2–2.0 = large; >2.0 = very large [7]. Significance was accepted when p value was <0.05. The statistical analysis was accomplished using SPSS (Version 20 for Mac; SPSS Inc., Chicago, IL, USA).

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Conflict of Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.dib.2019.104912.

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Study 5

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Article

Influence of the Amount of Instability on the Leg Muscle Activity During a Loaded Free Barbell Half-Squat

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Abstract: This study aimed to understand the acute responses on the muscular activity of primary movers during the execution of a half-squat under different unstable devices. Fourteen male and female high-standard track and field athletes were voluntarily recruited. A repeated measures design was used to establish the differences between muscle activity of the primary movers, the body centre of mass acceleration and the OMNI-Perceived Exertion Scale for Resistance Exercise (OMNI-Res) in a half-squat under four different stability conditions (floor, foam, BOSU-up and BOSU-down). A significant correlation was found between the highest performance limb muscle activity and body centre of mass acceleration for half-squat floor (r = 0.446, p = 0.003), foam (r = 0.322, p = 0.038), BOSU-up (r = 0.500, p = 0.001), and BOSU-down (r = 0.495, p = 0.001) exercises. For the exercise condition, the half-squat BOSU-up and BOSU-down significantly increased the muscle activity compared to half-squat floor (vastus medialis: p = 0.020, d = 0.56; vastus lateralis: p = 0.006, d = 0.75; biceps femoris: p = 0.000–0.006, d = 1.23–1.00) and half-squat foam (vastus medialis: p = 0.005–0.006, d = 0.60–1.00; vastus lateralis: p = 0.014, d = 0.67; biceps femoris: p = 0.002, d = 1.00) activities. This study contributes to improving the understanding of instability training, providing data about the acute muscular responses that an athlete experiences under varied stability conditions. The perturbation offered by the two BOSU conditions was revealed as the most demanding for the sample of athletes, followed by foam and floor executions.

Keywords: unstable; perturbation; electromyography; squatting

1. Introduction

Athletic performance is associated with specific neuromuscular adaptations improving the motor unit recruitment and the coordination of all the muscles involved in a given action. For such purposes, athletes perform different motor tasks searching for varied and effective training stimuli [1]. In this vein, a progression in load has been the ideal strategy for increasing muscular demands, but, in recent years, unstable environments have also been used with similar purposes [2–4]. Thus, different unstable devices have been used to enhance the effects of several exercises on muscle activation, force production, motor control, and consequently, athletic performance [1,5,6]. The design of these devices is intended to alter the relationship between the base of support, the body's spatial position, and the athlete's ability to maintain balance during the execution of a task. Therefore, the amount of instability depends on factors such as the nature of the task, characteristics of the subject (weight, height, muscle abilities and motor control) and the different features of the device (shape, material, friction, size and display) [1].

Performing conditioning exercises in an unstable environment, such as on a BOSU, Swiss balls, rubber discs, and freeman plates, or hanging loose objects on barbells, creates perturbations in whole-body stability. Thus, perturbation training represents a new challenge for somatosensory, vestibular, and visual systems [7]. Moreover, perturbed tasks increase the co-contractile activity, enhancing the role of antagonists to mitigate the uncertainty produced by the source of instability [8]. But how much instability does each device generate in the environment? Is this acute response the same for different athletic profiles? Which muscles are more demanded, and which are worked less when squatting? To address these questions, several studies have been conducted to assess the impact of instability on muscle activation during the execution of a squat [2,3,6,9,10]. As examined by Behm and Anderson [1], several authors have reported decrements of muscle activity of the primary squat movers under unstable conditions [3,6]. Specifically, McBride et al. [3] showed higher muscle activity of the vastus lateralis and biceps femoris under stable conditions (floor vs inflated disc) in three different loaded-squats in recreationally resistance-trained men, and Andersen et al. [6] found non-significant differences between stable and unstable squat conditions (foam) in the rectus femoris and both vastus muscles in males with a background in strength training. McBride et al. [2] found significantly higher muscle activity in the biceps femoris when squatting on two inflatable balance discs in recreationally resistance-trained males. However, no significant differences were found in both vastus muscles by Saeterbakken and Fimland (2013), when comparing the muscle activity of all the primary squat movers under different unstable conditions (Power Board, BOSU and Balance Cone). The authors established the instability properties of the devices used based on the number of unstable axes and the magnitude of contact with the floor. Unstable environments have been revealed in several studies to be a useful tool to elicit higher muscle activation in the core muscles when squatting [5,11,12].

As mentioned earlier, several groups of researchers have studied the acute responses of different unstable environments on muscle activation and force/power production in the past [3,10,13,14], but to the best of our knowledge, none of them have quantified the amount of instability created. According to the studies' designs, it can be inferred that some devices can create higher instability than others, but no data are available describing how unstable every condition is. Other studies have reported data using accelerometers in strength and conditioning settings. Thus, Vazquez–Guerrero et al. [15] compared the force output under different stability conditions of a flywheel squat providing mean values, and a correlation between thigh muscle activation and mean acceleration of body centre of mass has been found by Aguilera–Castells et al. [16] in a suspended lunge under unstable dual conditions. In other contexts, Thiel et al. [17] used different accelerometers to assess the quality of the movements in professional dancers, with lower acceleration peaks associated with higher performance in a series of demi-pliés. Moreover, Johnston et al. [18] used an inertial sensor to detect minor changes when performing the Y Balance test in healthy adults, calculating the postural adjustment from the XY axis and filtered data from a gyroscope. In the tested task, participants were required to explore their limits in stabilizing the whole body. Additionally, Barbado et al. [19] used the accelerometer of a smartphone to describe the intensity of core training through the quantification of the centre of pressure mean linear acceleration in different unstable environments. Thus, associating muscular activation with the amount of instability at each repetition of a set of exercises under different unstable conditions could explain the real effect of the different sources of instability on athletes. Nevertheless, the methods used in the cited studies present insufficient or questionable validity in some movements because mean acceleration was considered, instead of the sum of the integrated (x- and y-axis) acceleration peaks.

Other investigations have quantified or altered the balance with sufficient validity and reliability using different methods such as force platforms [20], stabilometers [21] and pressure mats [22] in the context of ankle and knee rehabilitation processes, fall prevention and postural balance in different populations. However, all these methods have several limitations when assessing the amount of instability in dynamic strength and conditioning tasks. One of the main limitations when using force platforms, stabilometers or pressure mats together with a BOSU, foams, or other devices providing ground instability, arises from the fact that the base of support and the ground reaction forces significantly change with respect to the execution on the floor. Indeed, the devices' characteristics change these parameters and, consequently, the validity of the amount of instability measured. Thus, measuring the dynamics of the body centre of mass far from the floor could address this issue. Understanding the amount of instability constitutes an essential factor in better explaining how challenging a task constraint is for the different sport profile [1]. In this regard, while some unstable environments are challenging for less-trained individuals, others are able to stabilize their posture even in the most unstable conditions. Therefore, muscle activity should reflect these differences.

Therefore, the first objective of the present study was to analyse the amount of instability in different half-squat conditions (floor, foam, BOSU-up and BOSU-down) experienced by high-standard athletes using an accelerometer, determining a protocol for its quantification. The second objective was to compare the muscle activity of the biceps femoris, vastus medialis, and vastus lateralis, and the global activity (sum of all the analysed muscles) of the highest performance limb during the execution of the half-squat and to assess the rating of perceived exertion (OMNI-Res) under the four aforementioned conditions. Thirdly, the relationship between the body centre of mass acceleration (BCMA) and the global muscle activity (sum of all the analysed muscles) was established. We hypothesized that the BOSU-down condition elicits higher BCMA than BOSU-up, foam and floor conditions, respectively, and follows this order of potential instability. In contrast, we expected lower muscle activity as the condition became more unstable. We also hypothesized a significant relationship between BCMA and global muscle activity, considering the different tested conditions.

2. Materials and Methods

2.1. Participants

Fourteen males (n = 5, mean age = 20.00 ± 1.41 years, range = 18–21 years; height = 1.73 ± 0.05 m, body mass = 64.00 ± 4.64 kg, body mass index = 21.48 ± 1.19 kg·m^{-2}) and females (n = 9, mean age = 20.44 ± 1.67 years, range = 18–23 years; height = 1.67 ± 0.03 m, body mass = 56.72 ± 4.89 kg, body mass index = 20.29 ± 1.43 kg·m^{-2}), all high-standard track and field athletes (i.e., 11 sprinters and 3 middle-distance runners), volunteered to participate in the study and were intentionally recruited. As inclusion criteria, all the participants were enrolled in a sport talent program, and all of them national finalists, training for at least 10 h per week (i.e., speed, endurance, and technical skill training) while engaging in international competitions. Participants were regularly checked by the sport talent program medical team, and none of them were excluded from the sample because they did not present any injury or pain related to cardiovascular, musculoskeletal or neurological disorders, following the American College of Sports Medicine exercise testing procedures. Before the familiarisation session and test session of the study, participants were encouraged to avoid consuming stimulants (e.g., caffeine), drinks or food 3 to 4 h before the session and to avoid high-intensity physical activity for 24 h before the test.

Before participating, each athlete was fully informed about the experimental procedures and the risks and benefits of participating in the study, as well as receiving and signing a written consent form. The Ethics and Research Committee Board in the Blanquerna Faculty of Psychology and Educational and Sport Sciences at Ramon Llull University in Barcelona, Spain, approved the study (ref. no. 1819005D). All protocols implemented in the study complied with the requirements specified in the Declaration of Helsinki (revised in Fortaleza, Brazil, 2013).

2.2. Experimental Procedures

A repeated measures design was applied to establish the relationship between muscle activity and body centre of mass acceleration (BCMA). Electromyographic activity, BCMA and results on the OMNI-Perceived Exertion Scale for Resistance Exercise (OMNI-Res) were compared during the resistance half-squat under different conditions of stability. The study was conducted in two sessions—familiarisation and test sessions—performed a week apart: both from 11 a.m. to 2 p.m. Firstly, the familiarisation session was conducted to acquaint the participants with the exercise technique and determine the highest performance limb, and the load lifted in a single maximum repetition (1 RM) in the half-squat. Secondly, the test session was used to assess muscle activity, BCMA and OMNI-Res results when performing the half-squat on four surface conditions: the floor, foam (Balance Pad; Airex, Sins, China), BOSU-up (BOSU, Ashland, OH, USA) with the dome side up, and BOSU-down with the dome side down.

The familiarisation session was held to collect the participants' age, weight, height, leg length, the width of the distance between the anterior superior iliac spine, and other descriptive variables (e.g., hours of training). Next, a general 10 min warm-up was performed (i.e., squatting exercise with bodyweight, dynamic stretches and joint mobility of the lower limb involved in the half-squat exercise) and a specific 10 min warm-up consisting of one set of 20 repetitions of the half-squat with the additional load of the squat bar (10 kg) and two sets of 10 repetitions of half-squats with a loaded bar (60–70% 1 RM). Before the values of 1 RM were recorded, participants performed a unilateral half-squat against an invincible resistance to determine their maximum voluntary isometric contraction in the concentric phase, measured with two force sensors anchored to the ground. To individualise the exercise but to allow all participants to apply force with knee flexion of 90°, two non-elastic straps were anchored between the force sensor and the bar following Saeterbakken and Fimland's protocol [10], used to establish the leg to be analysed under the different conditions of the exercise. The selected criterion was the highest-performing limb [23], defined as the side with the highest value in a specific task—in the study, the half-squat exercise. During the 1 RM test, the speed of the bar was controlled with a linear positional transducer (Chronojump-Boscosystem; Barcelona, Spain). During the warm-up, the velocity for the unloaded half-squat was determined to be >1.28 $m{\cdot}s^{-1}$ (<40% 1 RM) and for the loaded half-squat from 1.00 $m{\cdot}s^{-1}$ to 0.84 $m{\cdot}s^{-1}$ (60–70% 1 RM). To determine the value of 1 RM, participants performed a set of 10 repetitions of the half-squat on the floor condition, and according to the average speed and predictive equation $Load$ (% 1 RM) = −5.961 $MPV2$ − 50.71 MPV + 117.0, in which MPV refers to "mean propulsive velocity" [24], the value of each participant's load was individualised according to the relative value of 80% of 1 RM. That load value (i.e., 80% 1 RM) obtained for the half-squat on the floor condition was used for all exercise conditions.

The test session began with placing electromyographic electrodes (BIOPAC EL504 disposable Ag–AgCl) with an inter-electrode distance of 2 cm on the vastus medialis, the vastus lateralis and the biceps femoris of the highest-performing leg according to the recommendations of the SENIAM project. Before placement, the leg was shaved, exfoliated, and cleaned with alcohol to reduce the impedance of dead tissue surfaces and oils. Afterwards, a tri-axial accelerometer was placed on the waist for measuring the BCMA. Then, participants performed a standardised warm-up involving dynamic stretching, joint mobility and squatting in a set of 10 repetitions at 40% 1 RM. Next, participants began performing the half-squat protocol on the four surface conditions (i.e., floor, foam, BOSU-up and BOSU-down) in a random order (Figure 1). In each condition, they completed a set of five repetitions with a relative load of 80% of 1 RM at 60 beats per minute at an eccentric-to-concentric phase ratio of 1:1. A linear positional transducer used to control the range of movement in all repetitions of the different surface conditions was attached to the participant's hip. Between performing the half-squat exercise in each condition, participants received a 2 min rest period to prevent fatigue. Trials not performed with the proper technique were discarded and repeated.

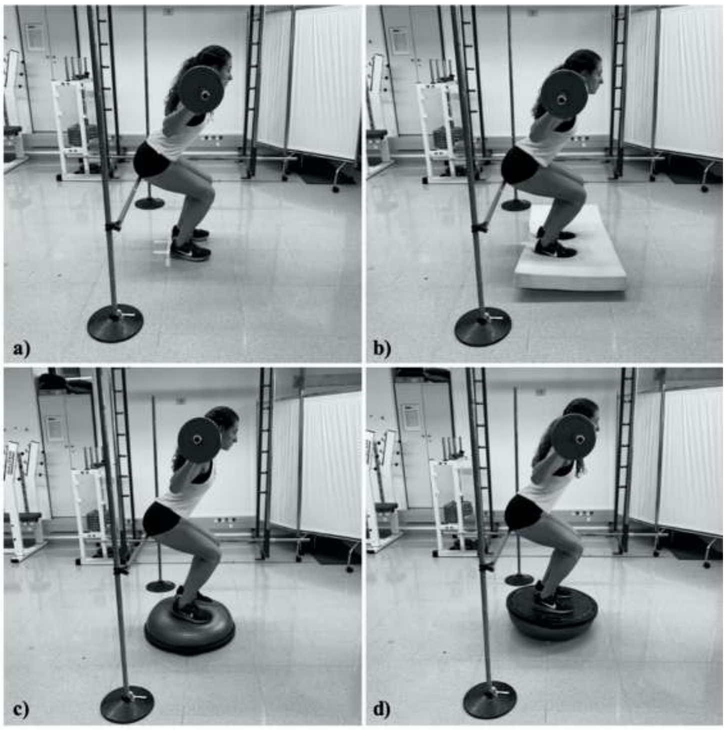

Figure 1. Exercise conditions: (**a**) half-squat floor, (**b**) half-squat foam, (**c**) half-squat BOSU-up, and (**d**) half-squat BOSU-down.

The half-squat depth was normalised to 75% of the participant's leg length, with the feet placed apart slightly wider than shoulder width and with toes pointed forward. The bar was placed across the shoulders on the trapezius slightly above the posterior aspect of the deltoids. Customised stoppers, similar to hurdles, were used to fix the lower position of the half-squat (Figure 1). Participants were instructed about the squat depth and when to commence the countermovement. Feedback regarding when to begin the half-squat and how to stand on the surface (i.e., upright, both feet planted and hands on the bar in a prone position) was provided. Participants' shoulders were placed at 90° of abduction with a slight external rotation, while the lower back maintained a neutral position. Participants lowered their body (i.e., eccentric phase) until their gluteus touched the customised stoppers and subsequently returned to the starting position with a full knee extension of the legs (i.e., concentric phase).

2.3. Surface Electromyography Signals

The data acquisition system BIOPAC MP150 was used to record all the surface electromyographical values at a sampling rate of 1.0 kHz, and these data were analysed using the AcqKnowledge 4.2 software (BIOPAC System, INC., Goleta, CA, USA). The electromyographical surface signals were bandpass filtered at 10–500 Hz utilizing a fourth order Butterworth filter. For each exercise, the root mean square surface electromyography signals were recorded.

The surface electromyography signals of all the exercise conditions were analysed by taking the average of the three middle repetitions, excluding the first and fifth repetition from the data analysis. The surface electromyography signal amplitude in the domain was quantified using the root mean square, and these values were selected for every trial. The global mean of all muscles (i.e., vastus medialis, vastus lateralis, and biceps femoris) was calculated (arithmetic mean), and the global activity (sum of the three analysed muscles) was also calculated.

2.4. Body Centre of Mass Acceleration

All BCMA values were measured by a tri-axial accelerometer TSD109 F (BIOPAC System, INC., Goleta, CA) with a sample rate of 2.0 kHz, a sensitivity of 40 mV/g, and a range of ± 50 g. Data were collected using BIOPAC MP150 and the AcqKnowledge 4.2 software. The tri-axial accelerometer was calibrated according to the manufacturer's recommendations.

Before analysing data from the BCMA, a bandpass filter fixed at 0.5 Hz (low), and 20 Hz (high) was applied, and then this signal was integrated with a root mean square. The BCMA values were analysed using the complete repetition on the anterior–posterior and proximal–lateral axes. The first and fifth repetitions were excluded from data analysis. The sum of amplitudes in the entire phase was analysed (Figure 2). This data analysis was based on the sum of all the maximum BCMA values reached in the entire phase. The global mean of the BCMA for each axis under all the exercise conditions was calculated. Next, the vector of acceleration was calculated as the quadratic combination of the global mean values of the anterior–posterior and proximal–lateral axes. After that, the global mean of this vector (arithmetic mean) was also calculated and analysed. This calculation method reflects the magnitude of the micro destabilizations necessary to maintain a balanced posture while squatting. A mean of all the acceleration data does not reflect this phenomenon, while the sum of peak values does.

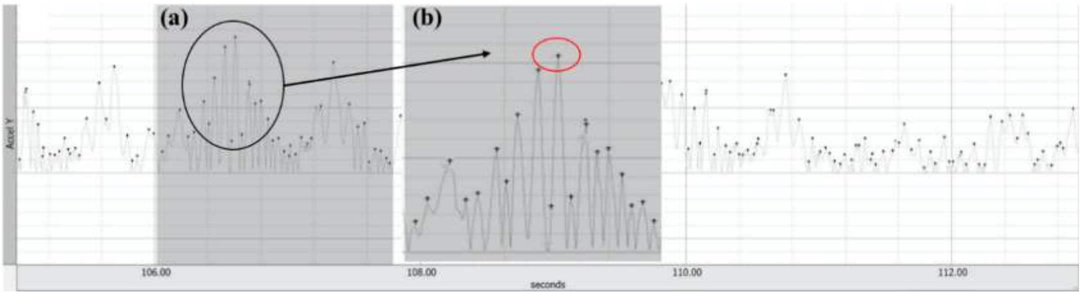

Figure 2. Body centre of mass acceleration signal (Y–axis). The signal shows all the changes in the body centre of mass acceleration (BCMA) during one repetition (entire phase) of the half-squat performed on the floor (**a**) = The shaded area shows the total number of amplitudes in the entire phase; (**b**) the magnified zone details each of the maximum BCMA values (red circle). These values were summed to determine the value of BCMA in the entire phase.

2.5. OMNI-Perceived Exertion Scale for Resistance Exercise

The OMNI-Res was used to rate the perceived exertion of the participants for each half-squat condition. Participants were asked to rate their perceived exertion for the overall body on completion of each exercise. During the familiarization session, participants were instructed to assign a rating of 1 to any perception of exertion that was less than that experienced during the unweighted repetition and a rating of 10 to any perception of exertion that was greater than that experienced during a 1 RM lift. The assessment of the OMNI-Res during the testing session was conducted following the Robertson et al. [25] instructions. Moreover, all the participants were instructed to establish a visual–cognitive link depicted visually by an athlete lifting weights at the top and bottom of the OMNI-Res scale. After collecting OMNI-Res values, the global mean of OMNI-Res (arithmetic mean) was calculated and analysed.

2.6. Statistical Analysis

The number of participants chosen was based on effect size 0.40 SD with an α level of 0.05 and power at 0.95, using G Power Software (University of Dusseldorf, Dusseldorf, Germany). The Shapiro–Wilk test was used to confirm that data were normally distributed to approve the use of the parametric techniques. The results were analysed by a statistical description of each of the dependent variables to obtain the mean values and standard error of the mean (SE) (mean ± SE). The intra-rater reliability of all quantitative dependent variables (muscle activity and BCMA) was assessed using an intraclass correlation coefficient (ICC), and their 95% confidence intervals were based on a mean rating (K = 3), absolute agreement, and a two-way mixed-effects model. Pearson's correlation (r) was used to determine the relationship between muscle activity (global activity) and BCMA of each repetition and exercise condition. Moreover, a linear mixed model analysis was used for global activity and included the exercise condition (half-squat on floor, foam, BOSU-up and -down) and BCMA as fixed effects, and participants were considered as random effects. The effect of every exercise condition on muscle activity (vastus medialis, vastus lateralis, biceps femoris, and global activity) was analysed using a linear mixed model, which was fitted to analyse whether the changes for muscle activity were influenced by exercise condition. The activation of vastus medialis, vastus lateralis, biceps femoris, and global activity were considered to be the dependent variables, the exercise condition (floor, foam, BOSU-up, and BOSU-down) was considered as a fixed effect, and participants were considered as random effects. Furthermore, another linear mixed model was used to examine whether the exercise condition modified the BCMA; the BCMA was considered as the dependent variable, the exercise condition (floor, foam, BOSU-up, and BOSU-down) was considered as a fixed effect, and participants were considered as random effects. For the previous models, the significance of the fixed effects associated with the outcome variable included in the model was assessed using the Wald test, with statistical significance set at $p < 0.05$. After the models were validated, the residuals of the final models were explored for normality, homogeneity, and independence assumptions. The normality assumption of the residuals was checked using a normal Q–Q plot of residuals. The OMNI-Res data did not meet the inferential parametric assumptions. A non-parametric Friedman test was used to examine the effect of exercise on the OMNI-Res. Post hoc Wilcoxon test analysis with Bonferroni correction was used in case of significant main effects. For pairwise comparison, the Cohen's *d* effect size was calculated [26], and the magnitude of the effect size was interpreted as <0.2 = trivial, 0.2–0.6 = small, 0.6–1.2 = moderate, 1.2–2.0 = large, and >2.0 = very large [27]. The ICC was interpreted using the recommendations of Koo and Li [28], i.e., poor (<0.5), moderate (0.5–0.75), good (0.75–0.90), and excellent (>0.90) reliability. Likewise, the magnitude of the Pearson's correlation values was interpreted as < 0.1 = trivial, 0.1–0.3 = small, 0.3–0.5 = moderate, 0.5–0.7 = large, 0.7–0.9 = very large, and 0.9–1 = nearly perfect. Statistical data were analysed using SPSS (Version 26 for Mac; SPSS Inc., Chicago, IL, USA) with a significance value of $p < 0.05$.

3. Results

The ICC demonstrated good to excellent reliability under all exercise conditions for all the analysed muscles and BCMA values (Table 1). The Pearson correlation between the highest performance limb activity and BCMA was significant for half-squat floor (r = 0.446, $p = 0.003$), foam (r = 0.322, $p = 0.038$), BOSU-up (r = 0.500, $p = 0.001$), and BOSU-down (r = 0.495, $p = 0.001$) exercises, all of them with a moderate effect (r = 0.3 to 0.5). Additionally, the linear mixed model showed a significant fixed effect for exercise condition [F (3,42) = 6.706, $p = 0.001$] and BCMA [F (1,46) = 19.209, $p = 0.000$] on global activity (Table 2). The effect of exercise condition on muscle activity showed a significant fixed effect for exercise condition on vastus medialis [F (3,42) = 6.350, $p = 0.001$], vastus lateralis [F (3,42) = 6.039, $p = 0.002$], biceps femoris [F (3,42) = 10.051, $p = 0.000$] and global activity [F (3,42) = 10.028, $p = 0.000$], and the results from linear mixed model are shown in Table 3. Post-hoc analysis showed a significantly greater vastus medialis activity for half-squat BOSU-up than half-squat floor ($p = 0.020$, d = 0.56) and foam ($p = 0.005$, d = 0.60) exercises, and vastus medialis recruitment was also significantly greater for half-squat BOSU-down than half-squat foam ($p = 0.037$, d = 0.53) lifts. A significantly greater activity for vastus lateralis was achieved under the half-squat BOSU-down condition compared to half-squat floor ($p = 0.006$, d = 0.75) and foam ($p = 0.014$, d = 0.67) repetitions. For the biceps femoris, activity was significantly greater for the half-squat BOSU-up and half-squat BOSU-down than for the half-squat floor activities ($p = 0.000$, d = 1.23; $p = 0.006$, d = 1.00, respectively). Moreover, the biceps femoris activity was significantly greater for the half-squat BOSU-up than half-squat foam exercises ($p = 0.002$, d = 1.00) (Table 4). The global activity was significantly greater for half-squat BOSU-up than floor ($p = 0.001$, d = 0.85) and foam ($p = 0.002$, d = 0.83) repetitions, also this activity significantly increased for half-squat BOSU-down in comparison with half-squat floor ($p = 0.003$, d = 0.84) and foam ($p = 0.004$, d = 0.79) movements (Figure 3a).

Table 1. Reliability values for each muscle analysed and body centre of mass acceleration under half-squat conditions.

	Exercise Condition	ICCs (Level of Reliability)	95% CI		SEM
			Lower	Upper	
Vastus medialis	Half-squat Floor	0.827 (Good)	0.57	0.94	0.11
	Half-squat Foam	0.934 (Excellent)	0.84	0.97	0.06
	Half-squat BOSU-up	0.859 (Good)	0.65	0.95	0.11
	Half-squat BOSU-down	0.772 (Good)	0.45	0.92	0.09
Vastus lateralis	Half-squat Floor	0.939 (Excellent)	0.85	0.98	0.06
	Half-squat Foam	0.816 (Good)	0.56	0.94	0.09
	Half-squat BOSU-up	0.846 (Good)	0.63	0.95	0.11
	Half-squat BOSU-down	0.820 (Good)	0.57	0.94	0.12
Biceps femoris	Half-squat Floor	0.937 (Excellent)	0.85	0.98	0.02
	Half-squat Foam	0.952 (Excellent)	0.89	0.98	0.02
	Half-squat BOSU-up	0.946 (Excellent)	0.87	0.98	0.04
	Half-squat BOSU-down	0.886 (Good)	0.70	0.96	0.05
Y-axis acceleration	Half-squat Floor	0.960 (Excellent)	0.90	0.99	0.47
	Half-squat Foam	0.792 (Good)	0.49	0.93	0.74
	Half-squat BOSU-up	0.859 (Good)	0.66	0.95	0.90
	Half-squat BOSU-down	0.908 (Excellent)	0.77	0.97	1.37
X-axis acceleration	Half-squat Floor	0.953 (Excellent)	0.89	0.98	0.49
	Half-squat Foam	0.843 (Good)	0.62	0.95	0.64
	Half-squat BOSU-up	0.919 (Excellent)	0.81	0.97	1.15
	Half-squat BOSU-down	0.830 (Good)	0.58	0.94	2.94

95% CI = 95% confidence interval; ICCs = Interclass correlation coefficients; SEM = Standard error of measurement.

Table 2. Linear mixed model with exercise condition and BCMA as the fixed effects and global activity as the dependent variable.

	Parameter	ES	SE	95%CI		Test (df)	*p*
				Lower	Upper		
Global activity	Intercept	0.83	0.24	0.35	1.31	t (54) = 3.460	0.001
	Half-squat Floor	0.76	0.12	−0.17	0.32	t (45) = 0.620	0.539
	Half-squat Foam	0.09	0.12	−0.15	0.34	t (45) = 0.728	0.470
	Half-squat BOSU-up	0.34	0.10	0.12	0.55	t (44) = 3.229	0.002
	BCMA	0.03	0.01	0.02	0.05	t (46) = 4.383	0.000
	σ_u				0.30		
	σ_ϵ				0.20		

ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t–value; *p* = *p*–value; BCMA = body centre of mass acceleration; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

Table 3. Linear mixed model with exercise condition as the fixed effects and muscle activity (vastus medialis, vastus lateralis, biceps femoris, and global activity) as the dependent variable.

	Parameter	ES	SE	95%CI		Test (df)	p
				Lower	Upper		
Vastus medialis	Intercept	0.73	0.06	0.60	0.85	t (20) = 12.116	0.000
	Half-squat Floor	−0.09	0.04	−0.17	−0.01	t (42) = −2.393	0.021
	Half-squat Foam	−0.11	0.04	−0.19	−0.03	t (42) = −2.886	0.006
	Half-squat BOSU-up	0.03	0.04	−0.05	0.11	t (42) = 0.721	0.475
	σ_u			0.19			
	σ_ϵ			0.10			
Vastus lateralis	Intercept	0.74	0.05	0.63	0.84	t (25) = 14.605	0.000
	Half-squat Floor	−0.15	0.04	−0.24	−0.06	t (42) = −3.532	0.001
	Half-squat Foam	−0.14	0.04	−0.22	−0.05	t (42) = −3.236	0.002
	Half-squat BOSU-up	−0.03	0.04	−0.12	0.05	t (42) = −0.821	0.416
	σ_u			0.15			
	σ_ϵ			0.11			
Biceps femoris	Intercept	0.33	0.02	0.27	0.38	t (31) = 11.875	0.000
	Half-squat Floor	−0.09	0.02	−0.15	−0.04	t (42) = −3.519	0.001
	Half-squat Foam	−0.07	0.02	−0.13	−0,02	t (42) = −2.763	0.008
	Half-squat BOSU-up	0.03	0.02	−0.02	0.08	t (42) = 1.199	0.237
	σ_u			0.07			
	σ_ϵ			0.07			
Global activity	Intercept	1.79	0.11	1.56	2.02	t (24) = 16.115	0.000
	Half-squat Floor	−0.34	0.09	−0.53	−0.16	t (42) = −3.794	0.000
	Half-squat Foam	−0.33	0.09	−0.51	−0.14	t (42) = −3.645	0.001
	Half-squat BOSU-up	0.02	0.09	−0.15	0.21	t (42) = 0.297	0.768
	σ_u			0.33			
	σ_ϵ			0.24			

ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t–value; p = p–value; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

Table 4. Root mean square surface electromyography values (mV) for each muscle analysed under half-squat conditions. Values are expressed as mean ± standard error of the mean (SE)**.**

	Half-Squat Floor	Half-Squat Foam	Half-Squat BOSU-Up	Half-Squat BOSU-Down
Vastus medialis	0.63 ± 0.06 †	0.61 ± 0.06 †‡	0.76 ± 0.07	0.73 ± 0.06
Vastus lateralis	0.59 ± 0.04 ‡	0.60 ± 0.05 ‡	0.70 ± 0.06	0.74 ± 0.07
Biceps femoris	0.23 ± 0.03 † ‡	0.25 ± 0.03 †	0.36 ± 0.03	0.33 ± 0.03

mV = microvolts; † Significantly different from half-squat BOSU-up; ‡ Significantly different from half-squat BOSU-down.

Table 5 shows the results of the linear mixed model between exercise condition and BCMA; a significant fixed effect for exercise condition [F (3,42) = 30.873 p = 0.000] was found on BCMA. The BCMA was significantly higher for the half-squat BOSU-down than half-squat floor (p = 0.000; d = 2.22), foam (p = 0.000; d = 2.28) and BOSU-up (p = 0.000; d = 1.53) (Figure 3b). For OMNI-Res, the exercise condition showed a significant main effect [X^2 (3) = 35.667 p = 0.000], and the OMNI-Res was significantly higher for half-squat BOSU-up and BOSU-down than half-squat floor (p = 0.006, d = 2.66; p = 0.008, d = 2.01, respectively) and foam (p = 0.005, d = 2.32; p = 0.009, d = 1.74, respectively) (Figure 3c). The raw data of this study is available as supplementary material.

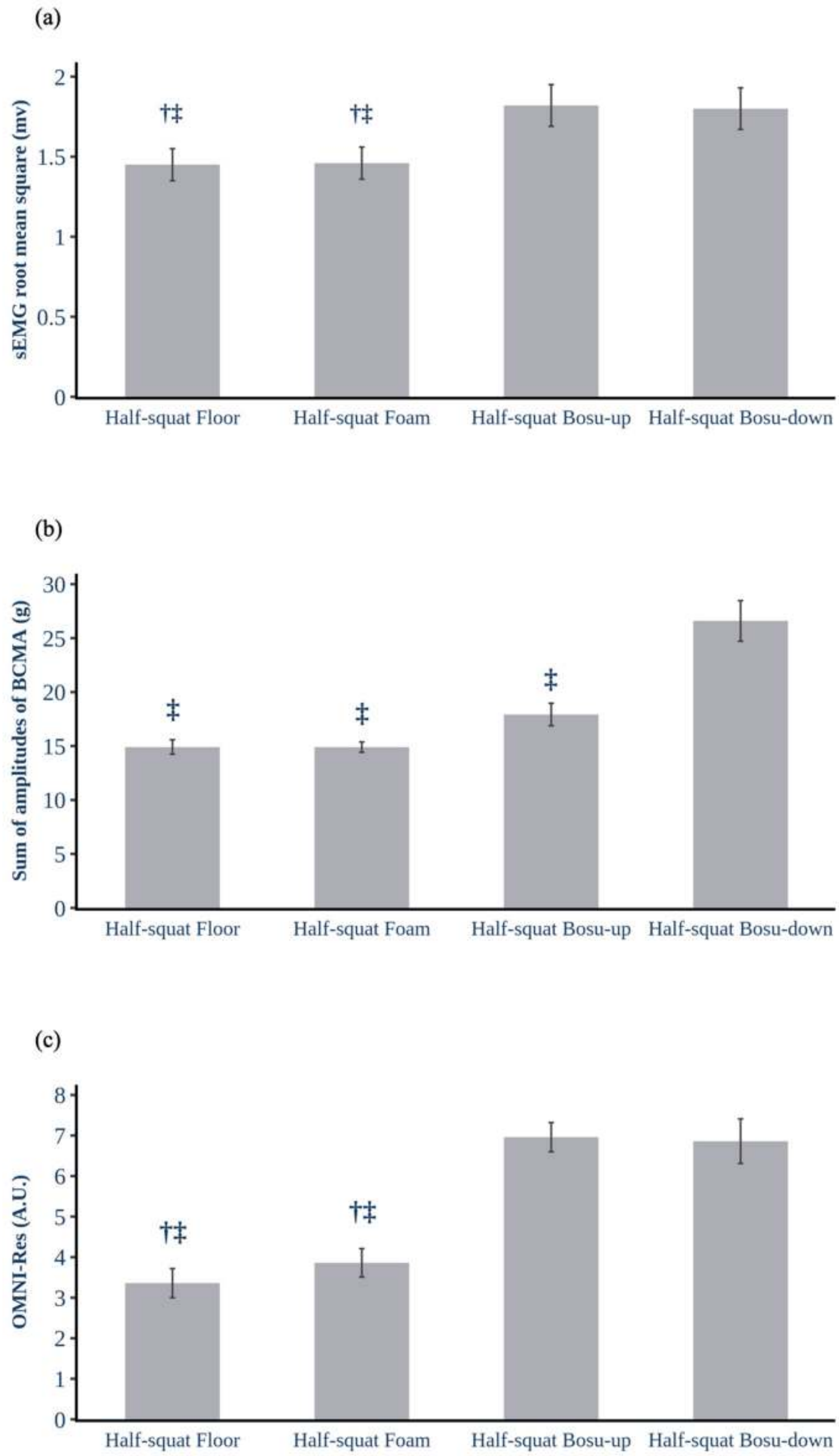

Figure 3. Comparison of the collected data under half-squat conditions: (**a**) global activity §, (**b**) body centre of mass acceleration, and (**c**) OMNI-Perceived Exertion Scale for Resistance Exercise (OMNI-Res). Each bar represents the mean, and the error bar represents the standard error of the mean (SE). § = Sum of the activity of the vastus medialis, lateralis and biceps femoris; sEMG = surface electromyography; mV = microvolts; BCMA = body centre of mass acceleration; A.U. = Arbitrary units; † Significantly different from half-squat BOSU-up; ‡ Significantly different from half-squat BOSU-down.

Table 5. Linear mixed model with exercise condition as the fixed effects and BCMA as the dependent variable.

	Parameter	ES	SE	95%CI Lower	95%CI Upper	Test (df)	*p*
BCMA	Intercept	26.59	1.10	24.37	28.81	t (50) = 24.043	0.000
	Half-squat floor	−11.69	1.41	−14.52	−8.85	t (42) = −8.307	0.000
	Half-squat foam	−11.69	1.41	−14.53	−8.85	t (42) = −8.309	0.000
	Half-squat BOSU-up	−8.67	1.41	−11.51	−5.83	t (42) = −6.166	0.000
	σ_u				1.80		
	σ_ϵ				3.72		

ES = coefficient estimate; SE = standard error; 95% CI = 95% confidence intervals; df = degrees of freedom; t = t-value; *p* = *p*-value; σ_u = standard deviation of participant; σ_ϵ = standard deviation of residual. We have used "half-squat BOSU-down" in the exercise condition variable as reference categories for this model.

4. Discussion

The first objective of the present study was to quantify the amount of instability in a half-squat using an accelerometer. The use of mean acceleration values might not be the best way to describe the amount of instability [15,19], and mean, or peak root mean square acceleration values do not reflect the ability to maintain the posture, because the moments when the participants are balanced are taken into consideration for the calculations [17]. Therefore, the sum of the peaks (Figure 2), considering the quadratic combination of the acceleration in anteroposterior and mid-lateral axes [29], seems to provide an accurate approach for quantifying the amount of instability (BCMA) in different unstable resistance training environments [30]. As expected, the results of the present study showed an increased BCMA from foam to BOSU-down conditions, and significant differences between all conditions and BOSU-down. The data also reflected differences between the two most stable conditions (floor and foam) and BOSU-up. This finding contributes to understanding the magnitudes of stability that a trained athlete experiences during the half-squat exercise on different unstable surfaces. The perturbation offered by the BOSU-down was the greatest, followed by the BOSU-up and the foam, and agreed with the Seaterbakken and Fimland [10] criteria to establish the magnitude of instability (unstable dimensions and magnitude of contact with the floor). Therefore, the BCMA does reflect how challenging it is for athletes to maintain their posture under the tested conditions, confirming the first hypothesis.

The second objective was to compare the global muscle activity, the rating of perceived exertion, and the BCMA during the execution of the half-squat under the four conditions. The analysis of variance showed a significant main effect for the three variables. The behaviour of muscle activity and OMNI-Res was similar, and significant differences were found in the BOSU compared to the floor and foam conditions. According to Andersen et al. [6], this study did not find significant differences in global muscle activation between stable and foam conditions. Moreover, although the authors reported differences in power and force outputs, Drinkwater et al. [14] found no significant differences between the foam and stable conditions in a loaded squat. However, when the load increased (100% of 1 RM), the foam condition became more 'stable', and the force output was higher in respect to other more unstable conditions (i.e., BOSU). In line with the studies mentioned earlier, the present results showed that the inclusion of foam pads during a squat might not be worthwhile for high-standard athletes, at least for increasing the activity of the knee extensor muscles. Furthermore, the use of high loads seemed to play a stabilizing role under unstable conditions, allowing higher muscle activation and, therefore, higher force production [10,14]. In the present study, participants squatted with extra loads corresponding to their body mass, which could have helped in stabilizing the posture and, consequently, perform higher muscle activity. Therefore, the second hypothesis was not confirmed in the athletes studied in the present investigation.

In recent years, the use of BOSU as a high-demand, unstable environment in strength and conditioning exercises has undoubtedly become widespread. BCMA described in the present study clearly shows the magnitude of the differences between BOSU and stable or foam conditions. In terms of muscle activity, the effects of performing squats on BOSU apparatus are unclear. Although McBride et al. [2] found lower muscle activity in the vastus lateralis and vastus medialis with similar unstable devices (Dyna Disc) in students, Saeterbakken and Fimland [10] found no significant differences in the same muscles comparing the stable condition with a Power Ball, BOSU, or Balance Cone in the same muscles in experienced resistance training participants. In contrast, the present study found higher muscle activity of the vastus lateralis and vastus medialis on the BOSU when compared to the more stable conditions (floor and foam). In the same vein, other studies [3,4,6], showed no significant lower-muscle activity in the biceps femoris under unstable conditions. Nevertheless, in line with Saeterbakken and Fimland [10], the present study found higher activity of this muscle in BOSU conditions. The experience of the athletes in the present study and their ability to maintain balance, even in the most perturbed conditions, might explain these differences. The contemporary trend of introducing unstable environments in training programs for experienced athletes might change the inhibiting effect of instability on the primary squat movers, and become a challenge for intramuscular coordination in highly trained and coordinated populations. Thus, using unstable resistance training exercises would force accommodation to an unstable environment, diminishing the loss of force and the extent of co-contractions [31]. Indeed, the present study was carried out with athletes who were able to perform squats in different conditions with good and excellent reliability scores (Table 1). Firstly, results confirmed the ability of the athletes to maintain the balanced posture in all conditions, including the most perturbed ones on the BOSU. Concretely, the BOSU-down condition presented the highest BCMA, but the athletes showed good and excellent reliability in both axes. These data reflect the excellent motor control of the athletes maintaining the posture in all conditions.

Regarding the differences in muscle activation between the two BOSU conditions, the present study found that the vastus medialis showed significantly higher activation in both BOSU conditions. It could be speculated that the tendency to avoid the dynamic knee valgus explains this finding. Indeed, although the BOSU-down condition created higher global instability, it offered a flat and rigid surface that compelled the participants to act differently in avoiding the knee valgus position. Although this study did not test this muscle, the role of the gluteus medius in stabilizing the posture can probably explain the lower activation of the vastus medialis in the BOSU-down [13,32,33] actions. Furthermore, the role of the biceps femoris co-contraction in the most unstable conditions seemed to be clear in a half-squat. In contrast to other studies [3,6,10], this study found significant increases in biceps femoris activation in the two BOSU conditions in comparison to the more stable conditions (floor and foam). The reason could be that BOSU creates higher anteroposterior instability. Only Saeterbakken and Fimland [10] used a BOSU, but the standard of their participants might explain the different findings.

The use of ratings of perceived exertion in resistance training exercises (OMNI-Res) is increasing. Its validity in terms of metabolic resistance training [25] and velocity-based training [34] has been pointed out. However, the relationship between perceived exertion and unstable environments is not clear [35,36]. The cited research investigated the effects of instability on bench press rating of perceived exertion in a trained population, but no research has studied the relationship between the amount of instability and muscle activity. In the present study, the OMNI-Res reflected similar increases to those in muscle activity throughout all the conditions. BCMA was slightly different concerning perceived exertion. The effect of performing a half-squat on a BOSU (up or down) caused almost the same perception of exertion, but the BOSU-down condition showed significantly higher BCMA than the BOSU-up condition. Therefore, beyond the instability role of the BOSU position demonstrated by the sum of BCMA values, muscle activity, and perceived exertion remained unchanged in both more unstable conditions (Figure 3).

There are several limitations to the present study. The particular characteristics of the sample, demonstrating high neuromuscular performance, prevents extrapolation of the results to the general population. The sample size, although the statistical power is acceptable, was also limited, as too was the number of analysed muscles. Further research should analyse the role of the stabilizers (e.g., gluteus maximus and medius, rectus abdominis, adductors, erector spinae) in the different conditions. Additionally, the present study was conducted using dynamic half-squats at 60 beats per minute. This controlled pace allowed an efficient and balanced execution, but the present results cannot be generalized to other rhythms and, of course, other motor skills. Further investigations should study this effect at different velocities and with explosive actions. Thus, the feedback provided in velocity-based resistance training might be complemented with BCMA data, monitoring how stable each repetition is. To summarize, the main strengths of the present proposal showed that the amount of instability can be quantified simply and suitably, especially on unstable surfaces, because nothing interferes with the relationship between the floor and the unstable device. In contrast, only the BCMA has been taken into account, but no acceleration measurements were obtained from other body parts such as the knee or the ankle. The data processing still requires the development of a proper algorithm for obtaining the BCMA in real time, while executing the movements.

5. Conclusions

The present study showed a higher muscle activity of the vastus lateralis, vastus medialis, and biceps femoris in BOSU conditions. This study contributes to understanding the magnitudes of stability that an athlete experiences during the squat exercise on different unstable surfaces. Moreover, OMNI-Res does not reflect the different level of perturbation (BCMA) found for the two BOSU positions, but this scale approximates the muscle activity of the primary movers in the studied half-squat conditions. Muscle activity in the primary half-squat movers increased under unstable conditions in elite athletes. These findings are in contrast to previous studies demonstrating insignificant differences between stable and unstable settings in this exercise. Experienced athletes and trained individuals showed different responses under unstable environments from those observed in other populations. Thus, the use of devices generating instability should be considered when the main objective is to increase the activity of the primary movers in this exercise and, potentially, in other exercises with similar muscular requirements. Therefore, the use of unstable conditions in strength and conditioning programs may increase variability, a crucial element to maintain chronic adaptations in long-term resistance training programs. Challenging experienced athletes by making their environment less stable seems to be a proper strategy to increase the acute responses and effects of lower-body resistance training. Nevertheless, the devices aimed at creating the mentioned challenging environments should be chosen accordingly to the ability of the individuals to control the movement while maintaining a balanced posture. Only by following this premise can the primary muscles be further activated to achieve better training effects. Thus, determining a BCMA limit could clarify how balanced the execution of a strength and conditioning exercise is, and the potential acute responses of the neuromuscular system. Moreover, monitoring the BCMA could be interesting in providing real-time feedback and quantifying the amount of instability in professional strength and conditioning contexts. The conclusions mentioned above, although in a very specific population, open up new possibilities in the fields of injury prevention and rehabilitation. As unilateral training revealed an essential element to be balancing the hamstring/quadriceps ratio, understanding which exercises generate more muscle activation when instability is a factor, and under what conditions they do so, allows a better prescription.

Supplementary Materials: The following materials are available online at http://www.mdpi.com/1660-4601/17/21/8046/s1, Spreadsheet S1: Raw data.

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Study 6

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sEMG Activity in Superimposed Vibration on Suspended Supine Bridge and Hamstring Curl

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Traditionally in strength and conditioning environments, vibration has been transmitted using platforms, barbells, dumbbells, or cables but not suspension devices. This study aimed to examine the effects on the lower limb of applying superimposed vibration on a suspension device. Twenty-one physically active men and women performed supine bridge and hamstring curl exercises in three suspended conditions (non-vibration, vibration at 25 Hz, and vibration at 40 Hz). In each exercise condition, the perceived exertion scale for resistance exercise (OMNI-Res) was registered, and the electromyographic signal was assessed for gastrocnemius (medialis and lateralis), biceps femoris, semitendinosus, gluteus maximus, and rectus femoris. A linear mixed model indicated a significant fixed effect for vibration at 25 Hz and 40 Hz on muscle activity in suspended supine bridge ($p < 0.05$), but no effect for suspended hamstring curl ($p > 0.05$). Likewise, the Friedman test showed a significant main effect for vibration at 25 Hz and 40 Hz in suspended supine bridge ($p < 0.05$) but not for suspended hamstring curl ($p > 0.05$) on OMNI-Res. *Post hoc* analysis for suspended supine bridge with vibration at 25 Hz showed a significant activation increase in gastrocnemius lateralis ($p = 0.008$), gastrocnemius medialis ($p = 0.000$), semitendinosus ($p = 0.003$) activity, and for semitendinosus under 40 Hz condition ($p = 0.001$) compared to the non-vibration condition. Furthermore, OMNI-Res was significantly higher for the suspended supine bridge at 25 Hz ($p = 0.003$) and 40 Hz ($p = 0.000$) than for the non-vibration condition. Superimposed vibration at 25 Hz elicits a higher neuromuscular response during the suspended supine bridge, and the increase in vibration frequency also raises the OMNI-Res value.

Keywords: instability, vibration, lower limb, suspension training, electromyography

INTRODUCTION

Nowadays, strength and conditioning practices combine resistance exercises and other training methods such as eccentric overloads, unstable surfaces, and suspension devices for improving strength and power performance (Maté-Muñoz et al., 2014; Behm et al., 2015; Suchomel et al., 2019). Similarly, coaches and fitness enthusiasts have also used mechanical vibrations as an alternative or complement to strength and explosive training (Hammer et al., 2018). The effects of vibration training have been widely studied on neuromuscular performance (Alam et al., 2018), flexibility (Fowler et al., 2019), and balance control (Ritzmann et al., 2014; Sierra-Guzmán et al., 2018). This method transfers the vibratory stimulus on the muscle belly and tendon directly (local) or indirectly (e.g., vibrating platforms) to elicit the tonic vibration reflex (Cardinale and Bosco, 2003). Platforms are the most commonly used piece of equipment in sports training to transfer whole-body vibration (WBV) and modify the stimulus through the type of vibration (side-alternating vibration or synchronous vibration), frequency (in Hz), amplitude (peak to peak amplitude), position, and time of exposure (Cardinale and Wakeling, 2005; Issurin, 2005).

WBV has been combined with different training methods, and lower-body resistance exercises (bodyweight or extra loads) performed under static and dynamic conditions (Rittweger, 2010). Several studies have shown the positive effects of performing WBV squats or other exercises such as lunges or Bulgarian squats on muscle strength and jump ability (Rehn et al., 2006; Fort et al., 2012; Osawa et al., 2013). However, the effect of vibration training on dynamic exercises with heavy loads (squats) did not improve maximal strength and jump performance using WBV at 40 Hz (Rønnestad, 2004) or 50 Hz at < 1 mm of amplitude (Hammer et al., 2018). Contrarily, dynamic squat training (6 sets of 6 reps; with an individual optimal load) performed on a vibration platform (30 Hz at 4 mm of amplitude) combined with repeated sprint training (3 sets of 6 reps of 20 meters shuttle run with 180° change of direction) (Suarez-Arrones et al., 2014) or functional eccentric-overload exercises (8 exercises between 6 to 10 reps with an inertial load ranged from 0.27 $Kg{\cdot}m^{-2}$ to 0.11 $Kg{\cdot}m^{-2}$) (Tous-Fajardo et al., 2016) elicited higher performance than traditional resistance training (lunges, half-squats, and calf raises; 50–100% body mass) on sprint, change of direction, and jumping performance. Furthermore, blood flow restriction training combined with WBV resistance training (30 Hz and parallel squat with dynamic loading) improved critical power, overall capillary-to-fiber ratio, and total lean body mass in endurance-trained men (Mueller et al., 2014). Considering acute effects, Bush et al. (2015) reported a post-activation potentiation effect on knee extension torque after exposing healthy participants to a WBV dynamic squat with bodyweight resistance (30 Hz and 4 mm of amplitude). Additionally, Aguilera-Castells et al. (2019) showed that combined WBV (40 Hz) with a suspended device elicited higher muscle activity than the suspended condition for hip and thigh muscles in the dynamic lunge bodyweight resistance.

In the studies mentioned above, the WBV was provided with a vibration platform to assess the effects of combining vibration and resistance training on different neuromuscular performance variables such as maximal strength, mechanical power, jumping ability, or muscle activity. However, to transfer the vibratory stimulus to the upper body, several devices with superimposed vibration have been used in the past, such as dumbbells (Bosco et al., 1999; Cochrane and Hawke, 2007), bars (Poston et al., 2007; Mischi and Cardinale, 2009; Moras et al., 2010; Xu et al., 2013), and cables (Issurin and Tenenbaum, 1999; Issurin, 2010). Likewise, superimposed vibration has been used to study the training effects on the lower body. Thus, the addition of vibration (30 Hz at 2.5 mm of amplitude) had no effects during four weeks of dynamic calf-raise on a seated rig (75–90% 1RM) (Carson et al., 2010). However, superimposed vibration on a BOSU (35–40 Hz and 2 to 4 mm of amplitude) enhanced the reaction time of peroneus brevis, longus, and tibialis anterior in athletes with chronic ankle instability during six weeks of training (Sierra-Guzmán et al., 2017). Furthermore, surface electromyography (sEMG) has been used to evaluate the activity of different muscles during an exercise with superimposed vibration (Xu et al., 2015). Thus, Marín and Hazell (2014) found higher activation of the gastrocnemius medialis, vastus medialis, and multifidus during 60° knee flexion static half-squats with superimposed vibration on a BOSU (30 Hz and 50 Hz and 1 mm of amplitude) in comparison to the stable condition. To the best of our knowledge, there are only four devices with superimposed vibration allowing the lower body training. Two of these devices are similar to vibration platforms, consisting of a small platform to improve flexibility in gymnasts (Sands et al., 2006; Kinser et al., 2008) and a platform with a bi-engine that provides vibration on a leg press machine (Pujari et al., 2019). The other two devices are Vibrosphere (ProMedvi), a superimposed vibration wobble board (Cloak et al., 2013), and Vibalance (Viequipment), a platform that combines vibration with different degrees of instability even though neither of these devices superimposed vibration on suspension straps.

Although the squat and its variations are the most used resistance exercises in WBV, the most demanded actions in team sports are sprinting, jumping, and cutting, generating numerous lateral actions and unilateral movements that demand horizontal force production (Gonzalo-Skok et al., 2017). Hence, the use of functional equipment such as suspension straps allowing exercises in multiple planes (Bettendorf, 2010), the inclusion of exercises based on the force-vector theory such as the barbell hip thrust to improve horizontal force production (Loturco et al., 2018; Neto et al., 2019), and preventive training on the hamstrings muscle complex (Rey et al., 2017; Bourne et al., 2018a) are commonly used in strength and conditioning team-sport programs. In the last decade, injuries to the hamstrings complex have increased in different team sports, especially in soccer, with an injury rate ranging between 15 and 50% (Al Attar et al., 2017). To strengthen the hamstrings complex (biceps femoris, semitendinosus, and semimembranosus), different bilateral and unilateral exercises, such as the deadlift, supine bridge, leg curl, glute-ham raise, or Nordic Hamstring have been included in injury prevention programs (Bourne et al., 2017). Thus, the suspended supine bridge and the hamstring curl were selected in the current study because of their popularity in hamstrings preventive programs (Malliaropoulos et al., 2012; Youdas et al., 2015). On the one hand, the supine bridge is a bodyweight exercise demanding the posterior hip and thigh muscles as gluteus maximus and hamstrings (Jang et al., 2013; Kim and Park, 2016; Lehecka et al., 2017; Marín and Cochrane, 2021), and it is a recommended exercise for strengthening and prevent injuries in hamstrings and lower back muscles (Ekstrom et al., 2007). This exercise is considered a variation of the hip thrust, where back and feet are placed on the ground, thus increasing the difficulty by modifying the position of the feet on a bench or an unstable surface (i.e., suspension device) (Tobey and Mike, 2018). Conversely, the hamstring curl is considered an open kinetic chain knee dominant exercise (Malliaropoulos et al., 2015) that uses body weight as resistance and aims to develop the strength and endurance of the hamstring muscles (Dawes, 2017).

Accordingly, a vibratory system for suspension training has been designed to provide an indirect and superimposed vibration on the suspension device, allowing a wide range of exercises in different planes. Therefore, the main objective of the present study was to examine the effects of the vibration device on muscle activation in the dynamic suspended supine bridge and hamstring curl exercises. It was hypothesized that the superimposed vibration on the suspension device would obtain a superior muscle activation than the suspended condition without vibration in both exercises. Additionally, it was also hypothesized that the OMNI-Res perceived exertion scale for resistance exercise would be higher in the suspended condition with vibration than the condition without vibration in each of the two exercises.

MATERIALS AND METHODS

Participants

Twenty-one physically active participants males (n = 15, mean age = 23.3 ± 2.8 years, height = 1.8 ± 0.0 m, body mass = 77.8 ± 6.9 kg, body mass index = 24.1 ± 1.8 kg·m^{-2}, suspension training experience = 4.2 ± 1.5 years) and females (n = 6, mean age = 22.6 ± 1.0 years, height = 1.6 ± 0.0 m, body mass = 56.6 ± 2.9 kg, body mass index = 21.5 ± 1.7 kg·m^{-2}, suspension training experience = 3.8 ± 1.9 years) were voluntarily recruited to take part in the study. Participants experienced in suspension training for less than one year, not performing 30 min of physical activity at least three times a week, or having pain or injury related to cardiovascular, musculoskeletal, or neurological diseases were excluded from the study. Additionally, before the familiarization session, an informed consent form was provided and signed by all participants after receiving a detailed explanation, both in verbal and written form, of the experimental procedures, benefits, and risks of participating in the study. They also answered the Physical Activity Readiness Questionnaire (PAR-Q) to determine potential health risks associated with physical exercise (Warburton et al., 2011). Before the familiarization and test session, all participants were asked to refrain from high-intensity physical activity 24 h before the test session and avoid drinking, eating, or consuming stimulant substances (e.g., caffeine) 3–4 h before the test session. This study was approved by the Ethics and Research Committee Board in the Blanquerna Faculty of Psychology and Educational and Sport Sciences at Ramon Llull University in Barcelona, Spain, with reference number 1819005D. The requirements specified in the Declaration of Helsinki (revised in Fortaleza, Brazil, 2013) were complied with and implemented in all study protocols.

Experimental Design

A cross-sectional study design was carried out to determine the effect of a vibratory system for suspension training on muscle activation in different lower limb muscles. Participants performed supine bridge and hamstring curl exercises in three suspension conditions: (a) non-vibration, (b) vibration at 25 Hz, and (c) vibration at 40 Hz. In all the above-mentioned conditions, muscle activation of the rectus femoris, biceps femoris, semitendinosus, gluteus maximus, gastrocnemius medialis, and lateralis was assessed and compared using sEMG. Muscle activation was normalized and expressed as a percentage of maximum voluntary isometric contraction (% MVIC). In addition, the OMNI-Perceived Exertion Scale for Resistance Exercise (OMNI-Res) was recorded to compare perceived exertion in each exercise condition.

Procedures

A familiarization session was conducted one week in advance of the test session. In this session, participants performed two sets of five repetitions of each supine bridge and hamstring curl under suspended conditions (non-vibration, vibration at 25 Hz and 40 Hz), and the researchers collected anthropometric data such as age, height, and weight. The test session took place one week later in the morning at the same time as the familiarization session. The test session began with a standardized warm-up consisting of 10 min of cycle ergometer while maintaining a cadence of 100 W at 60 revolutions per minute, two sets of eight repetitions of a unilateral stiff-leg deadlift, two sets of five repetitions of Nordic hamstring assisted with an elastic band, and two sets of eight repetitions of unilateral straight knee bridge. Next, surface electrodes were placed on the dominant lower limb (Criswell and Cram, 2011), which was established subjectively by asking participants which leg they would use to kick a soccer ball (Meylan et al., 2009). Before performing the different supine bridge and hamstring curl conditions, maximal voluntary isometric contraction (MVIC) tests were performed on the rectus femoris, biceps femoris, semitendinosus, gluteus maximus, gastrocnemius medialis, and lateralis in order to obtain a baseline value and normalize the electromyographic signal (Halaki and Ginn, 2012). Afterward, participants performed the different supine bridge and hamstring curl conditions in a randomized order. For the suspended supine bridge exercise, the distance between the crista iliac and the cradle of the suspension device was standardized as 75% of the leg length, and the hip elevation was controlled with customized stoppers (similar to hurdles), starting the exercise with the lower back, arms, and hands in contact with the ground (**Figure 1**). For the suspended hamstring curl, the distance between the crista iliac and the device's cradles was also 75% of the leg length, and the starting position of the exercise was standardized by laying the lower back and gluteus on a foam surface with a height corresponding to 20% of the leg length. Participants were instructed to begin with a complete knee extension in this exercise, release the lower back and gluteus on the foam surface, keep their arms and hands flat on the floor, perform a knee flexion, and then return to the starting position (**Figure 2**). The participants were instructed

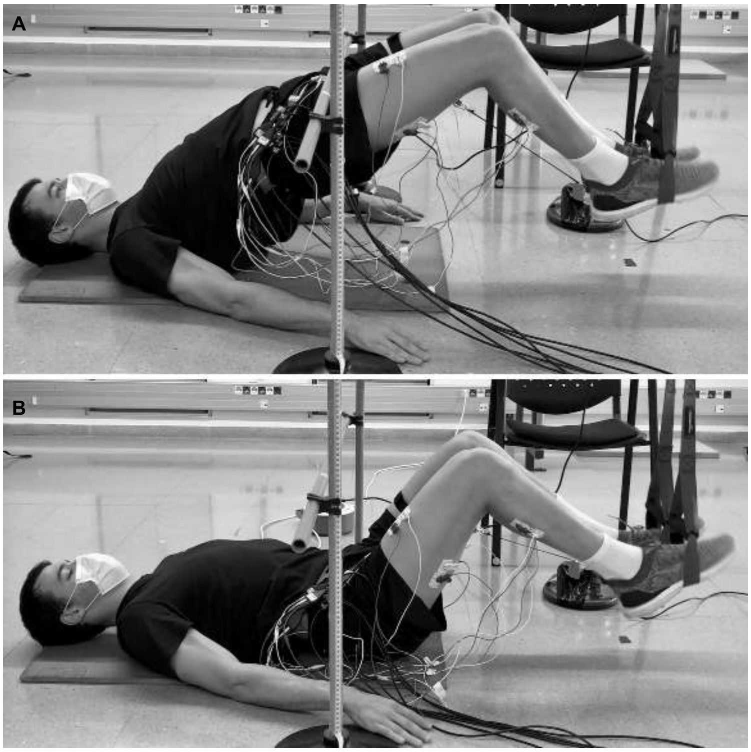

FIGURE 1 | Suspended supine bridge: upper **(A)** and lower **(B)** position.

to place their feet inside the suspension device cradles with plantar flexion and to hold this position during all the repetitions in both exercises.

From each dynamic condition of the exercise, participants performed five repetitions with a two-minute rest between attempts. The pace of each repetition was controlled with a metronome giving a rate of 60 beats per minute, and the range of movement was controlled with a positional encoder (WSB 16k-200; ASM Inc., Moosinning, DEU) by attaching the tether to the thigh or the cradle of the suspension device in the supine bridge and the hamstring curl, respectively.

The movement signal recorded by the positional encoder in each repetition of the exercises was used to determine the concentric and eccentric phases of the movement. The positional encoder signal was divided in two for each repetition, establishing that the concentric phase or the ascent phase for the suspended supine bridge ranged from the initial position to the maximum hip extension (highest position) and for the suspended hamstring curl from the initial position to the knee flexion (highest position). In both exercises, the eccentric phase ranged from the highest position to the initial position (lowest position). The positional encoder determined the beginning and the end of each repetition, thus establishing the range of motion in the same acquisition timeline of the BIOPAC MP-150 system (BIOPAC System, Inc., Goleta, CA, United States) sEMG signal. Those attempts that did not follow the proper technical execution indicated by the researchers were discarded and repeated, providing the two-minute rest between trials. A TRX Suspension Trainer (Fitness Anywhere, San Francisco, CA, United States) was used for both exercises, with the device anchored to the ceiling. The distance between the floor and the suspension device cradles was standardized as 30% of the leg length of each participant. A vibratory suspension training system was used under vibration conditions (25 Hz and 40 Hz) and fixed between the ceiling anchor point and the suspension device. The vibratory system provided vibration to the suspension device by converting the rotary motion of an electric motor into a vertical motion, which caused the displacement of a connecting rod with an amplitude of 8 mm (peak to peak), and the motor rotation frequency was regulated with a potentiometer.

Electromyography

The recording and analysis of sEMG of each muscle during each repetition under the suspended supine bridge and hamstring curl conditions (non-vibration, vibration at 25 Hz and 40 Hz) was performed with a six-channel BIOPAC MP-150 (sampling rate: 1.0 kHz) and AcqKnowledge 4.2 software (BIOPAC System, Inc., Goleta, CA, United States). Before placing the electrodes (Biopac EL504 disposable Ag-AgCl) over the rectus femoris, biceps femoris, semitendinosus, gluteus maximus, gastrocnemius medialis, and lateralis from the dominant leg, the skin area of the participants was prepared by shaving, exfoliating, and cleaning with alcohol to reduce impedance from dead surface tissues and oils. Following SENIAM recommendations (Hermens et al., 2000), the rectus femoris electrodes were placed at half the distance between the anterior superior iliac spine and

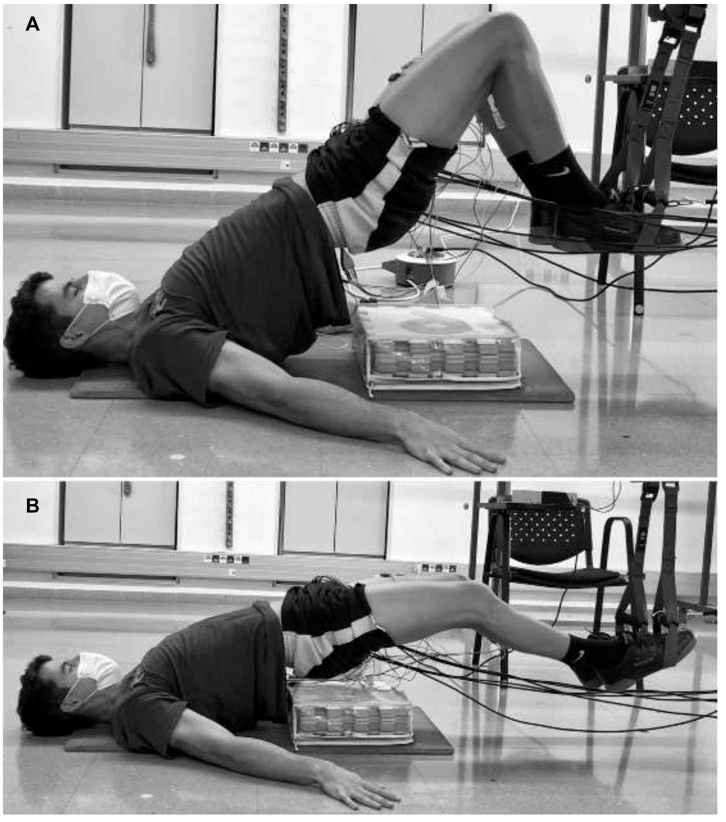

FIGURE 2 | Suspended hamstring curl: upper **(A)** and lower **(B)** position.

the superior part of the patella; for the biceps femoris and semitendinosus at half the distance between the ischial tuberosity and the lateral epicondyle (biceps femoris) or medial epicondyle (semitendinosus) of the tibia; the gluteus maximus at half the distance from the sacral vertebrae and the greater trochanter; for the gastrocnemius medialis over the most prominent bulge of the muscle, and in the gastrocnemius lateralis at 1/3 of the distance between the head of the fibula and the heel. All electrodes were placed at an inter-electrode distance of 2 cm and were oriented longitudinally to the direction of the muscle fibers. In addition, a reference electrode was placed on the crista iliac. The sEMG signal was bandpass filtered at 10–500 Hz using a 4th order 50 Hz Butterworth notch filter, and the root mean square (RMS) was calculated. In order to normalize the results of muscle activation of each of the muscles analyzed, MVIC tests were performed on the dominant leg with three MVIC of five seconds, recruiting gradually up to the maximum for two seconds and maintaining the MVIC for three seconds, with a three-minute rest between MVIC following Jakobsen et al.'s (2013) procedures. The position of each muscle used to achieve the MVIC was based on Konrad's (2006) protocol. Thus, the MVIC for the rectus femoris consisted of 90° seated single-leg knee extension; the MVIC for the biceps femoris and semitendinosus of 20–30° prone-lying single-leg knee flexion; the MVIC for the gluteus maximus in a supine-lying single hip extension; and the MVIC for the gastrocnemius medialis and lateralis in 90° seated ankle plantar flexion. All MVIC tests were against an immovable resistance; for the rectus femoris, biceps femoris, semitendinosus, and gluteus maximus, an ankle brace was used that was attached to a cable anchored to a stretcher. For the gastrocnemius medialis and lateralis, a horizontal leg press machine was used. The MVIC values obtained in each muscle mentioned above were used to normalize the RMS signal and report the muscle activation as % MVIC. For each exercise condition, the peak sEMG of each studied muscle during the concentric (ascending trajectory), and eccentric (descending trajectory) phase was analyzed, excluding the first and fifth repetition from the data analysis. Additionally, muscle activation levels recorded under the supine bridge and hamstring curl conditions were categorized as very high ($>$ 60% MVIC), high (41–60% MVIC, moderate (21–40% MVIC), and low ($<$ 21% MVIC) (Escamilla et al., 2010).

OMNI-Perceived Exertion Scale for Resistance Exercise

This scale was used to register the perceived subjective exertion experienced during the suspended supine bridge and hamstring curl conditions (non-vibration, vibration at 25 Hz and 40 Hz). Once participants completed an exercise condition, they were asked to assess their perception of exertion. Participants were instructed during the familiarization session to follow the instructions for the OMNI-Res assessment by Robertson et al.

(2003). During the familiarization and test session, a visual OMNI-Res scale was used, through which participants indicated the value of perceived exertion on a range from 0 to 10, where 0 indicated an extremely easy exertion (perception lower than that experienced during an unweighted repetition) and 10 an extremely hard exertion (perception higher than that experienced lifting 1 RM). The OMNI-Res values for each exercise condition were analyzed as mean OMNI-Res.

Statistical Analysis

Statistical data analyses were carried out using the SPSS statistical package version 26 (SPSS Inc., Chicago, IL, United States). G*Power (version 3.1.9.6; University of Dusseldorf, Dusseldorf, Germany) was used to calculate the sample size with power analysis and determined an effect size 0.29 SD with an α level of 0.05 and power at 0.95. All dependent variables showed a normal distribution, confirmed with the Shapiro-Wilk test, and met the inferential parametric assumptions, except the OMNI-Res. The global activity variable was calculated as the global mean of the six analyzed muscles. The effect of exercise condition on muscle activation (rectus femoris, biceps femoris, semitendinosus, gluteus maximus, gastrocnemius medialis and lateralis, and global activity) was assessed using a linear mixed model analysis considering the activation of each muscle as the dependent variable, the exercise condition as the fixed effect and the participants as a random effect. In case of a significant fixed effect, *post hoc* comparisons were made. Moreover, a non-parametric Friedman test was carried out to determine the effect of exercise conditions on the OMNI-Res. For significant main effects, a *post hoc* Wilcoxon test analysis with Bonferroni correction was applied. For pairwise comparison, Cohen's d effect size (Cohen, 1988) and 90% confidence intervals (CI) were also calculated. Effect size was interpreted as trivial ($d < 0.2$), small (d ranging from 0.2 to 0.6), moderate (d ranging from 0.6 to 1.2), large (d ranging from 1.2 to 2.0), and very large ($d > 2.0$) (Hopkins et al., 2009). Statistical significance was set at $p < 0.05$, and all data were expressed as mean ± standard error of the mean (SE).

RESULTS

The sEMG activity of each muscle and the global activity during the concentric and eccentric phase of the suspended supine bridge and the suspended hamstring curl under non-vibration, vibration at 25 Hz, and 40 Hz conditions are shown in **Tables 1**, **2**, respectively. Moreover, for the percentage of change of the analyzed muscles in the different suspended supine bridge and hamstring curl conditions, see **Supplementary Tables 1, 2**, respectively.

Suspended Supine Bridge

Supplementary Tables 3–5 shows the linear mixed model results. A significant fixed effect for exercise condition indicated that during the concentric phase, the suspended supine bridge with 25 Hz vibration showed a small increase with non-vibration condition for semitendinosus ($p = 0.003$, $d = 0.47$), gastrocnemius lateralis ($p = 0.008$, $d = 0.36$), and global activity ($p = 0.000$, $d = 0.60$). Moreover, the aforementioned conditions presented a moderate increase for gastrocnemius medialis (non-vibration vs 25 Hz vibration: $p = 0.000$, $d = 0.75$). The suspended supine bridge with 25 Hz vibration showed a small decrease with vibration at 40 Hz condition for gastrocnemius medialis ($p = 0.025$,

TABLE 1 | The sEMG activity for each analyzed muscle under suspended supine bridge conditions.

Exercise phase	Muscle group	Suspended supine bridge				
		Non-Vibration	Vibration at 25 Hz	Vibration at40 Hz		
		Mean ± SE	Mean ± SE	Mean ± SE	F	p
Concentric	Rectus femoris	1.7 ± 0.3	1.8 ± 0.4	2.0 ± 0.5	0.20	0.815
	Biceps femoris	19.1 ± 1.6	20.2 ± 1.6	19.6 ± 1.8	0.72	0.490
	Semitendinosus	19.7 ± 1.4	22.9 ± 1.5[a]	23.2 ± 1.7[a]	9.05	0.001
	Gluteus maximus	14.8 ± 1.7	16.1 ± 2.3	16.6 ± 2.2	1.79	0.178
	Gastrocnemius medialis	30.2 ± 2.0	37.4 ± 2.1[ab]	32.8 ± 1.8	9.71	0.000
	Gastrocnemius lateralis	36.5 ± 3.1	41.7 ± 3.1[a]	38.6 ± 3.1	5.19	0.010
	Global activity	20.3 ± 1.1	23.4 ± 1.0[a]	22.1 ± 1.1[a]	16.51	0.000
Eccentric	Rectus femoris	2.0 ± 0.3	1.9 ± 0.3	2.0 ± 0.3	0.25	0.780
	Biceps femoris	14.5 ± 1.3	16.5 ± 1.7	14.7 ± 1.4	3.11	0.055
	Semitendinosus	16.5 ± 1.3	18.1 ± 1.2[a]	18.3 ± 1.3[a]	4.73	0.014
	Gluteus maximus	8.6 ± 1.0	8.3 ± 0.8	8.6 ± 1.0	0.19	0.822
	Gastrocnemius medialis	24.4 ± 1.8	29.9 ± 1.9[a]	27.5 ± 1.9	8.91	0.001
	Gastrocnemius lateralis	37.6 ± 3.2	39.0 ± 2.9	36.4 ± 2.8	1.24	0.198
	Global activity	17.3 ± 0.9	18.9 ± 0.9[a]	17.9 ± 0.9	7.39	0.002

Data presented as normalized muscle activity (%MVIC); SE, standard error of the mean; Global activity, mean of the six muscles; [a]significantly different with non-vibration condition; [b]significantly different with vibration at 40 Hz condition.

TABLE 2 | The sEMG activity for each analyzed muscle under suspended hamstring curl conditions.

Exercise phase	Muscle group	Suspended hamstring curl				
		Non-Vibration	Vibration at 25 Hz	Vibration at 40 Hz	F	p
		Mean ± SE	Mean ± SE	Mean ± SE		
Concentric	Rectus femoris	1.3 ± 0.1	1.4 ± 0.1	1.2 ± 0.1	1.13	0.330
	Biceps femoris	23.6 ± 1.4	23.7 ± 1.3	24.0 ± 1.6	0.04	0.955
	Semitendinosus	24.9 ± 1.7	26.2 ± 1.6	25.8 ± 1.7	0.72	0.490
	Gluteus maximus	12.7 ± 1.1	13.1 ± 1.4	12.9 ± 1.1	0.16	0.848
	Gastrocnemius medialis	37.0 ± 3.0	37.6 ± 2.0	40.8 ± 3.4	1.61	0.210
	Gastrocnemius lateralis	52.8 ± 3.7	57.5 ± 3.8	56.2 ± 3.9	1.88	0.165
	Global activity	25.4 ± 1.1	26.5 ± 1.0	26.8 ± 1.2	2.60	0.086
Eccentric	Rectus femoris	1.4 ± 0.2	1.5 ± 0.2	1.8 ± 0.3	1.14	0.329
	Biceps femoris	22.0 ± 1.4	24.5 ± 1.7	22.6 ± 1.6	1.61	0.211
	Semitendinosus	20.6 ± 1.1	22.9 ± 1.5	22.5 ± 1.9	2.01	0.146
	Gluteus maximus	10.0 ± 0.8	11.7 ± 1.1	11.4 ± 1.0	3.48	0.060
	Gastrocnemius medialis	36.3 ± 2.1	37.0 ± 2.2	37.1 ± 2.2	0.17	0.838
	Gastrocnemius lateralis	51.5 ± 3.7	50.8 ± 3.6	51.2 ± 4.4	0.06	0.940
	Global activity	23.6 ± 0.9	24.7 ± 1.0	24.4 ± 1.1	1.85	0.169

Data presented as normalized muscle activity (%MVIC); SE, standard error of the mean; Global activity, mean of the six muscles.

d = −0.50). The semitendinosus and global activity showed a small increase between suspended supine bridge with 40 Hz vibration and non-vibration (p = 0.001, d = 0.46; p = 0.005, d = 0.34, respectively). For eccentric phase, the suspended supine bridge with 25 Hz vibration showed a small increase with non-vibration condition for semitendinosus (p = 0.046, d = 0.28) and global activity (p = 0.001, d = 0.40) and a moderate increase for gastrocnemius medialis (p = 0.000, d = 0.63). Additionally, the suspended supine bridge with 40 Hz vibration presented a small increase with non-vibration condition for semitendinosus (p = 0.024, d = 0.29). The standardized differences, expressed as Cohen d effect size, between exercise condition and muscle activity are shown detailed in **Figure 3**.

Suspended Hamstring Curl

The linear mixed model results are shown in **Supplementary Tables 6–8**. A non-significant fixed effect for exercise condition during the concentric phase neither eccentric phase was found on the analyzed muscles (**Table 2**). Additionally, the effect size analysis is shown in **Figure 4**.

OMNI-Perceived Exertion Scale for Resistance Exercise

Friedman test showed a significant main effect for suspended supine bridge [X^2 (2) = 26.462, p = 0.000] but not for suspended hamstring curl [X^2 (2) = 6.333, p = 0.052] on the OMNI-Res. Pairwise comparison showed a significantly higher OMNI-Res for suspended supine bridge with vibration at 40 Hz (4.86 ± 0.37) than for vibration at 25 Hz (4.33 ± 0.35, p = 0.024, d = 0.32 CI = −0.19, 0.83) and non-vibration condition (3.67 ± 0.40, p = 0.000, d = 0.67 CI = 0.15, 1.19). Moreover, OMNI-Res was significantly higher for suspended supine bridge with vibration at 25 Hz than for non-vibration condition (p = 0.003, d = 0.38 CI = −0.13, 0.89) (**Figure 5**). **Supplementary Table 9** shows the percentage of change for the OMNI-Res under suspended supine bridge and suspended hamstring curl conditions.

DISCUSSION

Superimposed vibration in a suspension device increased lower limb muscle activity in the supine bridge but not in the hamstring curl exercise. In the suspended supine bridge, a significant moderate increase of 14.8% (concentric phase) and a small increase of 9.7% (eccentric phase) was found under the 25 Hz vibration condition compared to the non-vibration global activity. Likewise, 40 Hz vibration significantly increased global activation by 8.7% (a small increase) during the concentric phase. Similarly, Marín and Hazell (2014) applied superimposed 30 Hz vibration on an unstable surface (BOSU) and found a higher muscle activity between 23.5% and 35% in the isometric half-squat compared to the unstable condition. The effect of additional vibration (30 Hz and 40 Hz with an amplitude of 4 mm) on unstable surfaces and suspension devices increased the demands of the exercise. Thus, eliciting a greater activation of the lower limb muscles (vastus medialis and lateralis, biceps femoris, and gluteus medius) during the suspended lunge combined with 40 Hz WBV than in unstable or suspended exercises without vibration (Aguilera-Castells et al., 2019). Understanding what exercises generate more muscle activation and under what conditions they do so is essential for practitioners. Previous scientific research reveals that different tasks involving the same muscle groups can present significantly different activation levels (Malliaropoulos et al., 2015); these findings are relevant in injury prevention and rehabilitation.

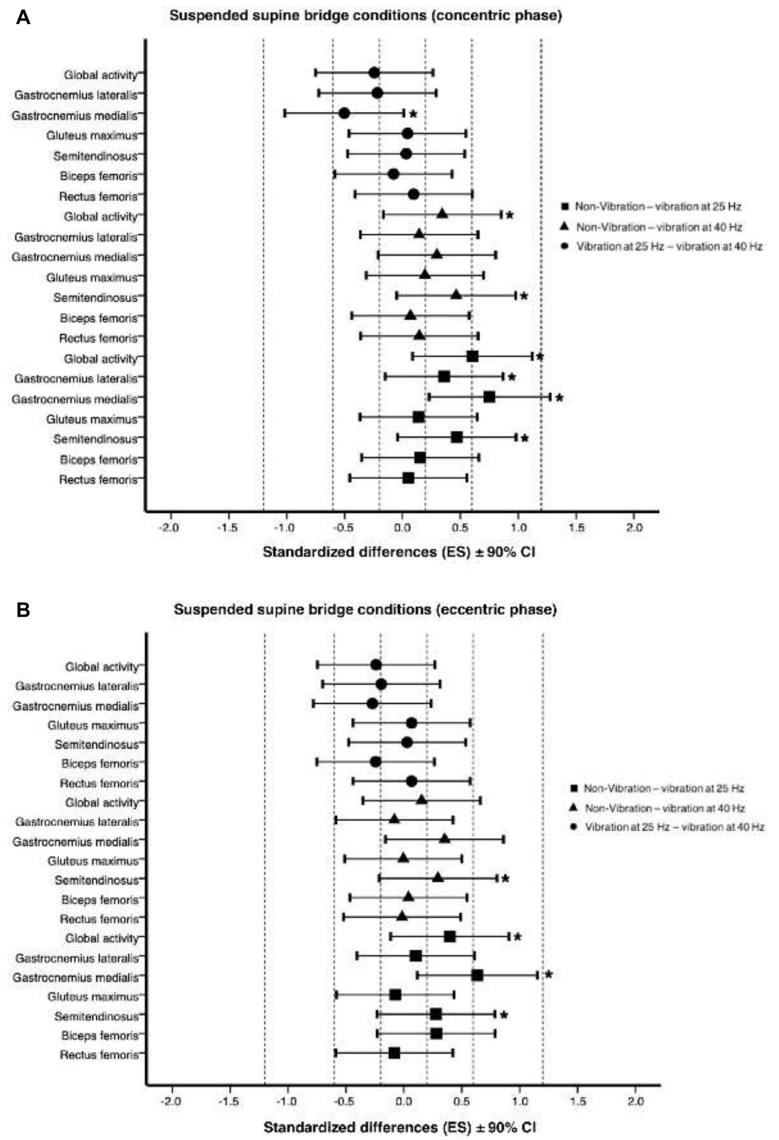

FIGURE 3 | Effects of suspended supine bridge conditions on muscle activity (%MVIC) at concentric **(A)** and eccentric phase **(B)** expressed as standardized differences (Cohen's *d*) ± 90% CI. Dotted line represents the effect size thresholds. * Significant differences at $p < 0.05$. ES, effect size; CI, confidence interval.

The effect of two different frequencies was studied in the present study, finding a small to moderate significant increase in semitendinosus, gastrocnemius medialis, and lateralis activation under 25 Hz vibration compared to the non-vibration condition. Likewise, there was a significantly small decrease in the gastrocnemius medialis activity at 40 Hz (**Figure 3**). Furthermore, no significant differences were found among frequencies for the other analyzed muscles. Overall, this study showed that performing the 25 Hz suspended supine bridge elicits a greater activation than at 40 Hz vibration in almost all the analyzed muscles. In the same vein, a progressive increase in vibration frequency (5 Hz to 30 Hz) gradually enhanced the neuromuscular response for the lower limb muscles (soleus, gastrocnemius, tibialis anterior, biceps femoris, vastus medialis, and rectus femoris), achieving the highest activations at 25 to 30 Hz frequencies (Ritzmann et al., 2013). On the other hand, 25 Hz vibration was consistently more demanding than 40 Hz vibration [concentric phase: biceps femoris (−3.0%, trivial), gastrocnemius medialis (−12.2%, small decrease), gastrocnemius lateralis (−7.4%, small decrease), global activity (−5.2%, small decrease); eccentric phase: biceps femoris (−10. 5%, small decrease), gastrocnemius medialis: (−8.0%, small decrease), gastrocnemius lateralis (−6.6%, trivial), global activity (−5.4%, small decrease)], per Cardinale and Lim (2003), who found lower but not significant muscle activity of 40 Hz vibration compared to 30 Hz. Regarding the effect of the different frequencies on the analyzed muscles, higher activation was found for the more proximal muscles exposed to the vibration. The

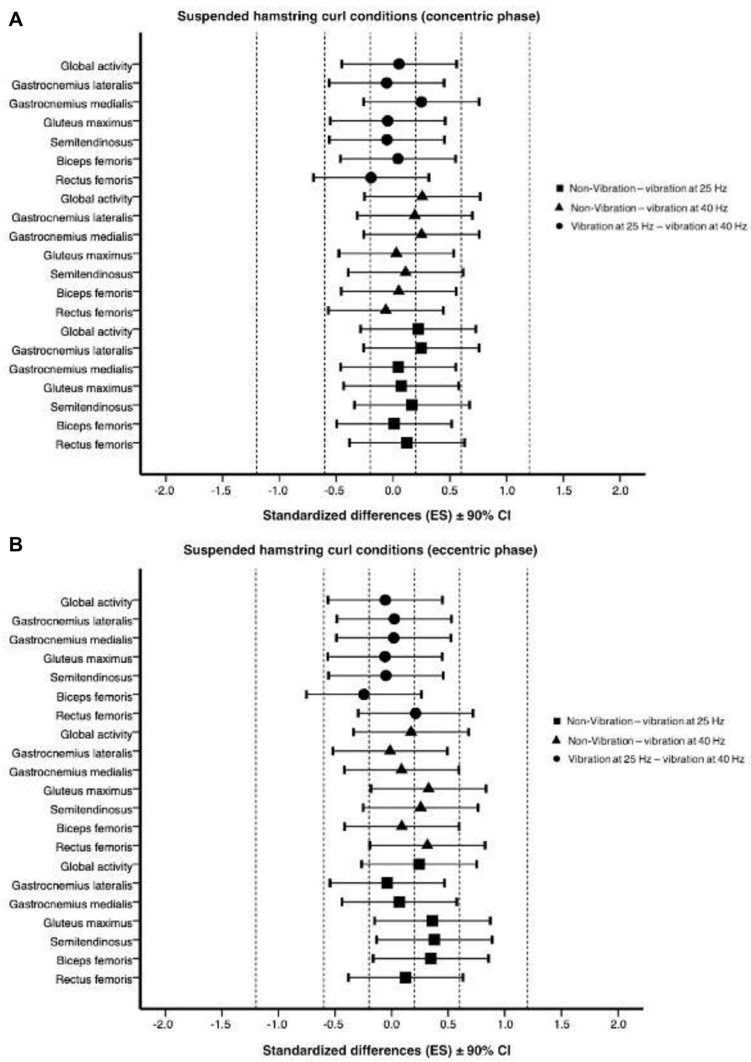

FIGURE 4 | Effects of suspended hamstring curl conditions on muscle activity (%MVIC) at concentric **(A)** and eccentric phase **(B)** expressed as standardized differences (Cohen's *d*) ± 90% CI. Dotted line represents the effect size thresholds. ES, effect size; CI, confidence interval.

additional effect of vibration at 25 Hz compared to the non-vibration suspended condition was significantly higher for the gastrocnemius (medialis and lateralis) and semitendinosus in the concentric and eccentric phase (from 9.8% to 23.8% with trivial to moderate effect). Previous studies also demonstrated that the more proximal to the vibration experimented higher activities than the more distal muscles (Hazell et al., 2010; Ritzmann et al., 2013). In this regard, the present study showed that in both vibration conditions (25 Hz and 40 Hz), the muscle excitation sequence (Neto et al., 2019), from higher to lower activation, was gastrocnemius lateralis, gastrocnemius medialis, semitendinosus, biceps femoris, gluteus maximus, and rectus femoris (**Table 1**). Thus, the magnitude of the neuromuscular response to the vibratory stimulus in those muscles that are closer to the most proximal joints (ankles) dissipates the effects of vibration for the more distal muscles, acting as a damper (Abercromby et al., 2007b). Indeed, the vibration induces different reflexes that favor increased muscle activation on the most proximal muscles, such as the tonic vibration reflex (Issurin, 2005; Ritzmann et al., 2010) or the stretch reflex on the soft tissues (Cardinale and Lim, 2003; Cochrane et al., 2009).

Of all analyzed muscles, gastrocnemius lateralis (41–60% MVIC) achieved a high activation under 25 Hz vibration and slightly lower (37.4% MVIC) for gastrocnemius medialis. Participants were asked to perform an ankle plantar flexion on the strap cradles instead of leaning their heels on the suspension cradles in the suspended supine bridge. Ritzmann et al. (2013) found that the variation of the foot position on the vibration platform increased the gastrocnemius medialis activity up to 48% (forefoot stance vs. normal stance). Although the feet remained in plantar flexion in the three suspended supine bridge conditions in the current study, the percentage of gastrocnemius activity significantly increased (14–23%, from small to moderate increase) under 25 Hz vibration to the non-vibration condition.

The lack of differences between the 40 Hz vibration and the non-vibration suspended condition could be explained because gastrocnemius is more predominantly activated at frequencies below 40 Hz (20, 25, and 30 Hz) (Di Giminiani et al., 2013), according to the findings of the present study (**Table 1**).

The hamstrings (biceps femoris and semitendinosus) muscle activity ranged from moderate to low (< 24% MVIC), with significant differences in semitendinosus activity at 25 Hz and 40 Hz in comparison to the non-vibration condition. However, following Abercromby et al. (2007a), the biceps femoris activity was slightly lower, with similar activation in all conditions. This low activation (< 21% MVIC) of the biceps femoris is related to 90° knee flexion in the suspended supine bridge. Ho et al. (2020) found a similar low activation (18% MVIC) of the biceps femoris in the dynamic supine bridge (90° knee flexion). However, the effect of WBV in the static supine bridge, maintaining the 90° of knee flexion, elicited a significant moderate activation (21–40% MVIC) of the biceps femoris at 30 Hz and 50 Hz, although the non-vibration condition also showed a moderate level of activation (27% MVIC). The authors supported that 50 Hz vibration was more demanding for the biceps femoris in the static supine bridge (Marín and Cochrane, 2021). Similarly, Hazell et al. (2007) found an increase in biceps femoris activation between 35 Hz and 45 Hz for dynamic and static squats. This suggested that superimposed vibration (25 Hz and 40 Hz) in the dynamic suspended supine bridge is insufficient to significantly stimulate the biceps femoris compared to the non-vibration condition significantly. Thus, an increased frequency of superimposed vibration on the suspension straps (> 40 Hz) and performing the exercise unilaterally, single-leg suspended supine bridge, could increase the demand of the biceps femoris to high activations (> 41% MVIC), as indicated by previous studies on sEMG on the single-leg supine bridge on the floor (Lehecka et al., 2017), or on a BOSU (Youdas et al., 2015). In this vein, the functional magnetic resonance imaging study conducted by Bourne et al. (2018b) found a predominant activation of the biceps femoris long head. Likewise, there could be several reasons for the small differences between the biceps femoris and semitendinosus in the suspended supine bridge. One reason is that the suspended exercise produces lateral instability, provoking a lateral rotation of the thighs and, consequently, an increased semitendinosus activity because of its role in counteracting this movement (Tobey and Mike, 2018). Furthermore, the amplitude of the vibrating machine (8mm, peak to peak) is suggested to provoke more horizontal oscillations and focus on the stabilizing structures that, in the present study, are stabilized by the semitendinosus (Cook et al., 2011). Another reason is that the necessity to keep the feet stable and maintain the anchor in a plumb line (perpendicular to the ground) of the suspension strap requires the participation of the posterior thigh muscles, similar to the feet-away hip thrust (Collazo García et al., 2020). This semi-stretched position provokes an increase in muscle tension and enhances the effects of the vibration in the hamstrings muscles (Cardinale and Lim, 2003; Marín and Cochrane, 2021). Overall, as a practical application, muscles with activations below 45% MVIC, such as biceps femoris and semitendinosus in suspended supine bridge conditions (non-vibration, 25 Hz and 40 Hz vibrations), would be targeted for muscular endurance, stabilization, and rehabilitation training programs (Ekstrom et al., 2007; Youdas et al., 2015).

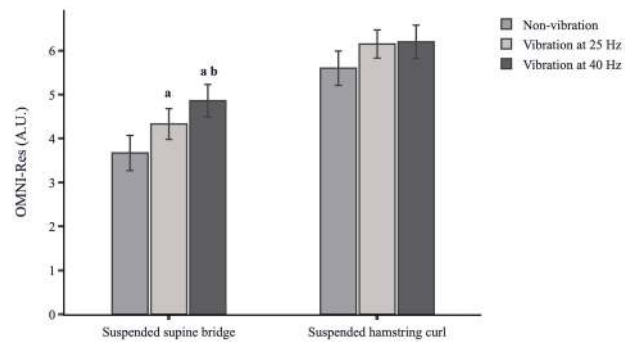

FIGURE 5 | OMNI-Res (mean ± SE) for suspended supine bridge and suspended hamstring curl under non-vibration, vibration at 25 Hz and vibration at 40 Hz conditions. Each bar represents the mean, and the error bar represent the standard error of the mean (SE). A.U., Arbitrary units; [a]significantly different with non-vibration condition; [b]significantly different with vibration at 25 Hz condition.

Although the barbell hip thrust is a very demanding exercise for gluteus maximus (> 60% MVIC) (Neto et al., 2019), the variation of suspended (and unloaded) exercise proposed in this study elicited low activation (< 23% MVIC) with a trivial and small effect among conditions (**Figure 3**). In this vein, previous studies have reported activation levels ranging from moderate to low (< 25% MVIC) for gluteus maximus in unloaded supine bridge on the floor (Ekstrom et al., 2007; Jang et al., 2013; Kim and Park, 2016). Thus, it appears that the suspended supine bridge (with an additional effect of vibration) is as demanding for the gluteus maximus as the traditional supine bridge exercise and are not sufficiently challenged to reach high and very high activation values (> 40% MVIC) in the gluteus maximus, as happens with the single-leg bridge (Ekstrom et al., 2007; Lehecka et al., 2017), the WBV supine bridge (Marín and Cochrane, 2021) or the barbell hip thrust (Contreras et al., 2016; Andersen et al., 2018; Williams et al., 2021). Therefore, although the gluteus maximus is the prime supine bridge mover, its activation is still low. Moreover, superimposed vibrations were dampened by the more proximal to vibration musculature, and the gluteus maximus were not overstimulated. In addition, the rectus femoris showed the lowest activation (< 2.0% MVIC) with a trivial effect in both phases of exercise without significant differences among conditions. Collazo García et al. (2020) showed a significantly (2.4%) lower rectus femoris activation in the feet-away barbell hip thrust (3.4% MVIC) compared to the original hip thrust condition (5.8% MVIC). Likewise, Lehecka et al. (2017) found similar rectus femoris activity in the unloaded single-leg bridge with 90° of knee flexion, agreeing with the present study results.

Conversely, as hypothesized, the additional effect of the superimposed vibration did not result in a significantly higher activation in any of the analyzed muscles, or the global activity, during the concentric and eccentric phases of the suspended hamstring curl (**Table 2**). Moreover, differences among exercise conditions ranged from trivial to small (**Figure 4**). Even though the muscle excitation sequence was similar to the suspended supine bridge. Thus, the activation increments of the most proximal muscles to the vibratory stimulus (gastrocnemius medialis and lateralis) were between 9% and 5% ranged from trivial to small increase in 25 Hz and 40 Hz vibration, respectively, to the non-vibration condition. The main difference in transmitting the vibration between the suspended supine bridge and the suspended hamstring curl was the suspension strap position. The straps remained in a plumb line in the supine bridge, whereas it acted as a pendulum in the suspended hamstring curl. Several studies suggested that vibration transmission via cable in pulley exercises such as biceps curl or one arm pulleying keep the perpendicular between the anchor point, vibration device, and handle to enhance the effects of local vibration (Bosco et al., 1999; Issurin and Tenenbaum, 1999; Issurin et al., 2012). Nevertheless, the pendulum motion in the suspended hamstring curl could attenuate vibration transmission because the vibratory system is designed to transmit the vibration. Moreover, it could be speculated that the pendulum motion could also exert a dampening effect by inhibiting the tonic vibratory reflex (Rittweger, 2010). On the other hand, the pendulum motion and plantar flexion to keep the feet on the cradles could explain the gastrocnemius activity in the suspended hamstring curl conditions. Additionally, Bettendorf (2010) suggested that the intensity variation in a suspended exercise is based on three fundamental principles. Thus, the pendulum principle could justify that the prime mover activations (biceps femoris and semitendinosus) in this study were slightly higher than low activations ($<$ 21% MVIC) reported by Árnason et al. (2014) in the suspended hamstring curl without pendulum movement and lower than high and very high activations ($>$ 50% MVIC) registered by Malliaropoulos et al. (2015) in the suspended hamstring curl with alternating knee flexion and pendulum motion.

Regarding OMNI-Res, the finding was that superimposed vibration increased the value of subjective perception of exertion compared to the non-vibration suspended condition around 10% (small increase) for both vibration frequencies in the suspended hamstring curl and from 18% to 32% (small to moderate increase) for the suspended supine bridge. Thus, it seems that the value of OMNI-Res increases progressively while increasing the vibration frequency, being consistent with the significant correlation ($r = 0.95$) between OMNI-Res and a range of vibration frequency (25 Hz to 45 Hz) and amplitudes (1 and 3 mm) found by Marín et al. (2011). Additionally, the validity and reliability of the intensity of exertion using subjective scales in exercises with superimposed vibration have been demonstrated for both vibration frequency and muscle activation (Marín et al., 2012).

There were some limitations in the study. The effect of superimposed vibration on suspended exercises has been assessed in physically active men and women, so the results obtained in the present study cannot be generalized to other populations. The footwear soles were different among participants, and since this area is the most exposed to vibration, this could slightly modify the vibratory stimulus due to the damping effect of the footwear soles. Therefore, future research should standardize the footwear for all participants. Likewise, the vibration transmitted through the suspension strap could have dissipated the vibration effect. While the distance between the suspension strap and the ground was standardized, it could be interesting to examine different suspension strap heights and their effects on muscle demand in the supine bridge in future studies. Another limitation was that the erector spinae and vastus (medialis and lateralis) requested in the supine bridge were not evaluated because the electromyography system employed only offers six channels. Further investigations could study the effects of superimposed vibration on neuromuscular performance in a loaded suspended supine bridge (kettlebell, barbells, weight plates) or variations of the exercise such as a single-leg or modifying the arm positions (crossed over the chest).

CONCLUSION

The additional effect of the superimposed vibration resulted in being more challenging for the suspended supine bridge than the suspended hamstring curl. Although the two vibration frequencies elicit the same activation level at the global activity level, the suspended supine bridge with a 25 Hz vibration provoked a higher activity of the most proximal muscles to the vibration device (gastrocnemius medialis, lateralis, and semitendinosus), with meaningless effects on the primary movers. Therefore, the amount of instability provoked by the suspended supine bridge with superimposed vibration increased the stabilizing role of the gastrocnemius and semitendinosus. In contrast, the anteroposterior movement of the suspended hamstring exercise seems to be less effective in transmitting the vibration. Regardless of the exercise, increasing the vibration frequency on the suspension device leads to a higher value of subjective perception of exertion (OMNI-Res).

Practical Application

The suspended supine bridge is as demanding as a traditional exercise for the gluteus maximus. However, the additional effect of the superimposed vibration in the suspended supine bridge provides greater gastrocnemius and hamstrings activity. Plantar flexion in the suspended supine bridge with superimposed vibration is a successful manner for strengthening the gastrocnemius, demanded in sports actions such as changes of direction, jumps, and sprints. Furthermore, this method allows dynamic tasks, changing the planes of the force production and offering a continuous exposition to vibration for the working muscles. Likewise, the increased instability generated through vibration to the suspension straps turns the suspended supine bridge into an exercise that demands the neutralization of the lateral rotation of the thighs, similar to other lateral actions in several sports actions. Moreover, superimposed vibration in a suspension device can complement traditional exercises such as the Nordic hamstring, leg curl, or deadlift to develop the strength and endurance of the hamstrings in strength and conditioning programs. Additionally, injury prevention and rehabilitation can benefit from the outputs of the present study to further evaluate the inclusion of superimposed vibration in the prescribed protocols since hamstrings injuries are prevalent in many sports.

DATA AVAILABILITY STATEMENT

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found below: Figshare (https://doi.org/10.6084/m9.figshare.14537001).

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics and Research Committee Board in the Blanquerna Faculty of Psychology and Educational and Sport Sciences at Ramon Llull University in Barcelona, Spain, with reference number 1819005D. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

JA-C, BB, AF-V, and JP contributed to the conception and design of the study. JA-C, JA-A, AF-V, and AM contributed to the acquisition of data. JA-C, JA-A, and AM analyzed the data. JA-C and BB wrote the original draft of the manuscript. BB and JP supervised the study. All authors wrote, edited, reviewed, and approved the submitted final version of the manuscript.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2021.712471/full#supplementary-material

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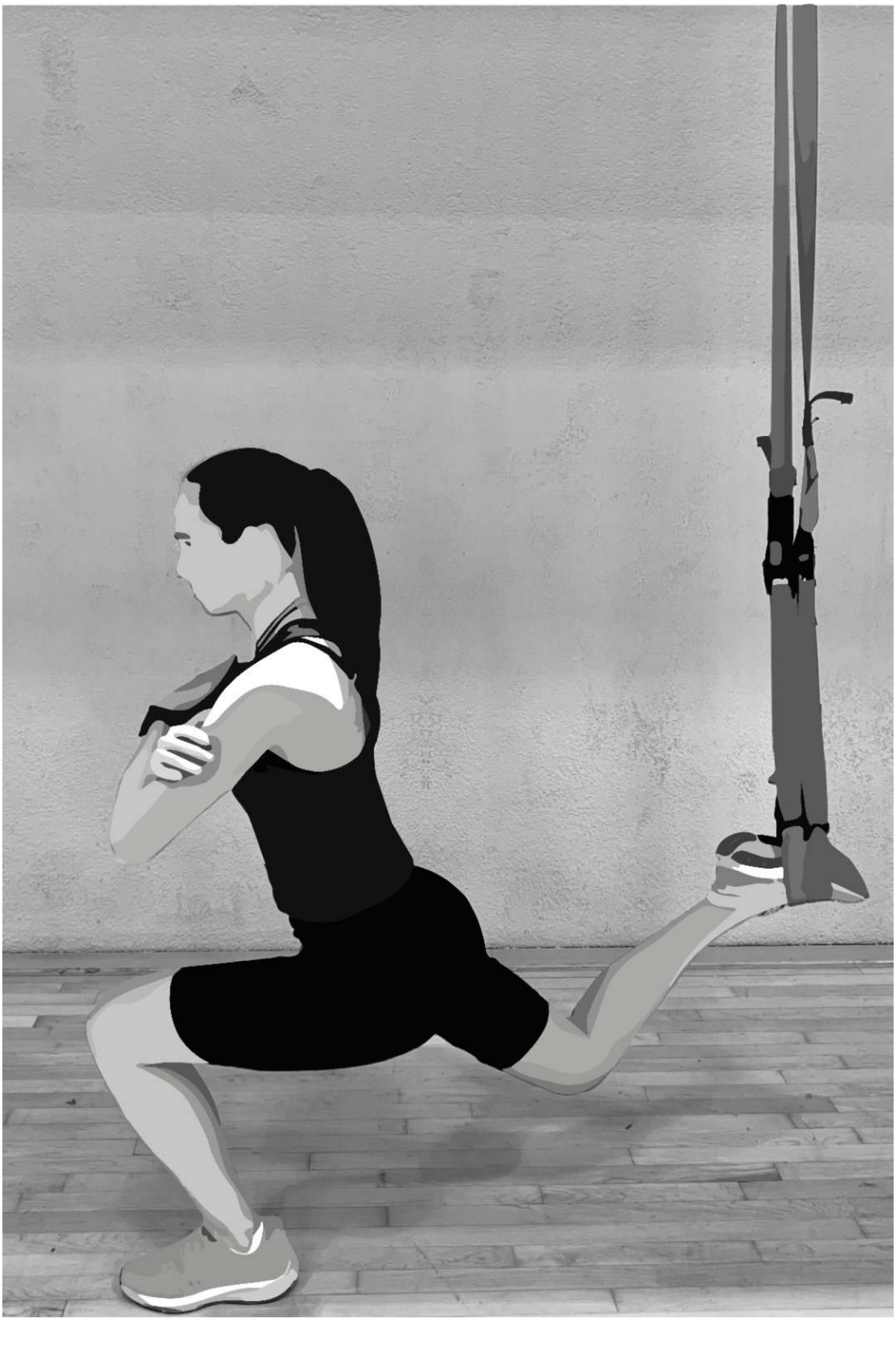
Conflict of Interest: JA-C and BB are affiliated to the Faculty of Psychology, Education Sciences, and Sport Blanquerna, of the Ramon Llull University, and JP is affiliated to the University of Vic - Central University of Catalonia, which have requested for a patent (number 202030652, Spanish Patent and Trademark Office - OEPM) enabling the use of superimposed vibration in athletic, fitness, and health settings.

The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.



PATENT

For confidentiality reasons the content of the patent section is not shown from page 313 to 329.