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UNDERSTANDING THE FACTORS INFLUENCING FUNCTIONAL TASKS FOR THE ASSESSMENT OF MUSCULOSKELETAL INJURY-RELATED BIOMECHANICS

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ABSTRACT

Background: The evaluation of lower limb biomechanics during functional tasks is extremely common and relevant in the current sports and exercise medicine literature. The choice of which task to use, which variation to adopt and which metric to extract is dependent on the research question and on the population investigated. However, each of those factors (in combination with many others) may affect the findings obtained by functional task, which, in turn, may lead to inadequate planning of rehabilitation and injury prevention protocols in clinical and sports practice.

Aims: To evaluate how distinct factors influence biomechanical assessment during functional tasks using different tasks, metrics, and populations.

Thesis structure: Four experimental studies were conducted in order to address five specific aims that will contribute to the current knowledge on the use of functional tasks.

Study 1: This study compared kinematics when executing three squat-based tasks (single-leg squat, anterior step-down, and lateral step-down) at three different speeds (slow, fast and self-selected. The study found that both task type and movement speed can influence several metrics commonly used to assess movement kinematics, albeit with small absolute difference in degrees.

Study 2: This study investigated if there were relationships between two metrics of muscle activation of lower limb and core muscles during single-leg squats and anterior step-downs. The findings show that, although present, this relationship is muscle, metric and task dependent.

Study 3: This study compared the differences between people with different running experience levels on linear and angular stiffness during running gait, finding that these metrics were not different between groups, suggesting that the increased injury rate in less-experienced runners is likely not explained by different gait patterns.

Study 4: This study measured the test-retest reliability of force measurements during the execution of functional tasks with progressive difficulty in a healthy and pathological population. The results show that these metrics is dependent on the task, the metric and the participants' injury status and that several metrics were not sufficiently reliable.

Conclusion: The thesis findings support the idea that results are highly dependent on many components that need to be taken into account when using functional tasks for the evaluation of healthy and clinical populations. Furthermore, it supports the necessity for the sports and exercise medicine literature to provide as many details as possible when describing these tasks and that care should be taken when comparing research findings with other studies that might have used different task variations, metrics and populations.

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THESIS INTRODUCTION

1. WHAT ARE FUNCTIONAL TASKS AND WHY DO THEY MATTER?

The evaluation of lower limb biomechanics during functional tasks is extremely common and relevant in the current sports and exercise medicine literature. Because these movement evaluations often rely on sophisticated and often bulky equipment, it is difficult to assess people while they are naturally executing the movements of interest and functional tasks are the most common method of replicating them. The first mentions of functional task in the context of sports medicine is found in the 1980s to describe tasks such as walking and stepping (Barrack et al. 1983; Freedman and Kent 1987; Wolf and Minkwitz 1989) and they have since included task such as jumping (Nakagawa et al. 2011; Guimaraes et al. 2023) and squatting (Glaviano et al. 2016; Mirzaie et al. 2019). However, there is currently no universally accepted definition for the term. Guimaraes Araujo et al., (2023) described them as being “able to simulate activities of everyday life and sports gestures”. This definition, however, does not specify what is being simulated and why they are being performed. Based on studies that use them, we can conclude that the load to the musculoskeletal system is what is being simulated and the reason they are being performed is to be able to measure specific parameters, such as performance or biomechanics. Therefore, we can tentatively expand this definition to state “Functional tasks are specific movements performed by an individual that simulate the load experienced during activities of everyday life and sports gestures in order to evaluate performance and biomechanics”.

Although we can classify several tasks as “functional”, many studies use other terms (e.g., screening tasks or dynamic tasks) (Lepley et al. 2013; Husted et al. 2016; Smale et al. 2019) or choose to just name the type of task executed or even the specific characteristics of the movement executed (Hinman et al. 2002; Mattacola et al. 2002). Therefore, it is important to understand what they are and how authors may choose to name them.

Squat-based tasks: These tasks involve a controlled lowering of the center of mass by using a combination of hip and knee flexion. The most common examples are double-leg squat, single-leg squat, anterior step-down, and lateral step-down (Figure 1).

Figure 1. Example of participant executing the double-leg squat (A), single-leg squat (B), anterior step-down (C), and lateral step-down (D).

Jump-landing tasks: These tasks typically involve a vertical jump and/or a landing from different heights. They can be done either double or single-legged and are likely the most popular type of functional task. Some examples are the land-and-hold, the countermovement jump and the drop jump (Figure 2).

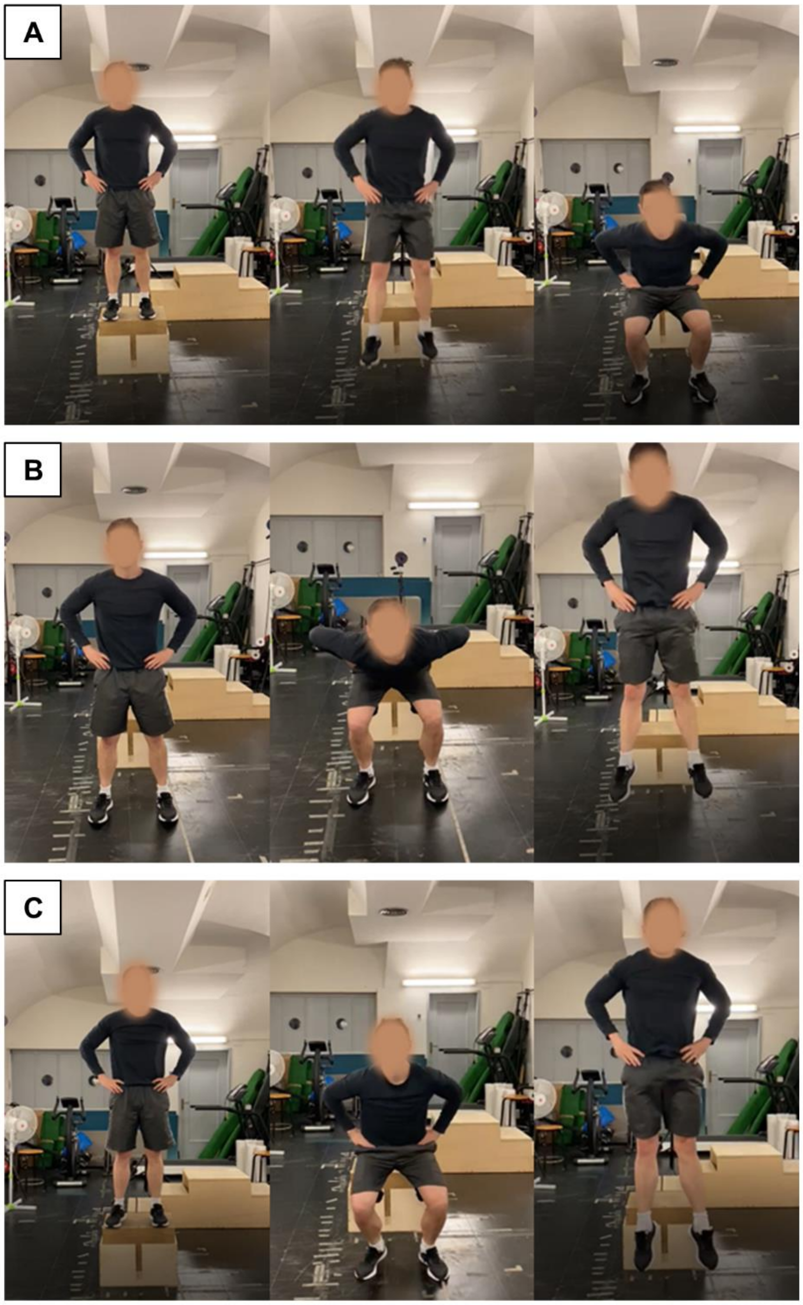

Figure 2. Example of participant executing the land-and-hold (A), countermovement jump (B) and the drop jump (C).

Hop tasks: These tasks require the participant to jump while moving forwards or laterally and land in a balanced way. The most used examples are the single-leg hop, the triple hop, and the crossover hop (Figure 3).

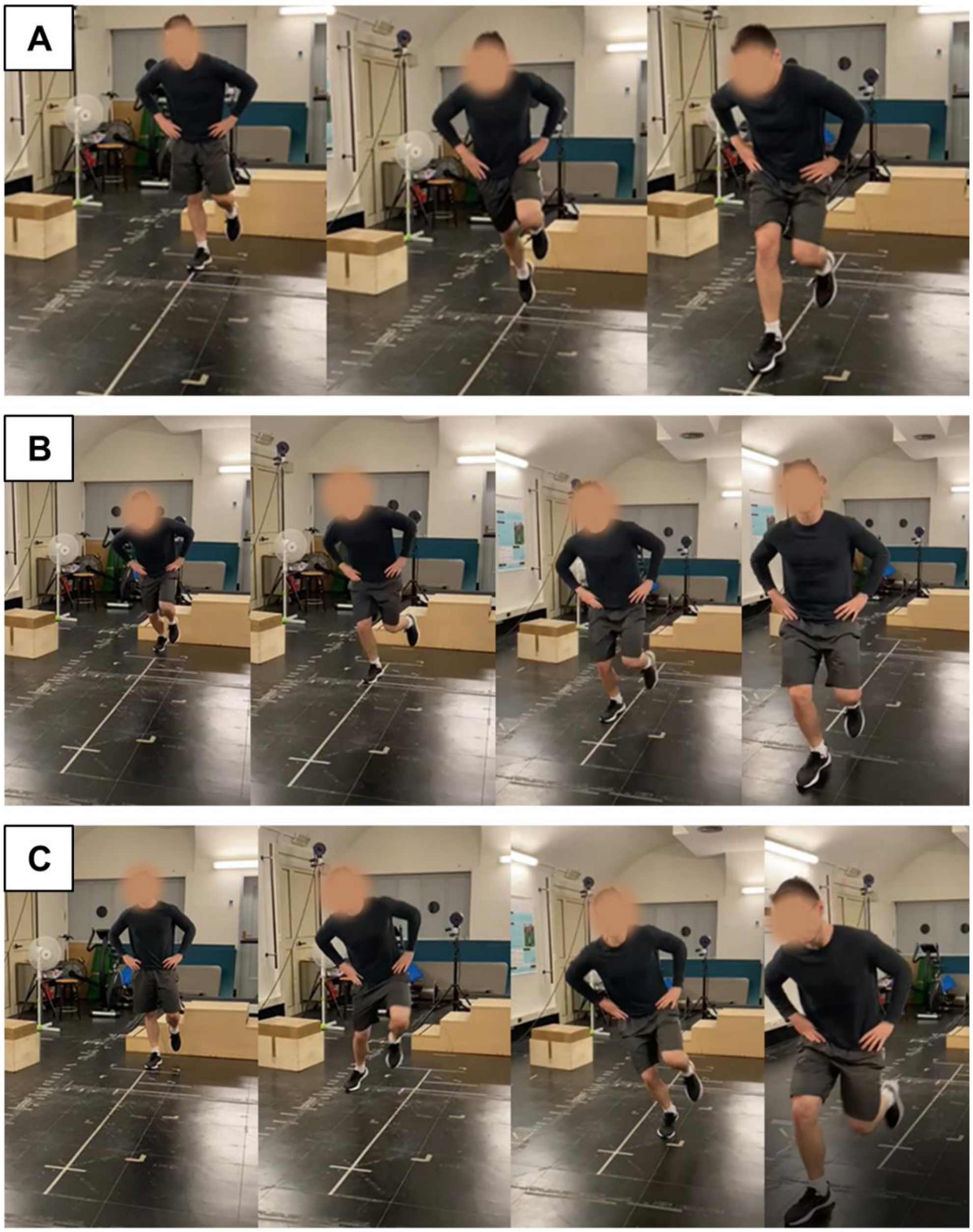

Figure 3. Example of participant executing the single-leg hop (A), triple hop (B) and crossover hop (C).

Cutting-based tasks: In these tasks a participant already running is asked to perform a sudden change of direction that can be in different directions and different angles. It requires deceleration followed by a quick acceleration towards the new direction. Common examples are the 90° cut, the 45° cut and the 180° turn (Figure 4).

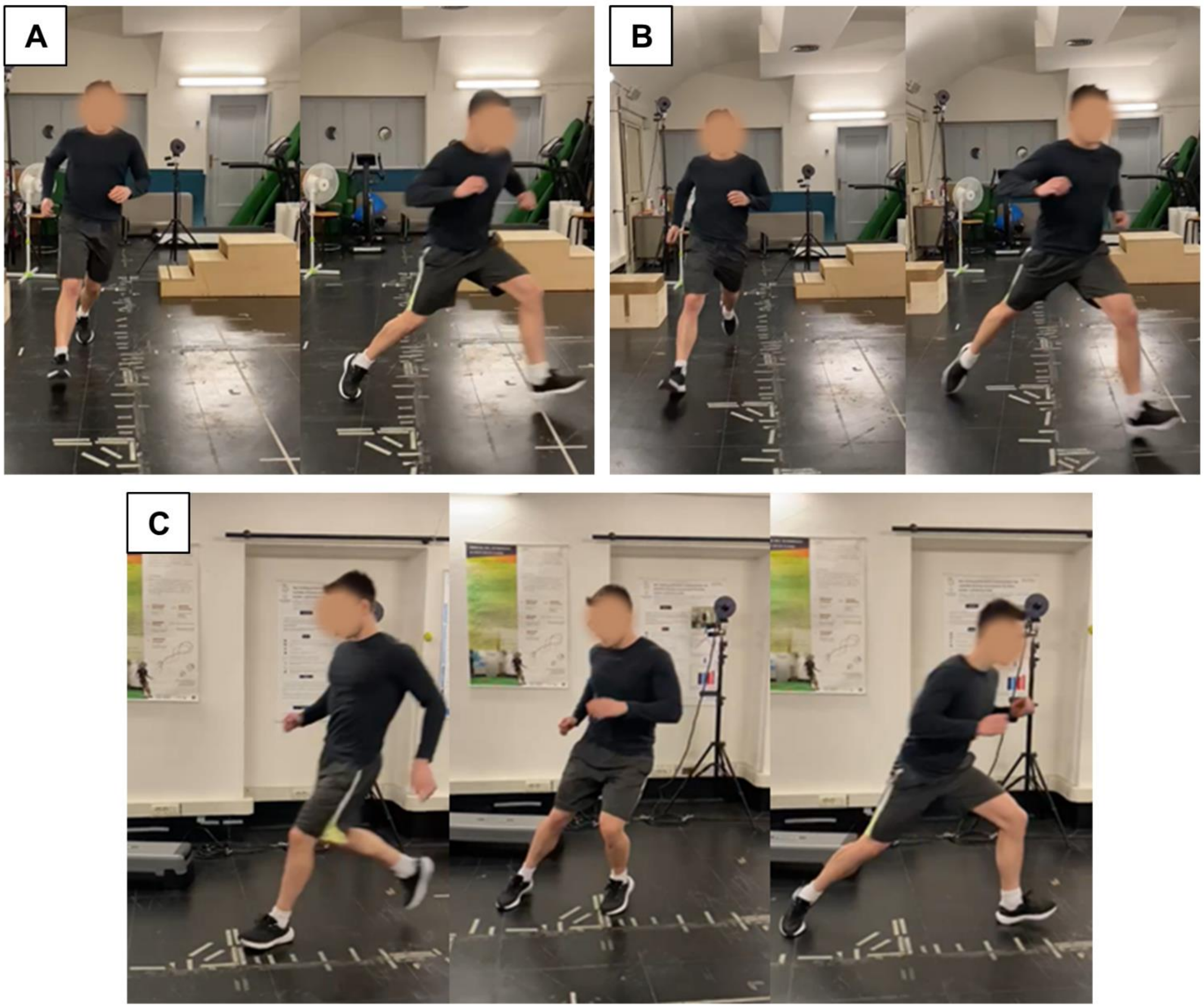

Figure 4. Example of participant executing the 45° cut (A), 90° cut (B) and 180° turn (C).

Static balance tasks: In these tasks the goal is to stay as still as possible in order to measure balance. The most common examples are the eyes-closed double-leg, the single-leg balance and the tandem balance tasks (Figure 5).

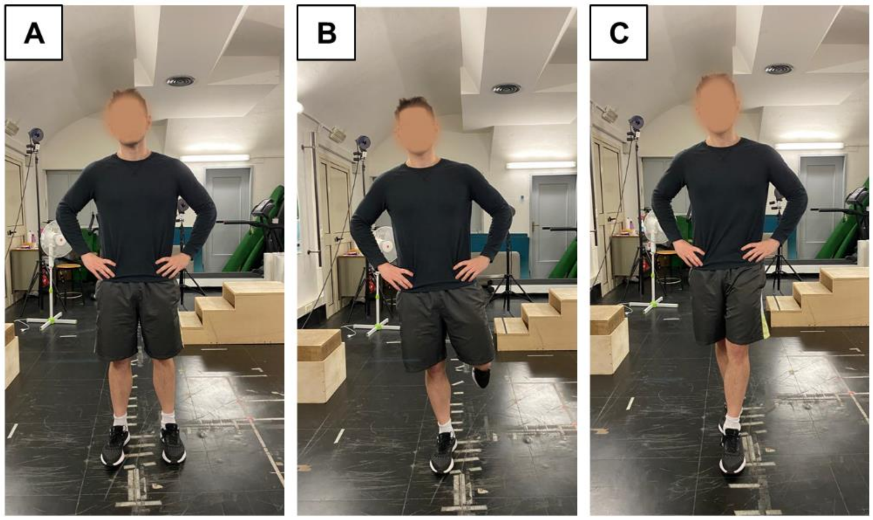

Figure 5. Example of participant executing the double-leg stand (A), single-leg stand (B) and tandem balance (C).

Gait-based tasks: There is some debate about whether these tasks should be included with the previous examples. While the other types of tasks seek to simulate the loads and challenges of different sport or daily activity gestures, gait-based tasks are effectively the gesture itself. However, given that in research they are often conducted on treadmills or small spaces, which are not how people usually perform them, we can consider them to also be a simulation of the gesture and, thus, functional tasks. The main examples of these tasks are walking, running, ascending stairs and descending stairs (Figure 6).

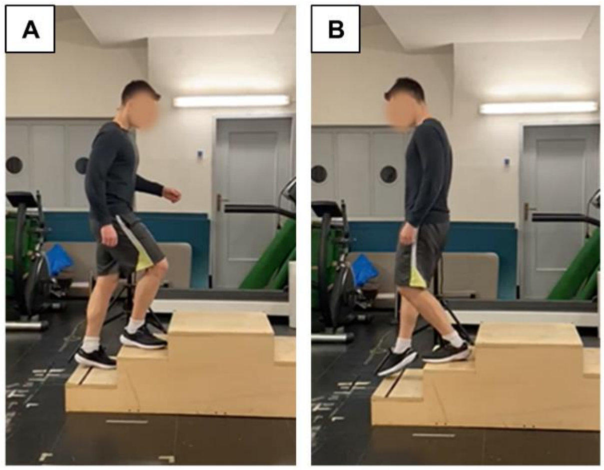

Figure 6. Example of participant ascending stairs (A) and descending stairs (B).

Combination tasks: These tasks are a combination of two or more of the previous task types. Examples are the single-leg land and cut and the stop jump (Figure 7).

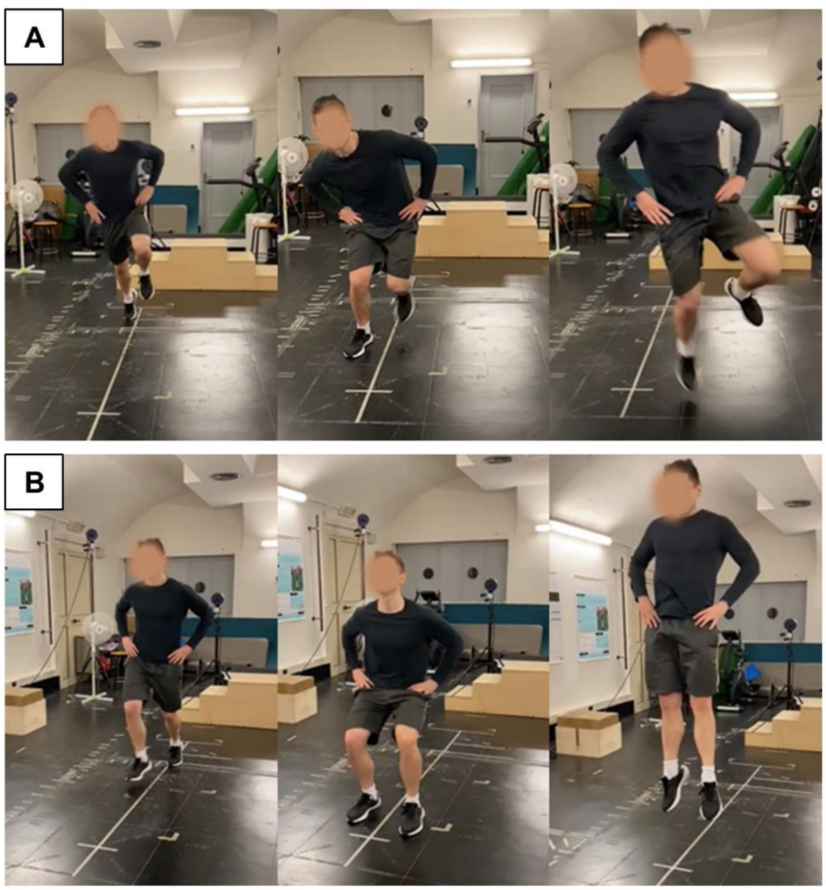

Figure 7. Example of participant executing the single-leg land and cut (A) and the single-leg hop and jump (B).

The choice of which task to use is dependent on the research question and on the population investigated. Studies might wish to only use “safer” tasks such as gait and squats when evaluating populations that are frailer, such as older adults and persons that have recently gone through surgery (Ko et al. 2011; Trulsson et al. 2015). On the other hand, tasks that are more challenging, such as the single-leg landing and the drop-jump, are more often used with athletic populations (Chinnasee et al. 2018; Everard et al. 2021). In addition, the choice can depend on which tasks best represent the most common movements performed by a given population. For example, runners are usually evaluated during gait (Desai and Gruber 2021) while athletes that often jump in their sport (e.g., basketball or volleyball) are usually evaluated in jump-landing tasks (Kulig et al. 2015; Harris et al. 2020). Finally, within a given task there can be modifications to make it even more activity-specific. For example, ballet dancers can be tested while wearing their ballet shoes (Pavlović et al. 2022) and perform jumps in the turned-out position (Mattiussi et al. 2023) and volleyball players can be asked to spike a ball during a jump-landing movement (Kulig et al. 2015).

As mentioned, these tasks are usually employed to assess either performance or biomechanics. While performance studies are important and make up a substantial part of the current literature, this thesis will focus on the biomechanical aspects of functional tasks. Biomechanics-focused studies typically evaluate kinematics, kinetics, and muscle activation, with a countless number of specific metrics that will be discussed in a later section. The research designs can be varied and can use the tasks to compare two distinct groups (e.g., compare healthy participants with those with patellofemoral pain (O’Sullivan et al. 2012; Nakagawa et al. 2013), compare different limbs (e.g., the injured with the uninjured side (King et al. 2018; Hetsroni et al. 2020), before and after an intervention protocol (e.g., neuromuscular electrical stimulation for the treatment of patellofemoral pain (Glaviano et al. 2020), before and after modifying a participant condition (e.g., with or without foot orthoses in people with osteoarthritis (Tan et al. 2020) or identify people that are at a greater risk of getting injured in the future (Hewett et al. 2005; Räisänen et al. 2018).

Currently, functional tasks are by far the prevailing method of evaluating movement biomechanics. Although there is movement towards evaluating athletes on the field and other people during their regular life (Rawashdeh et al. 2016; Horenstein et al. 2020; Abdollah et al. 2021), the majority of the literature and, therefore, the current knowledge stems from conclusions made from studies that used these functional tasks (Hewett et al. 2005; Nakagawa et al. 2012). On one hand, the execution of these tasks in a laboratory environment using functional tasks isolates the movement of interest and the assessment, presenting as an advantage due to not requiring complex data and statistical analysis to account for the variability present in the “real world”. On the other hand, the execution of tasks during more ecological contexts (i.e., during the actual sport or activity of interest), might help with the validity of the measurement. The dissonance of these two approaches is something that needs to be taken into account by all researchers while they seek to reach the ultimate goal of research, which is to apply the knowledge obtained in real life situations (e.g., in rehabilitation or injury prevention protocols). However, for a successful application of findings observed with the use of functional tasks, it is fundamental to fully understand the aspects of the tasks that can influence the results. In the next sections, I will discuss which factors can be modified during functional tasks, how they affect the movement biomechanics and how the choice of metric can alter the interpretation of studies that use functional tasks.

Influence of biomechanics on musculoskeletal injuries

As previously mentioned, there are several studies that conduct biomechanical evaluations using functional tasks in order to investigate their role in the occurrence of many musculoskeletal injuries (Ko et al. 2011; Nakagawa et al. 2012; O'Sullivan et al. 2012; Trulsson et al. 2015; King et al. 2018; Hetsroni et al. 2020). The conditions that are most commonly investigated in this context are anterior cruciate ligament tear, patellofemoral pain, ankle and foot injuries and running-related injuries. Examples of the current knowledge regarding the relationship between these conditions and biomechanics will be discussed below.

Because anterior cruciate ligament (ACL) tears are the most common traumatic knee injuries (Daniel et al. 1994) and result in serious consequences for the patient (Glogovac et al. 2019), it is likely the most investigated injury within the injury biomechanics literature (Von Porat et al. 2007; Husted et al. 2016; King et al. 2018; Gilmer et al. 2020). Its most common mechanism of non-contact associated with dynamic knee valgus has led to many studies seeking to identify which biomechanical factors may be associated with this injury, with the aim of modifying them in order to prevent future or recurrent injuries (Hewett et al. 2005; McLean et al. 2005; Richardson et al. 2020). There are also several other factors that have been identified as associated with ACL injuries, such as weak hip abductor strength, poor hip musculature control, increased femoral anteversion, wider pelvis, midfoot mobility, greater q-angle, poor postural control and more upright landings (Larwa et al. 2021).

Because of the large number of prospective, retrospective and observational studies, it is difficult to reach definitive conclusions about the role of biomechanics on the occurrence of this injury. Nonetheless, systematic reviews have tried to organize this knowledge and arrived at interesting conclusions. Larwa et al. (2021) evaluated nine video analysis studies and nine pre-screening studies, finding that ACL injuries were associated with stiff landings, increased knee valgus and landing on a heel strike, being particularly prevalent in female athletes. Focusing on a more specific aspect of motor control, Bertozzi et al. (Bertozzi et al. 2023) found that worse cognitive performance also influenced the ACL injury risk profile during movements that also required a cognitive challenge, which are more closely related to the reality of sports participation. Meanwhile, Petushek et al. (2019) found that neuromuscular training which included landing stabilization and lower body strengthening exercises were able to reduce the risk of ACL injury risk. Overall, there are many other studies that deal with ACL injury risk that were not included in a systematic review due to being overly specific (i.e., focusing on a specific population, using an uncommon metric or adopting a different study design). Therefore, a clear picture of how biomechanics affect ACL injuries is still lacking.

Another, less traumatic, but nonetheless common condition affecting the knee joint is patellofemoral pain (PFP). This condition affects women at a greater proportion (Boling et al. 2010) and describes a condition of pain in and around the patella. Although the reason for this condition may be multifactorial (Thomeé et al. 1999; Powers et al. 2012), several studies have specifically investigated the influence of poor biomechanics on the development of PFP (Lankhorst et al. 2013). In a systematic review including 47 studies, Lankhorst et al. (2013) found that several kinematics, kinetics and muscle activation metrics were different in people with PFP. However, most differences were based on single studies, making it difficult to reach a definite conclusion. The fact that there were 523 variables among the included studies and that diagnostic criteria was not the same contributed for this heterogeneity.

Besides the knee, the foot and ankle joints are likely the most common site of injuries during sports and physical activity. As the distal part of the kinetic chain, these joints play an important role in dealing with ground reaction forces and, therefore, may also suffer injuries. Examples of these injuries are hallux valgus, metatarsal injury and general ankle pain (Bowling 1989; Furia et al. 2010; Werber 2011). Although there are other risk factors for these types of injuries, the loading on these joints is a particular focus of the biomechanical studies, which are also affected by footwear. In another systematic review, Li et al. (2022) found that the condition of the ballet shoe was the most important factor for foot injuries in ballet dancers. In the same systematic review, they also found that the foot biomechanics is often investigated in the literature in the context of injuries (Prochazkova et al. 2014; McPherson et al. 2019) and the strengthening of the foot muscles should reduce injuries.

Because of the popularity of running as a physical activity and sports, but also its high injury incidence (van Gent et al. 2007), running related injuries are likely the most common topic of research within biomechanics (Bramah et al. 2018; Damsted et al. 2019; Schmida et al. 2022). The current accepted reason for the occurrence of running related injuries is that there is an accumulation of repetitive stress to the musculoskeletal structures, resulting in degeneration of the tissues without enough time to recover (Willwacher 2017). Although people have tissues and structures that present different characteristics of stress resistance, the way this stress is applied also plays an important role on whether someone will get injured or not (Edwards 2018). For this reason, there are many studies that evaluate the gait of people during running in several conditions and try to associate it with the incidence of injuries. Theoretically, identifying if there are biomechanical gait characteristics that are more injurious will allow runners and coaches to change them and prevent these running related injuries. Willwacher et al. (2022) conducted an important systematic review on the biomechanical risk factors for running injuries. This study included 66 articles and classified the level of evidence into groups that ranged from strong to no evidence. This review also presented the results separately for the different types of injury that may occur, as one biomechanical gait characteristic may decrease the stress in one structure while increasing it for another, ultimately not reducing the injury, but rather just shifting locations. Although there were 123 possible risk factors identified, none were identified as presenting strong or moderate evidence for Achilles tendinopathy. Similarly, iliotibial band syndrome and tibial stress fracture also presented only limited evidence, despite the investigation of 93 and 41 possible risk factors. Differently, the review found that increased duration of rearfoot eversion angle and contralateral pelvic drop angle had moderate evidence of being a risk factor for medial tibial stress syndrome, factors that were among 34 possible ones. Plantar fasciitis was another condition for which the authors were able to identify a risk factor with moderate evidence (among 46 possibilities). The authors found that increased average and instantaneous loading rate of the vertical ground reaction force was injurious. Finally, patellofemoral pain in runners was also investigated, finding that decreased braking ground reaction force impulse and increased contact time were associated with this injury (among 120 possibilities).

In this section, I discussed some of the most commonly investigated conditions within the injury biomechanics literature. I used examples from systematic reviews as they are the closest we can get to a comprehensive knowledge, as it is extremely difficult to reach a conclusion when considering the different methodologies, results and caveats of a number of studies concomitantly. Specifically to studies that use functional tasks, each one may adopt a different task (or the same task with variations), investigate different populations, measure different metrics and use different statistical analysis, all while seeking to answer the same question. Biomechanics may be a very important factor that leads to musculoskeletal injuries during sports and athletic activities. The forces that act on the different tissues and body structures can be measured by biomechanical analyses, but while each study adopts different methodologies, it is difficult to make an informed conclusion. Nonetheless, these studies provide knowledge that, when understood within its context, can provide invaluable information about the cause and possible preventative strategies for injuries. Therefore, researchers and clinicians need to be able to understand which and how the decisions made by the study authors affect the results. The next sections will delve into which aspects of functional tasks can be modified, how the movement can be quantified and used as dependent variables and other possible influencing factors found in these studies that need to be considered for a comprehensive and clinically applicable conclusion.

2. WHICH FACTORS CAN BE MODIFIED DURING FUNCTIONAL TASKS?

Besides the choice of tasks, there are several small modifications that can be made to a given task that can have an important impact on the measured biomechanics. These modifications can influence the movement to a degree that effectively makes it a distinct task. Although most authors do describe all the details of the tasks performed, this information is often not found in the title or abstract (which might be the only part that some readers examine) and some studies neglect to report relevant information altogether. This ultimately leads to an erroneous view of equivalence of the tasks by the reader, which impairs the ability to correctly interpret the findings, particularly when comparing to other studies. There are a number of common task alterations and the most relevant are presented in this section. In addition, researchers have recognized that these task modifications can have an important effect on their findings. Therefore, there have been studies that have compared movement biomechanics while participants performed different tasks and tasks with slight variations. The aim of this section is to discuss the factors that may influence results from functional tasks and mention a few of the most relevant studies that have compared task variations, allowing the identification of research that still needs to be conducted.

Overall task

There are several studies comparing the biomechanics evaluated during different types of tasks. These studies may focus on distinct metrics and also assess the effects of a secondary factor. However, given that there are a multitude of task types, not all combinations have been compared.

Donohue et al. (2015) compared three-dimensional knee and hip angles and moments when healthy participants performed double-leg or single-leg landings or squats. Their results showed significant differences for all measured dependent variables, with double-leg squat presenting the highest hip flexion, knee flexion and knee abduction angles, single-leg squat presenting the highest hip adduction angle and double-leg landing the highest external knee abduction moments. They also investigated the correlation between the values obtained in the four tasks, finding significant correlation in the frontal plane mostly between the two landings and between the two squats. Earl et al. (2007) investigated the differences between two other tasks while also considering the effect of gender on hip, knee, and ankle kinematics. The study compared the anterior step-down with a double-leg drop jump. Their findings demonstrated an overall greater knee flexion and smaller rearfoot eversion during the drop jump and an overall greater knee abduction in females. There were also several interactions between the two factors, showing that gender differences are also affected by the choice of task. Finally, Tanikawa et al. (2013) focused on cutting and hopping tasks while evaluating three-dimensional knee angles and moments, as well as ground reaction forces and knee anterior forces. This study also found several effects of task on biomechanics. Specifically, during the hopping task they found smaller flexion angle and greater extension moment, as well as smaller abduction angle and adduction moment, highlight how this task might be more appropriate to identify abnormal movements in the frontal plane.

There are also a few studies that have focused on the activation pattern of lower limb muscles during the execution of different functional tasks. Saad et al. (2011) compared the activation of the gluteus medius, vastus medialis obliquus, vastus lateralis obliquus and vastus lateralis longus between the step-up and step-down tasks. In this study, they also compared the activation of these muscles between healthy participants and those that present anterior knee pain. Their findings showed that healthy participants presented higher activation of all muscles, which was, in turn, consistently greater in the step-up than in the step-down. In another study, Husted et al. (2016) chose to focus on the activation of knee joint muscles (hamstrings and quadriceps) immediately before the foot contact during cutting, single-leg hop and drop jumps. Besides only finding low to moderate correlations between the values measured in different tasks, they found that there were statistically significant differences between all tasks for all muscles but the vastus lateralis. In a more recent study, Rabello et al. (2021) compared the activation of two hip abductors (gluteus medius and tensor fascia latae) and rearfoot invertors and evertors (tibialis anterior and peroneus longus, respectively) before and after a fatigue protocol. During a single-leg squat and the propulsion phase of a single-leg hop, no differences were found between the pre and post-fatigue moments, but there was higher activation of the peroneus longus and the gluteus medius during the single-leg squat and higher activation of the tibialis anterior during the single-leg hop.

Goal of the task

Even within the same task there can be factors that could highly influence the results. One of the most important characteristics of a task is the ultimate goal. Although different studies may refer to a task simply as “hop”, one might instruct the participants to hop as far as possible (Kingston et al. 2020), while another might present a pre-determined distance, which, in turn, can be based on the participant’s anatomical characteristic (Sritharan et al. 2020) or their maximal jump (Thompson et al. 2016; Read et al. 2017). Similarly, when performing a drop jump (landing followed by an immediate second jump), the researcher's instruction might be to execute the second jump as quickly as possible or as high as possible. Another example of variation in task goals that is often seen is during squat-based tasks. In those, the squat depth can either be "the deepest possible" or until a pre-specified angle (e.g., 45° knee flexion) (Mohr et al. 2019; Batty et al. 2020). Despite them likely being called by the same name in different studies, the actual movement would be different and could drastically affect the evaluated dependent variables.

The effects of the goal of the task on biomechanics is best exemplified in hop tasks, where the participant can be instructed to hop the maximal horizontal distance or a pre-specified one. In order to identify the effects of distance on lower limb biomechanics, Ali et al. (2014) compared ground reaction forces, power, work and joint angles when participants jumped 30, 50 or 70 cm, finding that as horizontal distance increases so do posterior ground reaction force, ankle dorsiflexion, hip flexion and trunk flexion. On squat tasks, the goal is related to how deep the participant is asked to squat, as some might go as deep as possible while others may stop at a pre-determined angle. In order to quantify how squat depth affects the biomechanics results, Bazett-Jones et al. (2022) compared the two-dimensional knee abduction, hip adduction, pelvic drop and lateral trunk flexion angles at five different knee flexion angles of the single-leg squat (30°, 45°, 60°, 75° and 90°). Their results showed that values significantly differed for all dependent variables but trunk angle, increasing together with knee flexion angles and indicating that the squat depth instruction affects the results obtained.

Movement speed

In gait and squat-based tasks, which are not ballistic, an important characteristic is the movement speed. On treadmills and step ergometers controlling the speed is very straightforward. However, when these tasks are performed overground or on staircases, researchers need to rely on precise timing using timing gates or stopwatches. Using these time measurement instruments also adds another source of error, as participants may vary their speed within the recorded period. Nonetheless, gait studies rarely do not report the chosen speed, as it is well accepted that gait speed has an important influence on biomechanics. The choice is usually between a self-selected speed, a set speed for all participants or a speed chosen based on some performance parameters such as best recorded time in a 5km race or a percentage of the ventilatory threshold (Boyer et al. 2014; Agresta et al. 2018; Maas et al. 2018). Speed is also an important factor in squat-based tasks, albeit has garnered less attention in the literature and is occasionally unreported. Studies may choose to let participants perform each repetition at their self-selected speed (Weir et al. 2010; Friesen et al. 2021), while others may try to standardize the speed, executing each repetition within a pre-determined duration, usually between two and five seconds (Crossley et al. 2011a; Almeida et al. 2016). The way the duration is measured is not without its difficulties, as participants may find it hard to be precise even with the help of a metronome, which adds variability to the data.

To confirm the influence of speed on gait, Fukushi et al. (2017) compared kinematics and kinetic data while participants ran at speeds of 2.5, 3.5 and 4.5 m/s, finding that most measured dependent variables were significantly affected by speed. Similarly, Hollis et al. (2021) used wearable sensors to compare results from self-selected fast and slow speeds, also identifying increased values of pronation excursion, braking and impact in faster speeds. Meanwhile, regarding the influence of speed on squats, Talarico et al. (2019) found that when squatting in slower speeds, participants presented higher center of pressure sway range and area, indicating they were less balanced during this condition. However, the literature has yet to show how movement speed can affect the kinematics of squat-based tasks. For this reason, the first study in this thesis will aim to investigate this topic.

Box and step height

A few tasks require additional equipment to be executed. That is the case of step-downs and drop-jump/landings. Although staircases can be found in most places, it's rare to have one inside a laboratory. Therefore, researchers often need to make or purchase a replacement. When doing so, they need to choose the step height and, since there is no standardization, they can usually vary between 15 and 25cm (Jones et al. 2014; Pairot de Fontenay et al. 2018). Drop-jump/ landings require the participant to step or jump from a box and land on the floor. The box height can play an important role in the task difficulty and consequently on the movement biomechanics. Studies can choose to have only one box height for all participants (Hewett et al. 2005) or adapt it, usually based on participant height or lower limb length. The one-height approach is the most common though, as it is difficult to find in the market an adjustable box with enough resolution and therefore, adjustable boxes need to be custom-made.

There are several studies that have investigated how box height influences the performance of drop jumps (Peng et al. 2017; Prieske et al. 2019), but this can also affect the biomechanics. Besides comparing different horizontal distances, Ali et al. (2014) also sought to verify the effect of box height on lower limb mechanics. They did so by having participants drop-land from boxes that were 20, 40 and 60 cm and measuring ground reaction forces and joint angles, power and work in the sagittal plane. Their findings indicated that higher jump heights lead to higher ground reaction forces, knee and trunk flexion angles as well as decreased knee power and work. Another study that focused on drop-jumps confirmed the importance of box height on biomechanics. Dickin et al. (2015) compared jumps from boxes of 30, 40 and 50 cm and also found and effect on hip and knee flexion angles and an increase in joint moments, power and ground reaction with greater heights. This study concurrently measured the effect of a fatigue protocol on the same set of dependent variables, finding differences between moments that were, however, also dependent on the box height.

Arms and contralateral limbs position

Although jump-landing, hop, balance and squat-based tasks focus on the movement of the lower limbs, the motion of the arms and trunk can play an important role on the position of the body's center of mass, and therefore on the overall forces that the lower limbs need to produce and absorb. In addition, arm position can also affect balance by changing the location of the center of mass in relation to the base of support, likely also requiring the lower limbs to modify the movement strategy in order to account for these deficits. Specifically during jump-landing tasks, the arm position is important as it affects both the balance required for landing and how high the jump is (Lees et al. 2004; Ashby et al. 2019), altering the height from which participants need to land from, which consequently affects biomechanics as previously discussed (Ali et al. 2014; Dickin et al.

2015). Arm position can be standardized by placing it at the shoulders or hips (Hetsroni et al. 2020; Friesen et al. 2021; Guimaraes et al. 2023) or can be free (Hewett et al. 2005; Von Porat et al. 2007). Researchers that adopt free arm movement argue that this is more approximate to the participants' regular movement pattern, but it is accepted that arm movement strategy can affect the overall biomechanics (Lees et al. 2004; Ashby et al. 2019).

Similarly to arms position, in single-leg tasks the position of the contralateral limb also is a component of the movement that needs to be considered. For example, in single-leg squats, the foot not executing the movement can be by the participants' side, in front or in the back (Khuu et al. 2016; Warner et al. 2019). This position may also affect balance, range of motion and other biomechanical factors. Although there can also be an effect of foot position on drop-landing tasks, the literature so far has only evaluated its effects on squats. On a series of studies, Khuu et al. (2016, 2021; 2019) compared the kinematics, kinetics and muscle activation of hip and knee muscles while participants squatted while keeping the foot behind, in front or next to the involved foot. In the first study, they measured three-dimensional joint angles and joint moments of trunk, pelvis, hip, knee and ankle. The findings showed significant differences in most analyzed dependent variables, however, there was no single squat variation that consistently showed higher or lower angles or moments (2016). In a second study, males and females were compared across these three variations, finding that there were several sex differences and that in some dependent variables the task variation interfered with the sex comparisons (2019). Finally, in their third study, they focused on the activation of the gluteus maximus, gluteus medius, tensor fascia latae, rectus femoris and hamstring muscles. Again, the findings confirmed that, for most muscles, the position of the contralateral foot had an important influence on the results obtained (2021).

Surface

Another aspect that may influence biomechanics during functional tasks is the surface where the testing takes place. This aspect is most commonly seen in gait studies, but its role should not be ignored in landing and cutting tasks. Although when studies conduct testing in a very specific surface, they generally report it clearly (Richardson et al. 2021; Zhou et al. 2021), even the “common” surfaces may be substantially different from each other. The stiffness of concrete, rubber, force plates or tile (which are often seen in biomechanics laboratories) is different and may influence how the participant deals with the forces generated during the movement (Ferris et al. 1998). However, the exact surface material present in the laboratory is often unreported. Gait evaluated on a treadmill is another situation where surface is a factor, in addition to the influence of wind resistance, inclination and belt width, limiting the possibility of comparison between studies (Van Hooren et al. 2020).

Similar to movement speed, most of the literature regarding the effects of surface on biomechanics has been conducted in relation to gait metrics. Boey et al. (2017) compared the tibial vertical acceleration (a metric that has been proposed to be associated with running related injuries) while participants of different experience levels ran on a concrete track, a synthetic running track or a woodchip trail. Although they did not find differences between the groups of runners, results showed that vertical acceleration was lower in the woodchip trail than in other surfaces. Besides comparing the effects of speed, Hollis et al. (2021) also compared running mechanics between a standard and a grass track, finding that pronation excursion, braking and impact were greater when running on the standard track. Although not all aspects of surface have been evaluated in drop-landing tasks, Richardson et al. (2020) found that knee abduction moment was smaller when landing on sand than when landing directly on a force plate. Further studies need to be conducted to examine the effect of surface on drop-landing kinematics and muscle activation.

Footwear and clothing

The final potential influential factors are clothing and other things that may be attached to the participants' bodies. If the clothing worn during the testing is different than the participants habitual one or have some particular characteristic, there can be some discomfort that may lead to unnatural movement patterns or even directly affect biomechanics (de Britto et al. 2017). The footwear can also play an important role in how people move. The type of shoe (e.g., minimalist, traditional, barefoot) may affect how people absorb the forces experienced during gait and landing and the balance required during static tests and squats (Bowser et al. 2017; Alghadir et al. 2018; Sun et al. 2020). In a given study, participants may be tested in a wide range of shoe types, possibly leading to bias (Hafer et al. 2019). On the other hand, laboratories that have more resources tend to opt for having a selection of shoes from the same make and model in all sizes and having their participants wear those during testing (Boyer et al. 2014; Maas et al. 2018). Although this approach eliminates the type of shoe bias, it can introduce another issue with people feeling uncomfortable in shoes they have not worn before. In addition, research involving kinematics or muscle activation often needs to place markers or probes in bony landmarks or muscle bellies. For securing this gear to the participant, a lot of tape is required and the overall feeling of having equipment and tape (albeit lightweight) can make the participant feel as they are not able to perform their regular movement, affecting results. There is movement towards developing technology that will eliminate or reduce the requirement of markers for kinematic evaluation, but it still seems to be a long way from becoming the norm (Horenstein et al. 2020; Abdollah et al. 2021; Wade et al. 2022). These new instruments will be discussed in more detail in a later section.

The most studied aspect of clothing during functional tasks is footwear. Particularly in running, this topic had garnered a lot of interest during this century, with the introduction of minimalist shoes and the popularity of barefoot running. Sun et al. (2020) conducted a systematic review to identify the effects of shoe construction (e.g., shoe lace, midsole, heel flare, minimalist, bending stiffness, etc.) on running biomechanics, focusing on those measurements that are associated with running related injury. By analyzing 63 studies, they've concluded, among other things, that softer midsoles result in decreased impact forces and that thicker midsoles can attenuate the shocks during impact, highlighting the importance of considering the type of shoe used during gait evaluations. Footwear is not studied as much during landing tasks, although impact forces and shock are also highly relevant for them. Wang et al. (2017) compared ground reaction forces and muscle activity of rectus femoris, biceps femoris, tibialis anterior and lateral gastrocnemius while wearing basketball shoes or minimally cushioned shoes. The tasks evaluated were a regular drop-jump and a passive landing, with findings showing lower ground reaction force, peak loading rate and activation magnitude while wearing basketball shoes. Although shoes are by far the most investigated aspect of clothing, one study has also assessed the effect of wearing compressive shorts on the kinematics of various jump-landing tasks. In this study, de Britto et al. (2017) found that compressive shorts led participants to land with reduced knee flexion and knee valgus, confirming that clothing can also affect biomechanics during functional tasks.

Conclusion

The previous paragraphs describe a series of choices that researchers need to make when deciding to use a functional task to perform their evaluations. Although studies may try to conduct testing in the same way as previous literature, it is very difficult to maintain all factors identical and even minor ones may have an important effect on the final findings. Despite the plethora of factors that can influence the biomechanical results obtained during functional tasks, there are still studies that fail to report all the details required to truly understand the findings within its full context. The current literature has done a good job in conducting studies directly comparing these influential factors, which can help researchers interpret and compare their findings to other studies when utilizing similar but not exactly the same tasks. The studies discussed in this section are only a sample of what has been done and published, however, there are still several other possible comparisons that can be made and many other measurements and metrics that have not been explored. The choice of which metric to use is another extremely important factor for the use of functional tasks in clinic and research environments and is, therefore, the theme of the following section.

3. HOW TO QUANTIFY FUNCTIONAL TASK PERFORMANCE?

Besides the choices of tasks and variations that were previously discussed, another fundamental factor for the development of studies wishing to use functional tasks to answer a research question is what is actually being measured. Although the terminology is not standardized, we can divide the assessment of biological signals in biomechanics into three main components: instrumentation, measurements and metrics. Instrumentation is the actual measurement system that is being used to collect the data and in biomechanics the most common are three-dimensional motion capture, two-dimensional video, force plates, inertial measurement units and electromyographers (Hewett et al. 2005; Nakagawa et al. 2011, 2012; Saad et al. 2011; Husted et al. 2016; Räisänen et al. 2018). Signal regard the type of data that are being extracted and/or calculated from the instruments, with the most common ones being joint angles, joint moments, ground reaction forces and muscle activation signal (Hewett et al. 2005; Saad et al. 2011; Agresta et al. 2018; King et al. 2018). Finally, the metrics are the specific values that represent the signal and are inputted into the statistical analysis (also commonly referred to as variables of interest or dependent variables depending on the context). Common metrics in biomechanics are peak vertical ground reaction force, average muscle activation during a time epoch, joint angle at foot contact and joint angle displacement (Ali et al. 2014; Trulsson et al. 2015; Agresta et al. 2018; Pairot de Fontenay et al. 2018). All these components and how they may affect the interpretation of the data will be discussed in this section.

Instruments

The main contributor for the choice of instrumentation is the availability at a given laboratory. Most instruments are often expensive and, therefore, not all researchers are able to simply choose, as they must use the ones they already have present in their laboratory. Three-dimensional motion capture is considered the gold standard to measure kinematics and is by far the most common instrument found in biomechanical studies (Hewett et al. 2005; de Britto et al. 2017; Glaviano et al. 2020; Desai and Gruber 2021) . Not all systems are the same and their differences may affect the quality of the data (e.g., markerless, passive or active marker systems, number of cameras, acquisition frequency, etc.). A low-cost way of measuring kinematics is using standard video recordings (from a smartphone or regular cameras)(Räisänen et al. 2018; Rabello et al. 2021; Bazett-Jones et al. 2022). This method obviously is only able to capture movement in two dimensions and tends to have a lower acquisition frequency as well. Another up-and-coming approach to evaluate kinematics is using inertial measurement units, which are usually equipped with a combination of accelerometers, magnetometers, gyroscopes and global positioning system (Nakagawa et al. 2012; Rawashdeh et al. 2016; Hollis et al. 2021). This instrument has the advantage of being smaller and therefore disturbing the participants to a lower degree. In addition, they can be used more easily on the field, which opens several research opportunities. However, researchers using these instruments need to conduct complex calculations to obtain the most common kinematic signals such as joint angles and they might not be as accurate (Faisal et al. 2019; Hollis et al. 2021; Ekdahl et al. 2023).

Exiting the instruments used for kinematic evaluation, the two most commonly seen are force plates and electromyographers. Force plates do often work in combination with 3D motion capture and are therefore, often seen together in biomechanics laboratories and manuscripts (Ali et al. 2014; Dickin et al. 2015). The gold-standard for force measurements are the three-dimensional force plates, which are able to separate the forces seen on the vertical, antero-posterior and medio-lateral directions (Dickin et al. 2015; Khuu et al. 2016). Another type of force plate, which has a lower cost, is the single-axis plates, which are only able to measure the resultant of the three force vectors (Hart et al. 2019). Finally, for the measurement of muscle activation the instrument used is the electromyographer, which reads the electrical signal coming from the muscle. The most common type of electromyographer uses surface electrodes, but there are also those that use needles and fine-wire electrodes to get closer to the actual muscle (Robb et al. 2021; Akuzawa et al. 2023). Both types have their pros and cons that affect the signal obtained and will be briefly discussed later on.

Because all these instruments acquire data at high frequencies and can be influenced by factors such as light, electricity and mechanical issues, most times the raw signal captured contains incorrect or missing data. Before obtaining the signals from which the metrics will be extracted, the raw signals must pass through processing steps that can also affect the data. There are a number of engineering-focused studies that seek to understand the effects of processing on data (Chang et al. 2013; Ma et al. 2021; Crenna et al. 2021), but, as this is outside the scope of this thesis, only a brief overview will be given. The most common processing steps are filtering, rectifying and interpolation. Filtering removes part of the data that are above, below or exactly a certain frequency which are judged to be likely due to errors and not what was truly happening on the participant. Another processing step is rectification, which simply converts the negative values found in the raw signal to positive values, making it possible to calculate averages. The final one to mention is interpolation, which is most commonly seen on 3D motion capture. Interpolation is necessary when the signal is absent for some frames (e.g., a marker that is occluded) and consists of estimating the missing values based on the ones occurring before and after. As the estimation is not perfect and depends on the number of missing data points and on the type of algorithm used (e.g., polynomial, spline, linear, etc.), there can be errors and differences between studies.

Measurements

Each instrument is able to measure different aspects of movement, requiring more or less calculation before arriving at a specific measurement. The measurements discussed in this paragraph all fall within the category of "kinematics" which is defined as "the branch of (bio)mechanics concerned with the motion of objects without reference to the forces which cause the motion". The main measurement obtained from 3D motion capture is joint angle, which can be in the three planes and represent the angle formed between one segment and another. The corresponding measurement obtained from 2D video are vector angles, which are only in one plane and do not consider the whole segment, but rather two lines (i.e., vector) that are used to represent the segment in that plane (e.g., the line from the center of the knee until the center of the ankle represents the leg). The differences between the 3D and 2D measurements are often ignored in the literature, which is one of the main sources of confusion or misinterpretation of findings obtained in functional tasks. Although one may assume that the angle calculated between leg and thigh vectors in 2D are the same thing as the angle between the leg and thigh segments in 3D while “ignoring” two of the planes, this is not exactly the case. Kingston et al. (2020) found that there was a moderate correlation between 2D and 3D trunk frontal plane angles during the single-leg hop, but no correlations during single-leg squats and drop-jumps. Correlations for the same tasks for hip frontal plane angle were moderate to strong. However, there were no correlations between 2D and 3D knee angles in the frontal plane during any of the tasks. This is particularly important because frontal plane metrics during functional tasks are usually adopted as a representative of injury risk, given that there are studies that have associated increased incidence of anterior cruciate ligament rupture and presence of patellofemoral pain with elevated frontal plane movement during functional tasks (Hewett et al. 2005; Willson and Davis 2008a, b; Nakagawa et al. 2012; dos Reis et al. 2015). The most accepted terminology for the frontal plane knee motion is “knee abduction” when referring to 3D measurement and “knee valgus” or “knee frontal plane projection angle” when referring to 2D. However, the lack of consistency in the nomenclature contributes to this confusion, as some studies use the terms interchangeably (Claiborne et al. 2006; Bazett-Jones et al. 2022). Besides angles, motion capture and 2D video also can measure the position of segments or specific points. Other possible measurements are linear velocities and acceleration and angular velocity and accelerations as they are the product of positions or angles and time, respectively. Inertial measurement units typically use the reverse route to obtain their measurements, that is, they start from the acceleration and through a series of calculations obtain the joint angles. However, studies using these wearable sensors often use different metrics that require only the acceleration measurements (Strohrmann et al. 2012; Boey et al. 2017).

Kinetics is defined as “the branch in (bio)mechanics that is concerned with the relationship between the motion and its causes, specifically, forces and torques”. The assessment of forces during the execution of functional tasks is conducted using force plates, which are able to quantify how much force is being applied to the ground (i.e, ground reaction forces). Forces can be represented by vectors and these vectors can be decomposed into the three vectors representing the vertical, medio-lateral and antero-posterior directions. These specific directions can give important information for assessments during functional tasks, as they allow for the calculation of the joint moments that are occurring at the three planes of movements. For example, the knee abduction moment is a very popular measurement for its association with knee injuries (Hewett et al. 2005). Although there might be differences in force plate technology (i.e., piezoelectric or strain gauge), those that are able to decompose the force vector are considered the gold standard. In the market, there are also force plates that unidirectional, meaning that they only record the resultant force (Hart et al. 2019). Although these plates can also be useful, they limit the assessments that can be conducted and disregard information that can be relevant. The measurements most commonly obtained from force plates are ground reaction forces and joint moments. Ground reaction forces are simply the force values that can be either three-dimensional or unidimensional, from these values there are several metrics that can be obtained and used for assessments in medical or performance contexts. By combining ground reaction forces and three-dimensional motion capture using inverse dynamics, it is possible to obtain joint moments, which represent the force that leads to angular acceleration of the joints.

The final measurement that warrants discussion because of its relevance for functional tasks is muscle activation. Using electromyographers, it is possible to capture the electric signal used by the many muscle fibers to start the contraction process, amplify it and reach a signal containing the motor unit action potentials presented in a superimposed manner (Konrad 2005). Because of this superimposition, muscle activation measurements are not as intuitive at first glance as joint angles, although it is possible to visualize the signal strength increasing as the muscle contracts, therefore, the correct reporting and understanding of the calculation of specific metrics is fundamental. However, there are some factors that need to be considered prior to this calculation that affect the actual measurement. The first is electrode positioning. Differently to reflective marker placement, electromyography electrodes are placed on the muscle belly as opposed to more specific anatomical points. The Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) project sought to standardize the electrode placement using the anatomical points as reference and it has been well accepted (Hermens et al. 2000). However, not all muscles are represented in SENIAM and, most importantly, there are inherent anatomical differences between participants that result in differences in positioning between participants. For example, it is recommended that the electrodes for the assessment of the soleus muscle are positioned at 66% of the distance between the medial condyles of the knee and the medial malleolus, however, some participants may have different conformations of the posterior calf muscles, with the Achilles tendon starting at different heights (Devaprakash et al. 2020). This translates as some participants having the electrodes positioned over the central portion of the muscle belly while others might have them at the distal portion of the same muscle. Another issue that arises concerning electrode positioning is the muscle crosstalk (Talib et al. 2019). This occurs when the electromyographer captures the signal from adjacent muscles and mixes it with that of the muscle of interest. Crosstalk affects particularly smaller muscles and can be especially relevant when the adjacent muscle is an antagonist of the muscle of interest. One possible solution for muscle crosstalk is using fine-wire electrodes, which are inserted into the desired muscle with a needle and are more precise (typically with the help of ultrasound). This method, however, has the downside of being much more invasive, which can possibly lead to participants altering their movements and ultimately affecting the results. Finally, because there are several factors that influence the signal recorded by the electromyographer (e.g., fat layer, skin impedance, etc.), the signal by itself cannot be used to compare participants or even a participant with itself in multiple days. Therefore, muscle activation measurements are often normalized by a reference contraction, which can be a maximal explosive movement or a maximal voluntary isometric contraction (Konrad 2005). However, the normalizing contraction can be conducted in many ways and therefore, significantly affect the interpretation of the results.

Metrics

The last level of decision-making before arriving at the variable of interest to be inserted into the statistical analysis is the choice of metric. The metric is how the researchers will choose to represent the measurement. This is usually done in two ways: whole time-series or discretization. The latter is by far the most common approach adopted and it consists in finding one or more values to represent the measurement. The predominant methods are averaging the values, finding the maximal or minimal, identifying the value at a specific event (e.g., at foot contact from a landing) or calculating the difference in values between two events. Although it makes analyses simpler, the discretization approach has the disadvantage of discarding most of the data in favor of only a couple of numbers, which can significantly alter the findings. In order to protect from this disadvantage, lately some studies have decided to input the whole time-series data into the statistical analysis and identifying the points in which significance is found (Pataky 2010; Pataky et al. 2013; Serrien et al. 2019). This approach is more complicated and often requires programming knowledge, making it more inaccessible, but techniques such as statistical parametric mapping, principal component analysis and cubic splines are currently found in the biomechanics literature (Boyer et al. 2014; Maas et al. 2018; Harrison et al. 2021).

Because there are countless combinations of measurement and metrics, researchers must limit the ones they select to reduce the possibility of type I error (false-positive). In most studies where functional tasks are used to compare groups, the effect of an intervention or to identify increased risk of injury, the measurements selected are based on previous literature. For example, for studies where anterior cruciate ligament injury is the focus, the measurements of knee abduction and knee abduction moment are usually adopted (Houck and Yack 2003; Ruan et al. 2017; Zago et al. 2021). Similarly, for patellofemoral pain, knee abduction is used in addition to hip adduction, pelvic drop and ipsilateral trunk lean (Nakagawa et al. 2012; Schmidt et al. 2019; Gilmer et al. 2020; Harput et al. 2020). In addition, studies where the goal is to identify a new variable that might be clinically relevant usually explain in more detail (from a neuro or biomechanical perspective) why that variable should matter. For muscle activation the choice of metric is even more relevant, as the raw signal interpretation is not very intuitive. Muscle activation analysis typically fall within the magnitude, time or frequency domains (Boling et al. 2006; Crossley et al. 2011b; Leporace et al. 2011) and the quality of the signal as well as the post-processing play a very important role.

The main issue with the plethora of metrics available is that researchers often interpret their findings based on the measurement rather than the specific metric. For example, a study that adopted the knee abduction displacement angle as the variable of interested might find significant differences between two groups and say that their findings agree/disagree with the current literature, even though the studies they cite might have used only the peak abduction angle. It is likely that those values will be correlated, but it is surely not a given considering that the initial knee abduction angle may vary substantially between participants or groups and this is accounted for in the displacement angle but not in the peak angle. This assumed equivalence of metrics within the same measurement can cause confusion and erroneous conclusions in the sports medicine literature. Another error that can occur in the interpretation of the findings related to the metrics is theoretical, but untested, correlation between metrics. The best example of this situation is the relationship between the activation of a muscle and the movement of the joint in which the muscle acts. The gluteus medius muscle is a prime mover of hip abduction and, therefore, when it is more activated it is assumed that there will be some degree of hip abduction or at least resistance against hip adduction from an external load. This is the case in open chain tasks or tasks where there is clear external resistance (e.g., strengthening exercises), however, this relationship during functional tasks may not be as direct as there are many other external and internal forces affecting the hip joint during these movements. Nonetheless, studies hypothesize, or even make conclusions about kinematics while only actually assessing muscle activation, which, once again, might lead to confusion and errors in interpretation. In order to test and discuss this issue, the second study in this thesis will focus on the relationship between muscle activation and kinematic metrics.

4. OTHER INFLUENCING FACTORS

There are two other elements of research, which are not exclusive to functional tasks, that warrant attention in such biomechanical studies. The first is the categorization of participants, which can happen in either an objective or an arbitrary manner. This can obviously influence the conclusions of studies that use some sort of group comparison, particularly if the groups are not well defined. The group comparison is a highly useful approach in sports medicine as it allows the researchers to use a cross-sectional design instead of prospectively following participants or conducting interventions (Houck and Yack 2003; Boey et al. 2017; Maas et al. 2018; Khuu and Lewis 2019; Zago et al. 2021; Hawkins et al. 2023; Araki et al. 2023). Examples of objective group categorization are people who have ruptured the anterior cruciate ligament (Houck and Yack 2003), people older than 60 years (Araki et al. 2023) or people that were assigned as females at birth (Khuu and Lewis 2019). However, some things cannot be as objective, as the criterion can exist in a continuum. In these situations, researchers must make a decision about the cut-off values necessary to be assigned to one group or the other, however, they are often arbitrary. Examples of arbitrary group categorization are elite or non-elite athlete (Zago et al. 2021), physically active or sedentary person (Hawkins et al. 2023) or experienced or novice in a given sport (Boey et al. 2017; Maas et al. 2018). One research might define as elite those that are within the top 5% of their sports, while another might set the cut-off value at 10%. If researchers adopt the same nomenclature when referring to groups with different characteristics due to the chosen cut-off values, there can be confusion and misinterpretation of findings. The third study in this thesis, besides the choice of metrics, deals with classification of runners within groups of difference experience levels.

Another element that is important and should receive attention is the reliability of the measurements (and consequently of the metrics). Reliability can be divided into two main categories: inter and intra-rater. The first compares the measurements obtained by two or more distinct raters, assessing if results are dependent on who conducted the assessment (Trajković et al. 2022). For functional tasks, this type of reliability is not as pertinent, given that the measurements are dependent on the instruments and the influence of the rater is limited to the instructions given and those are (theoretically) held constant in each study. Intra-rater reliability compares the assessment conducted in different moments (typically different days) and is much more relevant for functional tasks (Cavanaugh et al. 2017). When the same task is executed in different moments, the results are often not exactly the same. This can be simply due to measurement error/variability arising from the instrument or it can be that the participant is performing the task differently. It is normal for there to be some variability in performance, but if there are important and systematic differences within a sample it indicates that there are intervening factors affecting the results (e.g., learning effect, time of day, fatigue, etc.). Therefore, it is important to understand how reliable each measurement used is in order to interpret the results within the appropriate context. The fourth study in this thesis concerns the reliability of force measurements during a series of functional tasks.

5. THESIS RATIONALE

Functional tasks are a valuable way to simulate the loads faced by athletes and physically active persons during sports and other activities, allowing researchers to assess the status, detect the impairments and identify improvements in the way people move. Most of the current knowledge on movement biomechanics stems from studies that have employed functional tasks to assess the participants (Hewett et al. 2005; Nakagawa et al. 2012). From choice of task (and task variations) to instruments, measurements or metrics, there is an infinite combination of factors that affect the results of each study. For many reasons, we cannot expect all studies to be performed the same way. However, many studies seem to be missing some nuance when discussing their findings in the context of the literature. Although it might seem unpractical prefacing all comparisons with a complete discussion of the task used and the metrics calculated, not doing so oftentimes can result in researchers making misleading or incorrect statements that can ultimately lead to difficulty in applying the science in the clinical practice. Correct and detailed reporting of all influencing factors is a fundamental step on the path to high quality research. Regardless, the better understanding of current and future findings can be invaluably assisted by studies investigating how these factors influence biomechanical data. The present thesis will consist of four studies that, although not directly related to each other, focus on different aspect of influencing factor for the use of functional tasks in the sports and clinical literature and practice. Together, they will add to the current knowledge and allow the discussion of functional tasks aspects that are often overlooked.

6. AIMS

General:

To evaluate how distinct factors influence biomechanical assessment during functional tasks using different tasks, metrics and populations.

Specific:

1 – To assess the differences in lower limb kinematics between the single-leg squat, anterior step-down and lateral step-down (Study 1).

2 – To assess the influence of movement speed on the lower limb kinematics during three squat-based movements (Study 1).

3 – To identify the relationship between frontal plane knee and hip kinematics and two metrics of muscle activation of lower limb and core muscles during single-leg squats and anterior step-downs (Study 2).

4 – To compare the differences between people with different running experience levels on linear and angular stiffness during running gait (Study 3).

5 – To measure the test-retest reliability of force measurements during the execution of functional tasks with progressive difficulty in a healthy and pathological population (Study 4).

STUDY 1 – THE INFLUENCE OF TASK TYPE AND MOVEMENT SPEED ON LOWER LIMB KINEMATICS DURING SINGLE-LEG TASKS

1. CONTEXT

The project where the first two studies from the thesis stem from was conducted during the first year of the PhD. It was originally part of a larger study which focused on kinematics during the execution of landings in different conditions. However, given the relevance of squats and step-down in the literature, we decided to expand on its variations. Therefore, it was conducted as a stand-alone project which included kinematics and muscle activation measurements.

The aim of the present study was to represent how variations to the functional task (speed and task itself) influenced metrics commonly assessed in the literature in the context of injury risk. This deals with specific aims 1 and 2.

The study was submitted to Gait and Posture on January 28th, 2022, and accepted on May 17th, 2022. The following text differs from the publication in a few small ways in order to standardize the terminology and the structure. The study content is unchanged.

Rabello, R., Bertozzi, F., Brunetti, C., Silva Zandonato, L., Bonotti, A., Rodrigues, R., & Sforza, C. (2022). The influence of task type and movement speed on lower limb kinematics during single-leg tasks. Gait & Posture, 96(May), 109–116. https://doi.org/10.1016/j.gaitpost.2022.05.020

2. ABSTRACT

Background: Single-leg squats and step-downs are commonly used to assess kinematic patterns that may be linked to injuries. Task type and movement speed may influence the outcomes of interest because of different balance requirements. This study aimed to investigate the influence of task type and movement speed on lower limb kinematics.

Methods: This is a cross-sectional within-subjects study where 22 physically active females performed three single-leg functional tasks (Squat, Anterior step-down, and Lateral step-down) at three movement speeds (slow [5s], fast [2s], and self-selected), while three-dimensional kinematic metrics were recorded. Displacement values from the initial position in single-leg support until 60° or peak knee flexion were calculated. Two-way repeated measures ANOVA was used to compare tasks and speeds, and Cohen's d effect size (ES) was calculated for significant pairwise comparisons.

Results: At 60°, lateral step-down presented the greatest hip adduction (large ES) and internal rotation (small ES). The anterior step-down had the lowest knee abduction displacement while the squat had the greatest (small to medium ES). At peak knee flexion, values increased but differences between tasks followed a similar pattern. Slow speed induced smaller displacement angles at the knee and hip (trivial to small ES).

Conclusions: When knee abduction is the variable of interest, the single-leg squat may be the best task since it elicits the greatest displacement, but when evaluating hip motion, lateral step-down might be best. Knee abduction and internal rotation were lowest in the slow condition, suggesting that faster speed may be more appropriate to detect abnormal movement patterns. However, the small difference in absolute values (i.e., degrees of

movement) may indicate that the differences are not clinically significant, particularly for speed comparisons. Researchers and clinicians should take this into consideration when choosing the most appropriate task and the instruction to give during its execution.

3. INTRODUCTION

Functional tasks such as squats and step-downs are commonly used to assess kinematics in clinical and laboratory settings (Jones et al. 2014; Howe 2017; Fitarelli et al. 2020; de Albuquerque et al. 2021). A typical outcome of these tasks is knee mechanics in the frontal plane (named abduction when measured in 3D and named valgus when measured in 2D), since they are generally associated with lower extremity injuries (Nakagawa et al. 2012; Räisänen et al. 2018; Schmidt et al. 2019; Gilmer et al. 2020; Harput et al. 2020).

Functional task type influences joint kinematics because each type of movement may load the joints differently. Tasks such as squats and step-downs present lower external load than landings or changes of direction (Earl et al. 2007; Donohue et al. 2015). However, their popularity stems from being considered safer (useful in return to sport evaluations) and allowing visual detection of misalignments, providing immediate feedback (Earl et al. 2007; Yamazaki et al. 2010; Donohue et al. 2015). The single-leg squat (SLS), anterior step-down (SD_{ANT}) and lateral step-down (SD_{LAT}) are functional tasks commonly used to evaluate different populations (Howe 2017; Lopes Ferreira et al. 2019; Fitarelli et al. 2020; Harput et al. 2020; de Albuquerque et al. 2021). However, despite presenting similar movement patterns, the differences between these tasks (e.g., position of the contralateral limb and movement depth) may affect joint kinematics (Lewis et al. 2015; Hatfield et al. 2017).

Even within the same functional task, decisions regarding movement execution are required since tasks are not standardized. Previous research has identified differences in single-leg squats with altered contralateral limb position (Khuu et al. 2016) and in step-downs with different step-heights (Lewis et al. 2015). Another important factor is movement speed. Participants can perform tasks either at self- selected (Weir et al. 2010; Friesen et al. 2021) or controlled speed (Crossley et al. 2011; Alenezi et al. 2014b, a; Almeida et al. 2016; Lopes et al. 2021). However, even controlled speed can vary from two (Crossley et al. 2011) to five seconds per task (Alenezi et al. 2014b; Herrington 2014; Almeida et al. 2016). Talarico et al. found that faster SLS speeds resulted in decreased center of pressure sway range and area (2019), indicating that slow speeds may be more challenging.

Although these tasks are popular in the literature, it is still unclear how altering task type and execution speed influences kinematic metrics associated with injuries. The identification of a possible task- or speed-dependency may be useful for better interpretation of results and for choosing the more appropriate clinical test for a given population. Therefore, the aim of this study was to assess joint kinematic metrics associated with injuries when performing three single-leg functional tasks (SD_{ANT}, SD_{LAT} and SLS) at three different speeds (slow, fast, and self-selected). We hypothesized that the slow speed and the SLS and SD_{ANT} tasks would produce greater frontal and transverse plane motion due to it being the speed that requires greater balance and the tasks that require greater range of motion.

4. METHODS

Experimental overview

This is a cross-sectional within-subjects study. Ethical approval was obtained from the University ethics committee. After protocol explanation, all participants gave written informed consent. STROBE guidelines were used for reporting this study.

Each participant performed all tests in a single experimental session between June and July 2021 (Figure 1). They wore clothes that would not interfere with the reflective markers' positioning and their personal exercise shoes. After five minutes warming-up on a stationary bicycle, 38 reflective markers were positioned in selected anatomical landmarks. Afterwards, participants performed the SD_{ANT}, SD_{LAT} and SLS tasks at their self-selected speed and subsequently at a controlled speed.

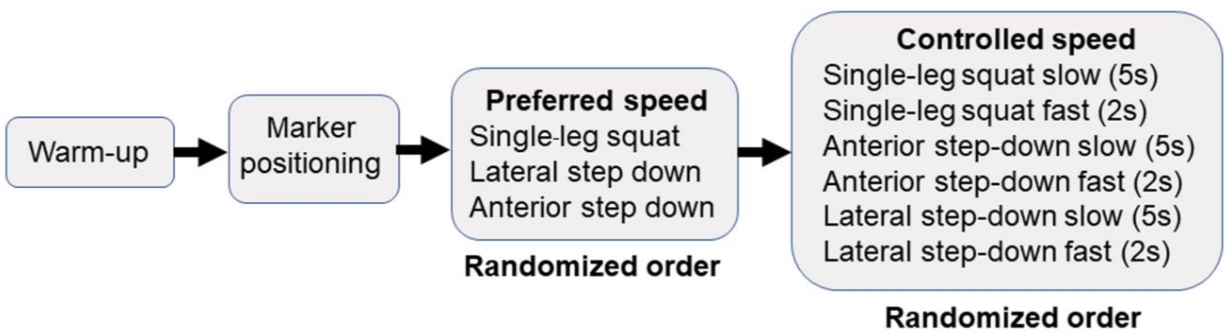

Figure 1. Flowchart of the protocol's steps.

Participants

The appropriate sample size was calculated a priori using G*Power (version 3.1.9.6; University of Trier, Trier, Germany), with a significance level of p = 0.05 and power of 1-β = 0.80. Effect size f was estimated at 0.33, according to a previous study (Lewis et al. 2015). A sample size of 21 was indicated and an additional participant was evaluated to anticipate data loss.

Participants had to be females aged between 18 and 40 years, perform physical activity at least twice a week and be capable of performing all tasks without pain or discomfort. History of knee surgery or injuries to ligaments or menisci, lower limb injury within the previous six months, conditions affecting balance, or cardiovascular disorders limiting the execution of the exercise were exclusion criteria. Only females were included in the study since the tasks used are more commonly found in studies investigating pathologies that affect women to a greater degree, such as patellofemoral pain and ACL injuries (Boling et al. 2010; Stanley et al. 2016).

Tasks and movement speed

Prior to starting each task, participants stayed in single-leg support with the ipsilateral limb fully extended and the contralateral knee flexed at 90° and the hip at 0° for one second to establish a baseline. For SD_{ANT} and SD_{LAT}, after recording of this standardized starting position, they positioned the contralateral foot next to the support one prior to the start of the tasks. SD_{ANT} and SD_{LAT} were performed on a step corresponding to 10% of the participants' height. Participants stood on the step while supporting the weight with the preferred foot and the contralateral foot next to it. For SD_{ANT}, they squatted and lightly touched the contralateral heel on the floor in front of the step before returning to the initial position, without lifting the support heel from the step (Figure 2B). The SD_{LAT} was performed similarly, but with the contralateral heel touching the floor to the side (Figure 2C). The SLS was executed with the preferred foot by squatting as low as comfortable (minimum 60° of knee flexion) and returning to the initial position (Figure 2A), keeping the contralateral knee flexed to 90° and the thigh perpendicular to the ground. For all tasks, the hands were crossed over the shoulders. A repetition was considered invalid when the participant lost balance, removed the hands from the shoulders, touched the floor with the contralateral foot (for SLS) or used the contralateral foot for support (for SD_{ANT} and SD_{LAT}).

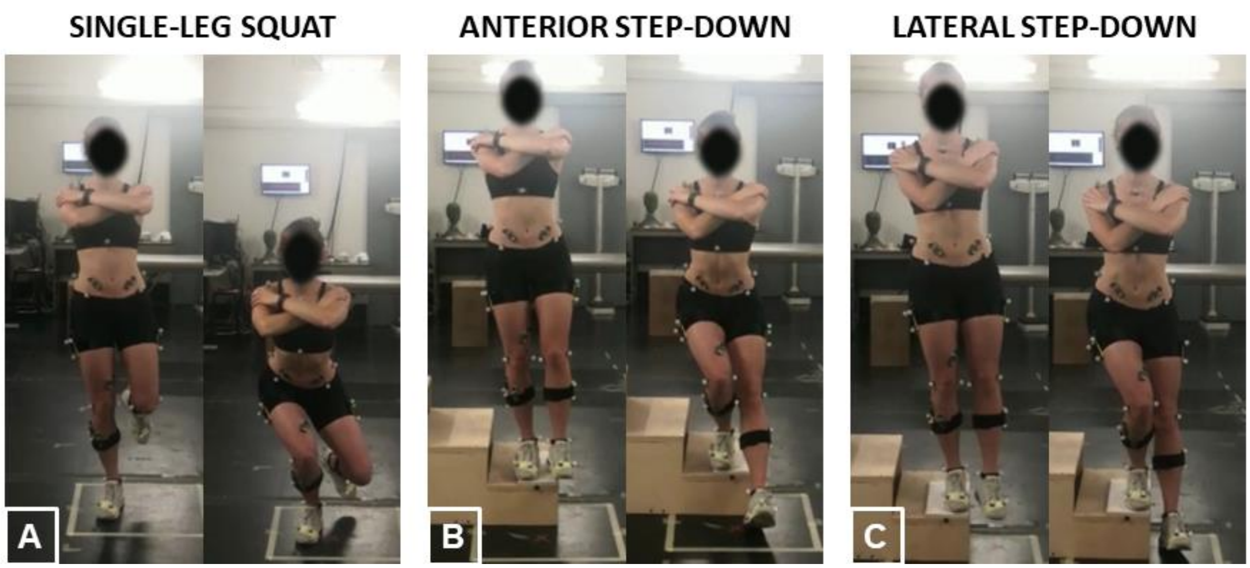

Figure 2. Representation of one participant performing the Single-Leg Squat (SLS), Anterior Step-Down (SD_{ANT}) and Lateral Step-Down (SD_{LAT}) at the initial and maximum knee flexion frames.

Participants performed five repetitions for each task, which were performed consecutively unless a pause was needed to regain balance in between repetitions. In either case, participants were required to start each repetition with the support knee fully extended. If a repetition was invalid, the participant performed an additional one at the end. If two or more repetitions were invalid in one trial, the whole trial was repeated after appropriate rest. A minimum interval of one minute was observed between trials to avoid fatigue.

Participants performed the SD_{ANT}, SD_{LAT} and SLS at three different speeds: fast, slow, and self-selected. In the slow speed condition the task had to be performed in five seconds (three seconds for the eccentric and two seconds for the concentric phase) (Alenezi et al. 2014b; Herrington 2014; Almeida et al. 2016). The fast one was performed in two seconds, (one second per phase) (Crossley et al. 2011). For self-selected, participants performed the tasks at whichever speed they felt more comfortable. This condition was performed before the other two so the controlled speeds would not interfere with the participants' choice of speed. Prior to data collection, participants practiced all tasks until they felt comfortable with the movement's execution and timing. Tasks' order within the preferred speed block (three tasks) and within the controlled speed block (six tasks) was randomized a priori.

Data analysis

A nine-camera motion capture system (60Hz, BTS S.p.A, Garbagnate Milanese, Italy) and marker set containing 38 reflective passive markers were used (Figure 3). A ten-second static recording was performed at the beginning of each experimental session. The markers' three-dimensional position was reconstructed using SMARTTracker (BTS S.p.A, Garbagnate Milanese, Italy), and analyzed using Visual3D (C-Motion, Inc., Germantown, USA). The pelvis segment angle was expressed according to the laboratory's global reference frame, while knee and hip joint angles were expressed according to the proximal segment. A fourth-order Butterworth low-pass filter with 6Hz cut-off frequency was used (Robertson and Dowling 2003). A Visual3D hybrid model with a Cardan X-Y-Z rotation sequence was used to determine joint angles, corresponding respectively to the mediolateral, anteroposterior, and vertical axis.

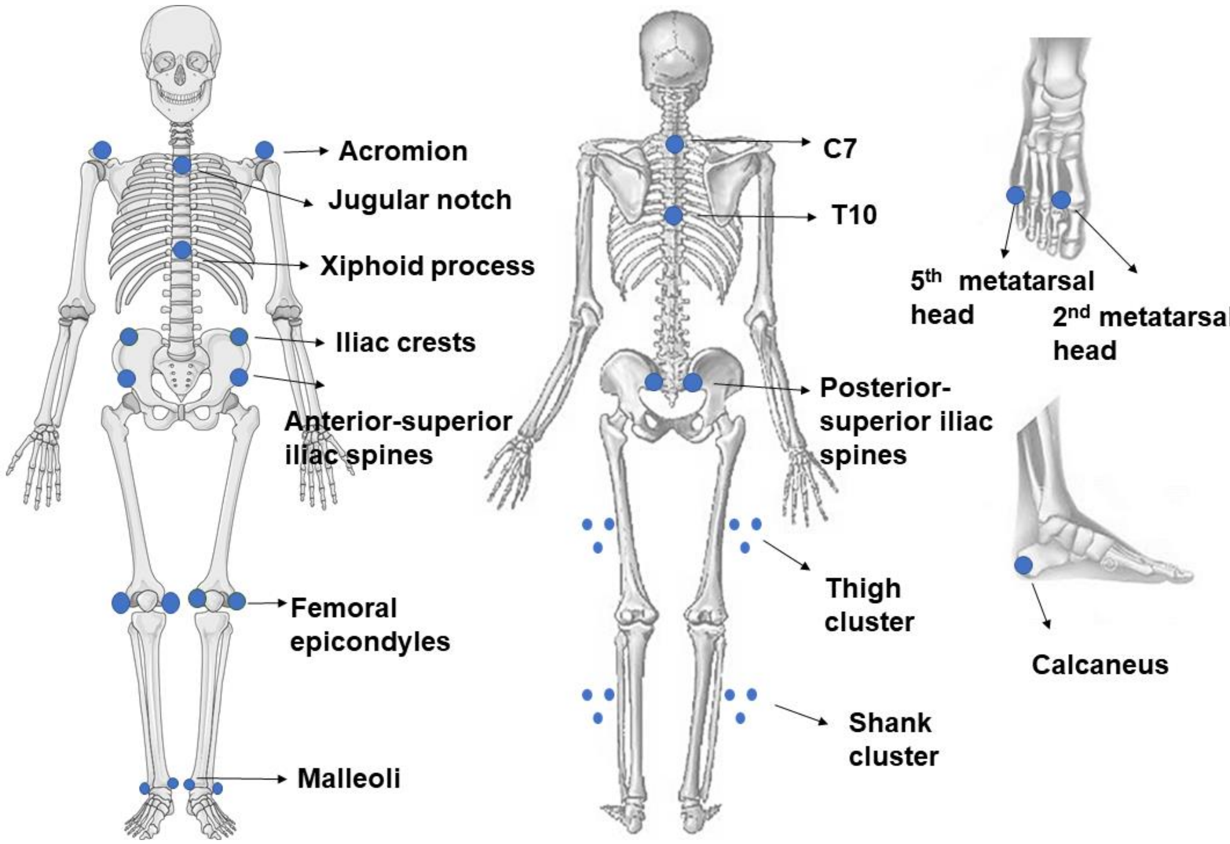

Figure 3. Anatomical landmarks where the reflective markers were positioned.

The following metrics were measured and considered the dependent variables: pelvic contralateral drop, support hip adduction and internal rotation, support knee flexion, abduction, and internal rotation. The metrics were evaluated at 60° of knee flexion in the descent phase and at peak knee flexion of the support limb. The angular displacement was computed as the difference between angles at the initial position (average of the standardized single-leg support position values for the nine conditions) and at the recorded event (either 60° or peak knee flexion). The first repetition was discarded in the subsequent analysis.

Statistical analysis

Data normality was evaluated using the Shapiro-Wilk test. Sphericity was determined using the Mauchly Test of Sphericity and the Greenhouse-Geisser correction was applied when the assumption was violated. Separate 3x3 Two-Way ANOVAs with repeated-measures on both factors (speed and task) were performed to compare the dependent kinematic variables (mean of four repetitions) at both peak and 60° of knee flexion. When significant main effects were found, post-hoc pairwise t-tests with a Bonferroni correction were used. When interactions were present, the individual effects were analyzed separately. Analyses were performed on IBM SPSS Statistics version 20 (IBM Corporation, Armonk, USA) and alpha was set at 0.05. Effect sizes (ES) were calculated for each significant pairwise comparison to assess the magnitude of the significant differences using Cohen's *d*. Values were defined as: <0.2 = trivial, 0.2 to 0.5 = small, 0.5 to 0.8 = medium and ≥0.8 = large (Cohen 1988).

5. RESULTS

Twenty-two participants completed all conditions of the study (24.6 ± 3.4 years, 166.1 ± 7.2 cm; 59.1 ± 9.2 kg, 3.1 ± 1.3 weekly training sessions; three participants were left-footed). Average absolute peak knee flexion values were 77.7 ± 4.9 °, 69.4° ± 4.7 and 80.1° ± 7.5 for SD_{ANT}, SD_{LAT} and SLS, respectively. All participants reached a minimum of 60° of knee flexion for all tasks. Tasks performed at self-selected speed had a duration of 2.9 ± 0.7s, 2.6 ± 0.6s and 2.9 ± 0.7s for SD_{ANT}, SD_{LAT} and SLS, respectively. Figures 4 and 5 display the angular displacement values for task and speed for the selected metrics at 60° and peak knee flexion, respectively. Supplementary material 1 shows the individual value for each combination of task and speed.

Knee flexion at peak knee flexion was greater for SLS and SD_{ANT} than for SD_{LAT} (large ES). Knee abduction was greater for SLS and SD_{LAT} than SD_{ANT} at both 60° and peak knee flexion (small to medium ES) and greater for SLS than SD_{LAT} only at 60° (small ES). Knee internal rotation was greater for SD_{ANT} than SD_{LAT} at 60° and peak knee flexion and greater than SLS only at peak knee flexion (trivial to small ES). Hip adduction was greater for SD_{LAT} than SD_{ANT} exclusively at 60° (large ES). Hip internal rotation was greater for SD_{LAT} and SD_{ANT} than SLS at peak knee flexion (large ES) and greater only for SD_{LAT} at peak knee flexion (small ES). Contralateral pelvic drop was higher for SLS and SD_{ANT} than SD_{LAT} (medium ES), but only at 60°.

Knee flexion at fast and self-selected speeds were greater than slow (small ES). Fast speed presented greater knee abduction than slow only at 60° (trivial ES). Fast and self-selected speeds showed greater knee internal rotation than slow at 60° and peak knee flexion (trivial to small ES). Hip internal rotation was lower for slow than fast only at

60° (small ES). No effect of speed was found for hip adduction and contralateral pelvic drop. Only knee flexion displayed interaction between factors, where the speed differences are not as pronounced during the SD_{LAT} task.

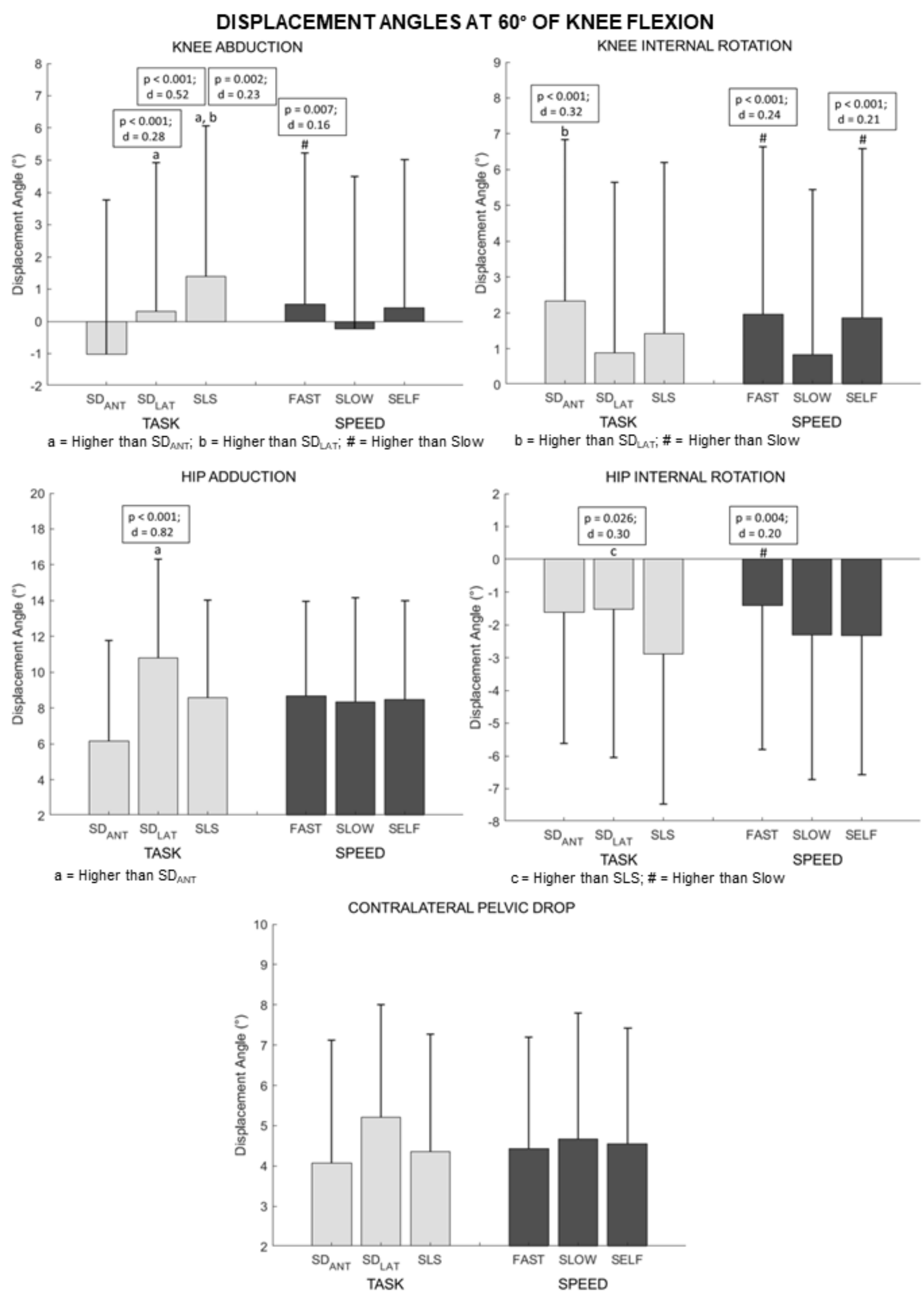

Figure 4. Displacement angle values for each variable of interest at 60° knee flexion. Task values are presented as the average of each task across the three speeds. Speed values are presented as the average of each speed across the three tasks.

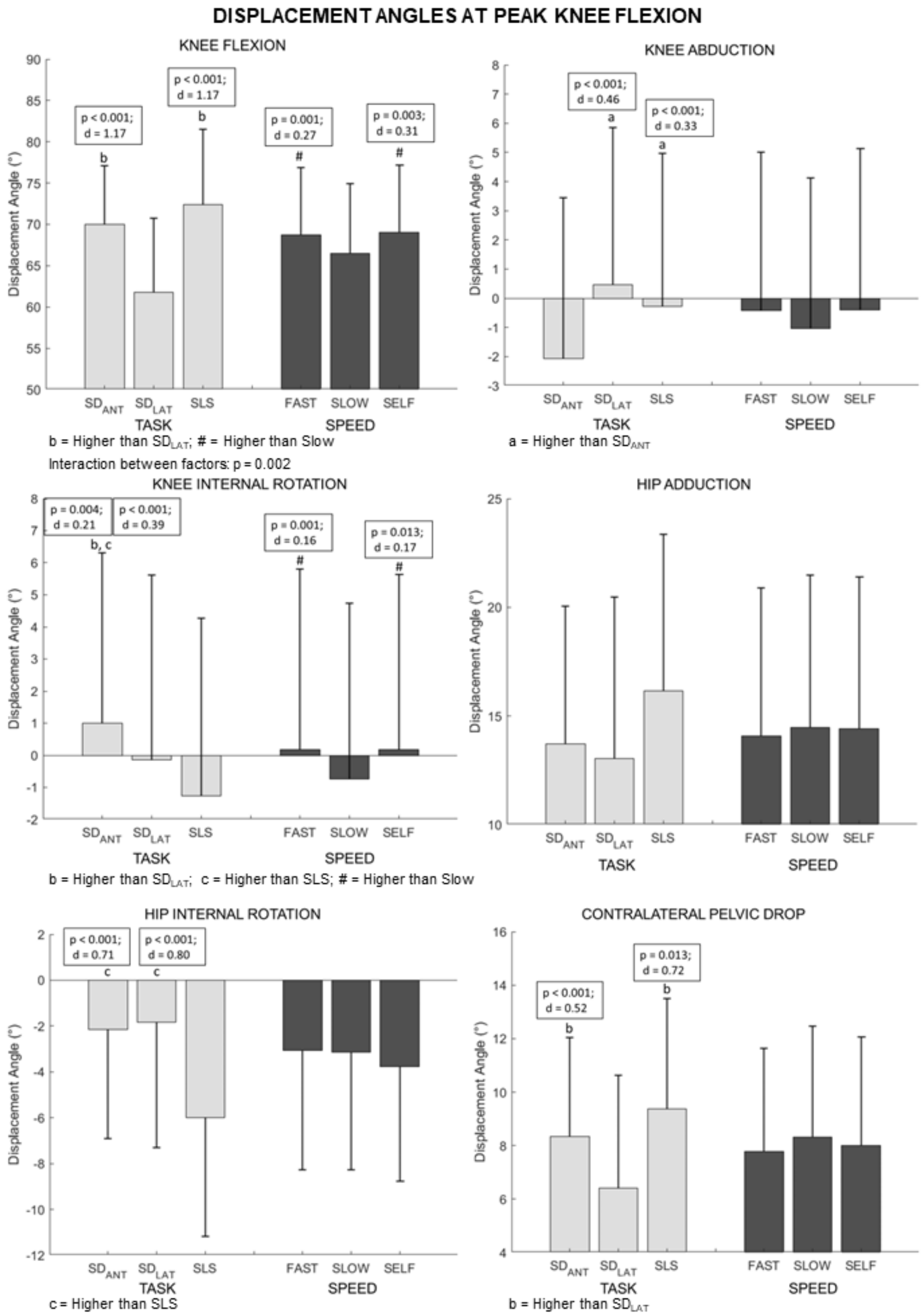

Figure 5. Displacement angle values for each variable of interest at peak knee flexion. Task values are presented as the average of each task across the three speeds. Speed values are presented as the average of each speed across the three tasks.

6. DISCUSSION

This study evaluated the effect of task type and movement speed on kinematic measurements associated with increased lower limb injury risk (Nakagawa et al. 2012; Räisänen et al. 2018; Schmidt et al. 2019; Gilmer et al. 2020; Harput et al. 2020). We found that task type influenced all metrics. At 60°, the lateral step-down presented greater frontal and transverse displacement at the hip but lower at the knee. The anterior step-down had the lowest frontal plane movement while the squat had the greatest knee abduction displacement. At peak knee flexion, values tended to increase and followed the same pattern for knee abduction and internal rotation and hip internal rotation. Speed affected primarily the knee angles, where the slow speed induced smaller displacements in the three planes.

The measurements evaluated in this study were chosen based on their association with injury (Nakagawa et al. 2012; Schmidt et al. 2019) or with increased load at the knee (Koga et al. 2010; Weiss and Whatman 2015). It has been shown that people with patellofemoral pain displayed greater hip adduction, knee abduction and contralateral pelvic drop (Nakagawa et al. 2012) and that increased hip adduction, hip internal rotation and knee abduction were predictors of pain in this population (Nakagawa et al. 2013). Increased hip adduction and contralateral pelvic drop has also been found in subjects with chronic hip joint pain compared to those with patellofemoral pain (Schmidt et al. 2019). In addition, increased knee valgus, which is a two-dimensional measurement that encompasses knee abduction, knee internal rotation, hip adduction and hip internal rotation, has been found in populations with reconstructed medial patellofemoral (Harput et al. 2020) and anterior cruciate ligaments (Gilmer et al. 2020). More specifically, excessive hip adduction and internal rotation move the knee joint center medially relative to the foot. As the foot is fixed to the floor, excessive frontal and transverse plane motions at the hip generate medial motion of the knee joint, increasing knee load (Powers 2003).

Slower squat speeds have been suggested to require increased balance than faster ones (Talarico et al. 2019). We hypothesized this would reflect in greater displacement values in the frontal and transverse planes during this condition. However, our hypothesis was not confirmed since the slow speed resulted in lower values for knee abduction and internal rotation, suggesting that faster speeds may be more appropriate when seeking to identify misalignments at the knee. A previous study that compared two kinematically similar tasks (single-leg squat and propulsion phase of a single-hop) found that the slower task presented greater adduction (Rabello et al. 2021). However, we did not find any influence of movement speed on hip adduction or pelvic drop, suggesting that the influence of movement speed is more relevant at the knee than in the proximal joints.

We evaluated kinematics at two moments during the squat in order to account for the influence of knee flexion on the frontal and transverse measurements, since SLS and SD_{ANT} were performed to a higher depth than SD_{LAT}. Knee abduction and internal rotation of the knee and hip followed similar patterns at both time points, while hip adduction and pelvic drop only showed differences at 60° and peak knee flexion, respectively. Besides depth, the position of the contralateral limb is the main difference between tasks. Therefore, the support limb and pelvic compensations required to maintain balance while descending with the contralateral limb positioned in the back, side or front (position in the

SLS, SD_{LAT} and SD_{ANT}, respectively) are better represented in the 60° values (Khuu et al. 2016).

Knee abduction and hip adduction are two of the most commonly assessed measurements during squats and step-downs (Donohue et al. 2015; Bell-Jenje et al. 2016; de Albuquerque et al. 2021). Knee abduction was smaller for SD_{ANT} and greater for the SLS, as previously reported (Lewis et al. 2015). However, the small absolute difference (< 3°) in the current study indicates that this statistically significant difference may not be clinically important. Meanwhile, hip adduction was only greater at 60° for SD_{LAT} than SD_{ANT}, likely due to a strategy to shift the center of mass toward the support limb in order to maintain balance, since during SD_{ANT} the participants could also adduct the contralateral limb to achieve this goal, which was not possible for SD_{LAT}.

Other commonly assessed measurements are contralateral pelvic drop and hip internal rotation (Nakagawa et al. 2012). Pelvic drop was lowest for SD_{LAT} at peak knee flexion, which is somewhat unexpected, since this action would also approximate the contralateral heel to the ground and make the task easier. However, since it required lower knee flexion (and therefore considered easier), subjects may not have used this strategy to accomplish the task. Finally, hip external rotation was higher (particularly at peak knee flexion) for SLS than the other tasks, a possible strategy to avoid medialization of the knee while maintaining balance.

The presented values cannot be fully compared to previous studies which compared variations of single-leg functional tasks such as squats and step-downs (Lewis et al. 2015; Khuu et al. 2016). Instead of reporting absolute values, we reported angular displacement from the initial position (in single-leg support) to 60° or peak knee flexion to eliminate the effect of anatomical characteristic and of being on single-leg support on outcomes. This measurement allowed us to understand how much motion occurs during task, and in particular what are the differences in motion between the tasks and speeds. When using functional tasks to assess kinematic measurements associated with injuries, researchers and clinicians should consider movement speed and task used, since most metrics were affected by these factors. The slow condition was where knee abduction and internal rotation were lowest, suggesting that faster speed may be more appropriate to detect abnormal movement patterns during these tasks. When knee abduction is the variable of interest, the SLS may be the best test since it elicits the greatest displacement, but when evaluating hip motion, the SD_{LAT} might be best. It is important to note, however, that these differences may not be clinically significant, since the absolute differences were of few degrees, particular for the speed comparisons.

To our knowledge, this was the first study in which movement speed was directly controlled in order to evaluate its effects on kinematics (Talarico et al. 2019). However, this study has limitations that should be mentioned: (1) we only evaluated healthy females, so the results may not apply for males or pathological populations; (2) although the measurements we evaluated have been used to differentiate between pathological and healthy populations, they were not prospectively associated with increased lower limb injury risk and; (3) only kinematics were evaluated, therefore we do not have information about the forces that caused the movements during the different conditions; (4) data were discretized in order to compare the results with the current literature, however, in the process of selecting only the peak knee flexion and 60° of knee flexion data points, data that could have been important was lost. Future studies are necessary to confirm that the found differences also apply to different populations and if different tasks or speeds are prospectively associated with greater injury risk.

7. CONCLUSION

Task type and movement speed can influence several metrics commonly used to assess movement kinematics, albeit with small absolute difference in degrees. During clinical and return-to-sport assessments, lack of standardization may significantly affect the interpretation of the resulting angles. Therefore, researchers and clinicians should consider this aspect when choosing the most appropriate task and the instruction to give during its execution. There isn't one task or speed that consistently displays higher or lower displacement angles. However, single leg squats and anterior step-downs at faster or self-selected speeds might be more appropriate when there is greater interest on the knee joint while the lateral step-down might be best when evaluating movement at the hip joint.

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9. SUPPLEMENTAL MATERIALS

Supplementary material 1 - P-values and effect sizes (Cohen's d) for pairwise comparisons within tasks and speeds.

CONDITION	TASK						SPEED						INTERACTION
	SD_{ANT} **vs** SD_{LAT}		SD_{ANT} **vs SLS**		SD_{LAT} **vs SLS**		**FAST vs SLOW**		**FAST vs SELF**		**SLOW vs SELF**		
	p	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	
Knee Flexion													
PKF	**<0.001**	**1.17**	p = 1.000		**<0.001**	**1.17**	0.001	**0.27**	p = 1.000		0.003	**0.31**	**0.002**
Knee Abduction													
60° KF	**<0.001**	0.28	**<0.001**	0.52	**0.002**	0.23	**0.007**	**0.16**	p = 1.000		p = 0.095		0.230
PKF	**<0.001**	0.46	**0.045**	0.33	p = 0.718		p = 0.151						0.184
Knee Internal rotation													
60° KF	**<0.001**	**0.32**	p = 0.140		p = 0.673		**<0.001**	**0.24**	p = 1.000		**<0.001**	**0.21**	0.955
PKF	**0.004**	**0.21**	**<0.001**	**0.39**	p = 0.162		**0.001**	**0.16**	p = 1.000		**0.013**	**0.17**	0.338
	p	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	
Hip Adduction													
60° KF	**<0.001**	**0.82**	p = 0.172		p = 0.142		p = 0.770						0.965
PKF	p = 0.059						p = 0.635						0.729
Internal rotation													
60° KF	p = 1.000		p = 0.193		**0.026**	**0.30**	**0.004**	**0.20**	p = 0.066		p = 1.000		0.236
PKF	p = 1.000		**<0.001**	**0.71**	**<0.001**	**0.80**	p = 0.189						0.322
	p	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	*p*	**d**	
Contralateral Pelvis drop													
60° KF	p = 0.067						p = 0.759						0.852
PKF	**<0.001**	**0.52**	p = 0.676		**0.013**	**0.72**	p = 0.203						0.307

ANOVA p-values are presented when no main effect was found and t-test pairwise p-values are presented when a main effect was found. "d" values are only presented for significant pairwise comparisons.

PKF = Peak knee flexion; SD_{ANT} = Anterior step-down; SD_{LAT} = Lateral step-down; SLS = Single-leg squat; SELF = Self-selected

Supplementary material 2 - Displacement angles for each task and speed.

CONDITION	SD_{ANT} Fast	SD_{ANT} Slow	SD_{ANT} Self	SD_{LAT} Fast	SD_{LAT} Slow	SD_{LAT} Self	SLS Fast	SLS Slow	SLS Self
Knee Flexion									
60° KF	-	-	-	-	-	-	-	-	-
PKF	71.3 ± 7.1$^{b,\#}$	68.4 ± 6.3^{b}	70.4 ± 7.3$^{b,\#}$	62.8 ± 7.4$^{\#}$	60.8 ± 7.0	61.7 ± 7.2$^{\#}$	72.0 ± 11.1$^{b,\#}$	70.3 ± 10.2^{b}	75.0 ± 10.7$^{b,\#}$
Knee Abduction									
60° KF	-0.8 ± 4.6$^{\#}$	-1.2 ± 4.8	-1.0 ± 4.7	0.5 ± 5.2$^{a,\#}$	-0.3 ± 4.5^{a}	0.7 ± 4.8^{a}	1.9 ± 4.6$^{a,b,\#}$	0.8 ± 4.4^{a,b}	1.5 ± 4.5^{a,b}
PKF	-2.1 ± 5.6	-2.2 ± 5.5	-1.9 ± 5.5	0.6 ± 5.7^{a}	-0.3 ± 5.0^{a}	1.1 ± 5.4^{a}	0.2 ± 5.1^{a}	-0.6 ± 4.9^{a}	-0.4 ± 5.7^{a}
Knee Internal rotation									
60° KF	2.8 ± 4.3$^{b,\#}$	1.6 ± 4.5^{b}	2.7 ± 4.7$^{b,\#}$	1.3 ± 4.8$^{\#}$	0.1 ± 4.6	1.2 ± 4.1$^{\#}$	1.8 ± 4.5$^{\#}$	0.9 ± 5.3	1.6 ± 5.2$^{\#}$
PKF	1.4 ± 5.9$^{b,c,\#}$	0.1 ± 5.49^{b,c}	1.5 ± 5.4$^{b,c,\#}$	0.1 ± 5.5$^{\#}$	-0.8 ± 5.2	0.2 ± 4.6$^{\#}$	-1.0 ± 5.6$^{\#}$	-1.5 ± 6.3	-1.3 ± 6.0$^{\#}$
Hip Adduction									
60° KF	6.4 ± 5.1	6.0 ± 4.9	6.0 ± 7.2	10.9 ± 5.9^{a}	10.5 ± 4.8^{a}	11.0 ± 5.9^{a}	8.8 ± 5.8	8.5 ± 5.0	8.5 ± 5.1
PKF	13.3 ± 6.0	14.1 ± 5.9	13.8 ± 7.7	13.2 ± 5.9	13.0 ± 6.3	13.0 ± 6.0	15.8 ± 8.7	16.3 ± 8.1	16.4 ± 7.8
Hip Internal rotation									
60° KF	-1.3 ± 4.1$^{\#}$	-2.0 ± 3.8	-1.6 ± 3.9	-0.9 ± 4.3$^{c,\#}$	-2.0 ± 4.0^{c}	-1.6 ± 3.9^{c}	-2.0 ± 5.2$^{\#}$	-2.9 ± 4.9	-3.7 ± 5.1
PKF	-2.2 ± 5.5^{c}	-2.0 ± 4.8^{c}	-2.2 ± 4.9^{c}	-1.4 ± 4.8^{c}	-2.0 ± 4.3^{c}	-2.1 ± 4.3^{c}	-5.6 ± 5.8	-5.4 ± 6.2	-7.0 ± 5.6
Contralateral Pelvic Drop									
60° KF	4.0 ± 2.8	4.1 ± 2.5	4.1 ± 3.5	5.2 ± 3.3	5.3 ± 2.8	5.1 ± 3.4	4.1 ± 2.9	4.5 ± 2.2	4.5 ± 2.8
PKF	8.1 ± 3.6^{b}	8.8 ± 3.4^{b}	8.1 ± 4.4^{b}	6.3 ± 3.6	6.8 ± 3.6	6.2 ± 3.5	9.0 ± 4.7^{b}	9.4 ± 4.1^{b}	9.8 ± 5.0^{b}

All values are presented as degrees of displacement (position at the event – position at initial single-leg stance).

a = Higher than SD_{ANT}; b = Higher than SD_{LAT}; c = Higher than SLS; # = Higher than Slow

PKF = Peak knee flexion; SD_{ANT} = Anterior step-down; SD_{LAT} = Lateral step-down; SLS = Single-leg squat; Self = Self-selected

STUDY 2 - DIFFERENT NEUROMUSCULAR METRICS ARE ASSOCIATED WITH KNEE ABDUCTION AND HIP ADDUCTION ANGLES DURING FUNCTIONAL TASKS

1. CONTEXT

This study also stems from the same project as the first study. It was written afterwards as the muscle activation data required a longer time to analyze, given that most of the MatLab code used for it was written from scratch.

This study focused on the interrelation between the metrics commonly used during functional tasks (both kinematics and muscle activation) and its possible effects on the interpretation of the current literature. This study deals with specific aim 3.

The study was submitted to the Journal of Electromyography and Kinesiology on July 22nd, 2023, and accepted for publication on October 13th, 2023. The following text differs from the publication in a few small ways in order to standardize the terminology and the structure. The study content is unchanged.

Rabello, R., Brunetti, C., Bertozzi, F., Rodrigues, R., & Sforza, C. (2023). Different neuromuscular parameters are associated with knee abduction and hip adduction angles during functional tasks. Journal of Electromyography and Kinesiology, 73(July), 102833. https://doi.org/10.1016/j.jelekin.2023.102833

2. ABSTRACT

Background: Knee abduction and hip adduction during functional tasks may indicate increased joint injury risk and discriminate between pathological and healthy people. Muscles' neuromuscular metrics such as amplitude (EMG_{AMP}) and onset (EMG_{ONSET}) have been used to explain kinematics. The study aimed to evaluate the correlation between two EMG metrics of seven trunk and lower limb muscles and 3D kinematics during two tasks.

Methods: Eighteen physically active women participated in the study. The following metrics were obtained during single-leg squat and anterior step-down: (i) EMG_{AMP} and EMG_{ONSET} of fibularis longus (FL), tibialis anterior (TA), vastus medialis (VM), biceps femoris (BF), gluteus medius (GMED), ipsilateral (OB_IL) and contralateral (OB_CL) external obliques and (ii) knee abduction and hip adduction angular displacement (initial angle – angle at 60° of knee flexion). Spearman's correlation coefficient was calculated between kinematic and EMG metrics.

Results: Greater knee abduction was correlated with delayed TA_{ONSET}, $GMED_{ONSET}$ and OB_IL_{ONSET} during step-down. Greater hip adduction was correlated with lower VM_{AMP}, BF_{AMP} and delayed VM_{ONSET} during step-down.

Conclusions: Although task-specific, these results suggest that EMG_{ONSET} may influence knee abduction, while both EMG_{ONSET} and EMG_{AMP} may affect hip adduction. The identification of muscle activation patterns in relation to kinematics may help the development of injury prevention and rehabilitation programs.

3. INTRODUCTION

Kinematics of functional tasks may indicate increased injury risk and discriminate between pathological and healthy people (Nakagawa et al. 2012; Räisänen et al. 2018). In particular, increased knee valgus has been observed in people that went on to have a knee injury (Räisänen et al. 2018) and in people with patellofemoral pain (Nakagawa et al. 2012). Moreover, individuals with chronic condition such as patellofemoral pain have also displayed increased hip adduction angles during single-leg squats (Nakagawa et al. 2012). Although these measurements are considered relevant for injury risk estimation, the neuromuscular factors that are associated with specific motion patterns are insufficiently studied in the literature.

The relationship between knee and hip kinematics and forces generated by the muscles are complex. While frontal plane hip movement is largely controlled by muscles that primarily act on the joint (e.g., gluteus medius and tensor fascia latae), there are no muscles whose primary function is to move the knee in the frontal plane given that it is a bicondylar joint (Neumann 2010). Despite that, knee abduction still occurs passively during functional tasks where the foot is in contact with the ground (Nakagawa et al. 2012). During these closed-chain tasks, the forces that generate this passive knee abduction are exerted by the ground reaction forces through the distal segments (foot and shank) and by the upper body mass through the proximal segments (trunk and hip) (Tiberio 1987; Powers 2010). Therefore, the association between frontal plane hip and knee angles and local (knee or hip), distal and proximal muscle activation can help us understand why these movements occur. In particular, we can hypothesize that the muscles that act as prime movers in frontal plane movement, such as the tibialis anterior and fibularis longus at the ankle, gluteus medius at the hip and external oblique at the trunk, could have a greater influence on frontal plane kinematics than those that primarily act on the other planes.

Single-leg squats and anterior step-downs are commonly used functional tasks due to their lower speed, greater safety and possibility of immediate visual feedback (Rabello et al. 2022), which are beneficial for the clinical evaluation of injuries such as patellofemoral pain, knee osteoarthritis and femoroacetabular impingement syndrome (Nakagawa et al. 2012; Cabral et al. 2021; Malloy et al. 2021). Other popular functional tasks are landing and cutting movements, which are more commonly used in the context of acute non-contact injuries such as the anterior cruciate ligament rupture (Hewett et al. 2005). Activation patterns of several muscles have been evaluated during the execution of these movements, albeit with a greater focus on the hip (Nakagawa et al. 2012; Hollman et al. 2014) and knee muscles (Hatfield et al. 2017; Mirzaie et al. 2019). Furthermore, different metrics of muscle activation have been evaluated, such as amplitudes (peak or mean) and onsets (Brindle et al. 2003; Neamatallah et al. 2020; Rodrigues et al. 2022c).

Because of its role in controlling movement, studies have sought to make inferences and hypothesis regarding joint kinematics (and consequently injury risk or mechanism) based on muscle activation results (Boudreau et al. 2009; Motealleh et al. 2015; Orozco-Chavez and Mendez-Rebolledo 2018; Mirzaie et al. 2019; Krause and Hollman 2020). However, the relationship between muscle activity and injury-related kinematics (such as knee abduction and hip adduction) has been investigated in few studies and only focused on activation amplitude and on the gluteal muscles (Hollman et al. 2009, 2014; Neamatallah et al. 2020). Concurrent evaluation of different metrics and of muscles acting on different joints should provide a more comprehensive picture of the activation strategies employed. Therefore, this study aimed at identifying the association between knee and hip frontal plane kinematics during the single-leg squat and the anterior step-down with the activation amplitude and onset of muscles acting on the trunk, hip, knee, and ankle joints.

4. METHODS

Participants

Twenty-two participants were recruited for a larger study in our laboratory, whose results are presented elsewhere (Rabello et al. 2022). Among those, muscle activation data were successfully obtained from 18 physically active women, who performed all evaluations in a single day. Sample size was calculated on G*Power (version 3.1.9.6; University of Trier, Trier, Germany), adopting and Effect size $|\rho|$ of 0.5, α of 0.05 and Power ($1-\beta$) of 0.75 (Hollman et al. 2009). Participants were included if they performed physical activity two or more times a week, were between 18 and 40 years old, had no pain or discomfort on the evaluation day and had no history of lower limb or back surgery. All participants signed an informed consent form prior to taking part in the study, which was approved by the university's ethics committee. The participants were free to stop the experiment at any moment.

Tasks and instrumentation

Participants performed five repetitions of two tasks while kinematics and muscle activation were recorded for the dominant limb, which was determined with the question "which foot would you use to kick a ball?" (Figure 1). For the single-leg squat, participants kept the contra lateral knee flexed at 90°, the thigh perpendicular to the ground and were instructed to squat until a comfortable depth before returning to the initial position. For the anterior step-down, participants were instructed to lightly touch the ground in front with their contralateral heel before returning to the initial position. This task was executed on a step with height between 15 and 17 cm according to the participants' height (i.e., 15 cm for participants shorter than 159 cm, 16 for those in between 160 and 175 cm and 17 cm for those taller than 175 cm). For all tasks, participants were instructed to keep their hands across their shoulders and movement speed was controlled using a metronome (three seconds eccentric and two seconds concentric). Prior to data collection, all participants performed as many trials as they required to familiarize themselves with the tasks and timing, resting as long as necessary to avoid fatigue effects.

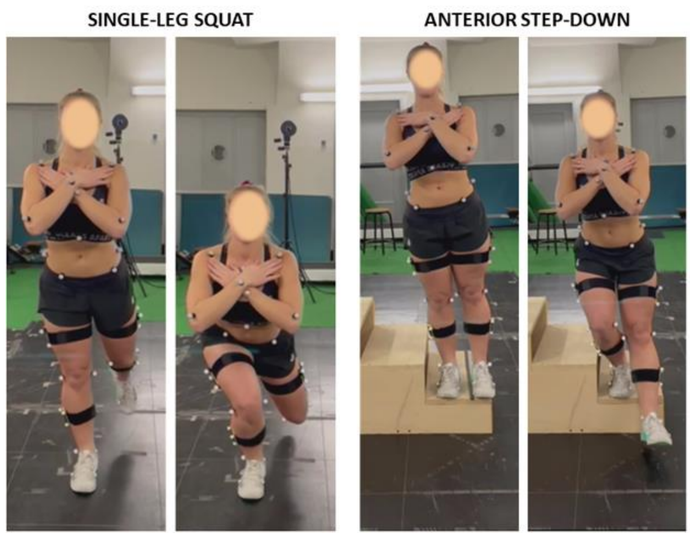

Figure 1. Execution of the Single-Leg Squat and the Anterior Step-Down.

Knee and hip kinematics were measured with a 9-camera 3D motion-capture system (60 Hz, BTS S.p.A, Garbagnate Milanese, Italy) using 38 passive retroreflective markers (Rabello et al. 2022). Muscle activation was estimated using surface electromyography (EMG) with 1020 Hz sampling rate, 16-bit resolution and differential amplifiers (bandwidth: 10–500 Hz) with common mode rejection ratio >110 dB at 50–60 Hz and input impedance >10 GΩ (FreeEMG, 300 BTS S.p.A, Garbagnate Milanese, Italy). Disk-shaped silver-silver chloride bipolar electrodes (diameter: 24 mm; interelectrode distance: 2 cm; Covidien, Dublin, Ireland) were positioned over the bellies of seven muscles and aligned with the fiber orientation: fibularis longus, tibialis anterior, vastus medialis, biceps femoris, gluteus medius, external obliques on the ipsilateral and contralateral sides. These muscles were selected because they represent primary movers in the frontal plane for the ankle, hip and trunk and the most important muscles acting on the knee joint. Skin preparation and electrode positioning were conducted according to the Surface ElectroMyoGraphy for the Non-invasive assessment of Muscles (SENIAM). Since they were unavailable on SENIAM, positioning on the oblique muscles followed previous literature (Rabello et al. 2022). Maximum voluntary isometric contractions were conducted for the normalization of average EMG data after appropriate warm-up. Isometric trunk flexion was performed (in neutral trunk position) for the oblique muscles, hip abduction (in 10° of hip abduction and 0° of hip and knee flexion) for gluteus medius, knee flexion (in 30° knee flexion and 0° hip flexion) for biceps femoris, knee extension (in 90° of knee and hip flexion) for vastus medialis, ankle dorsiflexion (in neutral ankle position) for tibialis anterior and rearfoot eversion (in 0° of eversion) for fibularis longus (Figure 2). Two trials were performed for each muscle with contractions lasting five seconds.

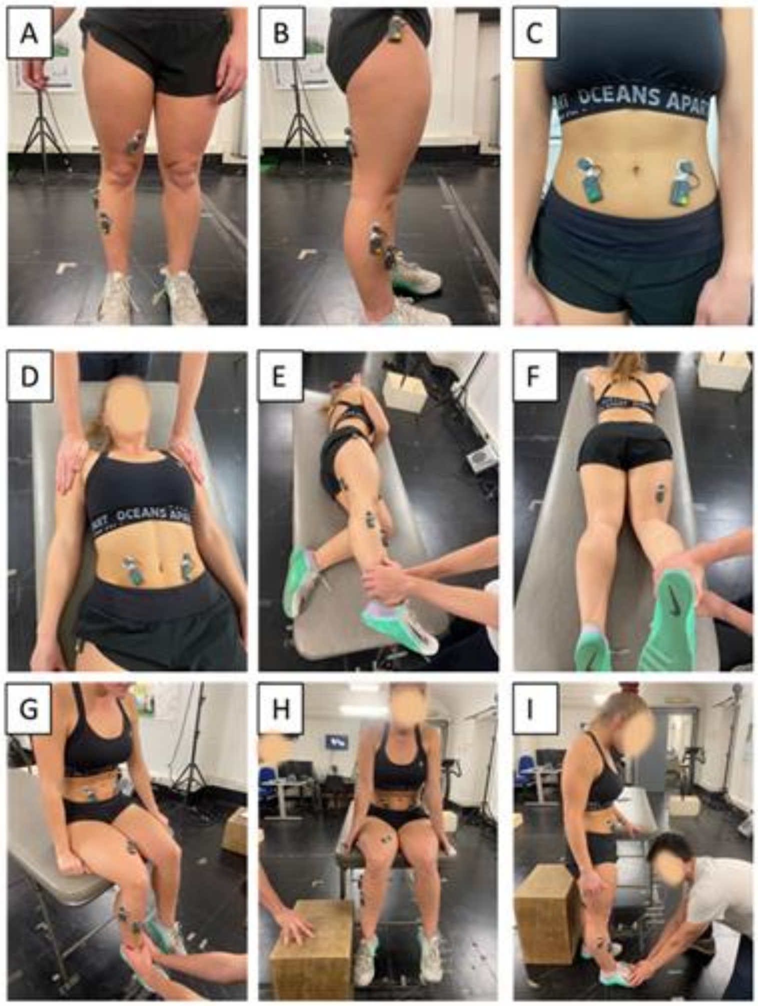

Figure 2. Positioning of the seven EMG probes (A-C). Maximum voluntary isometric contractions against manual resistance for the external obliques (D), gluteus medius (E), biceps femoris (F), vastus medialis (G), fibularis longus (H) and tibialis anterior (I) muscles.

Data processing

Kinematics

Markers' 3D positions were reconstructed using SMARTTracker (BTS S.p.A, Garbagnate Milanese, Italy) and used to calculate knee and hip displacement angles in the frontal plane with Visual3D (C-Motion, Inc., Germantown, USA). For each participant, a model was created based on a 10-s static recording and applied to the movement trials. Three-dimensional knee and hip angles were calculated with a Cardan X-Y-Z rotation sequence. The eccentric phases of the two tasks were evaluated from the start of the movement (when knee flexion begins) until 60° of knee flexion. Movement start was considered the frame in which the knee flexion curve started to increase from the baseline (visual determination). Knee abduction and hip adduction angles for repetitions two to five were extracted during single-leg quiet standing (recorded prior to the task) and at the instant of 60° of knee flexion. The difference in angle between the two moments was calculated and adopted as the angular displacement. Trial one was discarded as it was considered a final practice trial. A standardized 60° angle was chosen for all tasks because they are performed with different knee flexion ranges and we sought to eliminate its influence on the frontal plane angles (Rabello et al. 2022).

Electromyography

In addition to the maximum voluntary isometric contractions, two common muscle activation metrics were obtained for each muscle using the raw EMG data and the start and end events: amplitude average (EMG_{AMP}) and onset (EMG_{ONSET}). The values for each metric for trials two to five were recorded and averaged. Data were processed and exported using Visual3D and analyzed using custom-written MATLAB code (Version 2021b; Mathworks Inc., Natwick, USA.)

For the analysis of EMG_{AMP}, data were rectified, high-pass filtered (20 Hz, Butterworth 4th order), smoothed (500 ms root mean square) and time-normalized to 101 samples. EMG_{AMP} was the average activation value from the start until 60° of knee flexion and expressed as the percentage of the EMG activation during the maximum voluntary isometric contractions. The EMG signal was not usable for 0-17% of the trials, resulting in the exclusion of 0-4 participants from the analysis due to lack of data, depending on the muscle. Using rectified and filtered data (not smoothed nor time-normalized), the EMG_{ONSET} of each muscle was defined as the moment in which the EMG signal amplitude signal rose above three standard deviations from the baseline (a 200 ms interval before the start of the movement) and was maintained for at least 25 ms following the start of knee flexion (i.e., start of the movement). The onset is expressed in milliseconds. Out of the 144 trials collected, the percentage of trials that did not present an onset and the consequent number of participants that did not have at least one onset recorded for each task were 45% and 1 (fibularis longus), 36% and 1 (tibialis anterior), 21% and 3 (vastus medialis), 36% and 5 (biceps femoris), 49% and 3 (gluteus medius), 40% and 3 (ipsilateral oblique) and 33% and 0 (contralateral oblique), respectively.

Statistical analysis

Fourteen EMG (7 muscles x 2 metrics) and two kinematic (knee abduction and hip adduction) metrics were extracted for each of the two tasks. The Shapiro-Wilk test was conducted to verify data normality, finding that a large number of metrics presented a non-normal distribution. Therefore, in order to find the association between the kinematic and EMG metrics, the Spearman's correlation coefficient was calculated adopting 0.05 as the significance threshold. The correlation coefficients' magnitude were interpreted with Cohen's criterion: <0.1 = trivial, 0.1– 0.29 = small, 0.3–0.49 = moderate and >0.5 = large.

5. RESULTS

Figure 3 shows the recorded values for the EMG metrics of the seven muscles and the angular displacement in the frontal plane of knee and hip joints. Table 1 shows the correlations between each EMG and kinematic metrics.

Knee abduction

During the single-leg squat, no significant correlations were found between knee abduction and any EMG metric. During the anterior step-down, greater EMG_{ONSET} delay of three muscles were significantly correlated with greater knee abduction. Tibialis anterior EMG_{ONSET}, gluteus medius EMG_{ONSET} and ipsilateral oblique EMG_{ONSET} all presented large correlations ($\rho > .596$).

Hip adduction

During the single-leg squat, no significant correlations were found between any EMG metric and hip adduction. During the anterior step-down, greater hip adduction was significantly correlated with lower EMG_{AMP} and delayed EMG_{ONSET}. Lower biceps femoris EMG_{AMP} and vastus medialis EMG_{AMP} presented a large and moderate correlation with greater hip adduction, respectively. A greater delay on vastus medialis EMG_{ONSET} presented a moderate correlation with greater hip adduction.

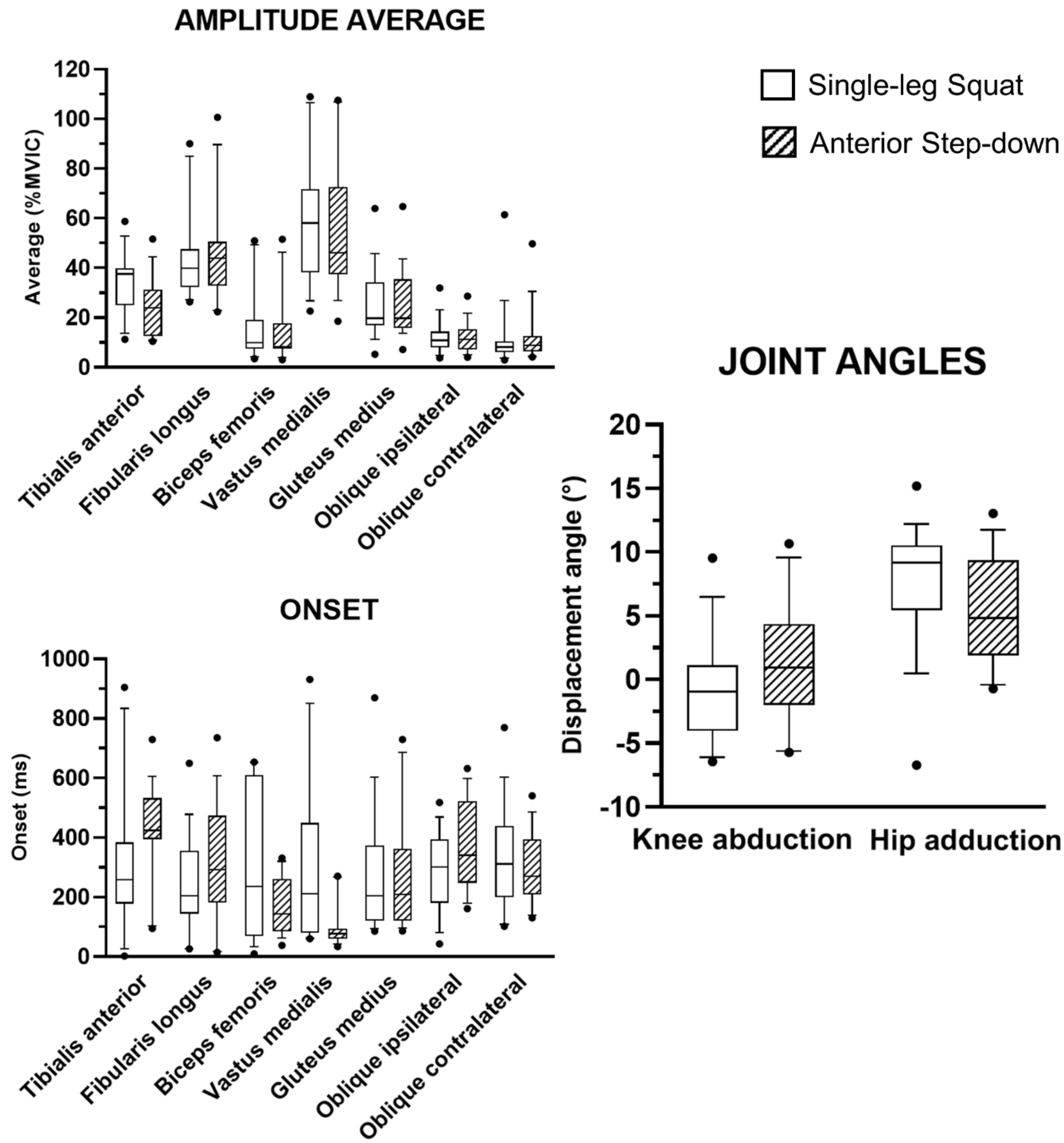

Figure 3. EMG and kinematic metrics. Boxplots show median, 10-90% range, maximum and minimal values.

Table 1. Spearman's p (p values) for the correlations between knee abduction, hip adduction and EMG metrics.

Knee Abduction	**Single-leg squat**		**Anterior step-down**	
Muscle	Average	Onset	Average	Onset
Tibialis anterior	-.182 (.533)	-.113 (.667)	-.275 (.321)	**.618 (.011)**[b]
Fibularis longus	.182 (.516)	-.176 (.498)	-.185 (.492)	.294 (.269)
Biceps femoris	.120 (.646)	.132 (.668)	.083 (.751)	.445 (.064)
Vastus medialis	.154 (.554)	.186 (.508)	.036 (.887)	.441 (.067)
Gluteus medius	.392 (.119)	-.432 (.108)	.401 (.099)	**.696 (.004)**[b]
Oblique ipsilateral	.146 (.603)	-.111 (.694)	.218 (.435)	**.596 (.019)**[b]
Oblique contralateral	-.098 (.699)	.013 (.958)	-.240 (.336)	.102 (.687)
Hip adduction	**Single-leg squat**		**Anterior step-down**	
Muscle	Average	Onset	Average	Onset
Tibialis anterior	-.459 (.098)	-.012 (.963)	.021 (.940)	-.159 (.557)
Fibularis longus	-.407 (.132)	-.267 (.300)	-.021 (.940)	.382 (.144)
Biceps femoris	-.265 (.305)	.297 (.325)	**-.667 (.003)**[b]	.245 (.328)
Vastus medialis	-.355 (.162)	-.421 (.118)	**-.490 (.039)**[a]	**.472 (.048)**[a]
Gluteus medius	.015 (.955)	-.239 (.390)	-.185 (.463)	.000 (1.00)
Oblique ipsilateral	-.082 (.771)	.096 (.732)	-.129 (.648)	-.150 (.594)
Oblique contralateral	-.152 (.548)	.311 (.210)	.129 (.610)	-.020 (.938)

Significant correlations are presented in bold. [a] = Moderate correlations; [b] = Large correlations, according to Cohen's criteria.

6. DISCUSSION

Although muscle activation measurement are commonly evaluated with the goal of understanding knee injury-related kinematic patterns, the relationships between these measurements are still unclear. Depending on the metric and on the task, different muscles presented moderate or large correlations with either knee abduction or hip adduction, with varied p-values. Given that there are few similar correlation studies (Hollman et al. 2009, 2014; Neamatallah et al. 2020), in the next sections of the discussion our findings will also be compared to studies that investigated differences between people with patellofemoral pain and controls and between poor and good performers, as these studies compare activation between a group that displays larger frontal plane kinematics values.

Distinct EMG metrics are employed in the literature given that they actively measure different aspects of muscle activation. For studies using squats and step-downs, the amplitude of the EMG signal is the most frequently used metric, usually reported with its peak or, even more commonly, with its average (Hatfield et al. 2017; Rabello et al. 2022). In our study, we found no association between knee abduction and average activation of any muscle in both tasks. These results agree with previous studies that found no differences in activation amplitude between controls and people that displayed medial knee displacement (Mauntel et al. 2013) or were bad performers (categorization that included excessive knee valgus) (Hollman et al. 2014). In addition, a recent meta-analysis (Rodrigues et al. 2022a) reported no differences in activation amplitude of the gluteus muscles in similar tasks between controls and people with patellofemoral pain, a pathology that has been previously associated with increased knee valgus (Nakagawa et al. 2012). Finally, three studies also reported no correlation between gluteus medius amplitude and knee valgus angles in women (Hollman et al. 2009, 2014; Neamatallah et al. 2020).

Meanwhile, decreased biceps femoris and vastus medialis amplitude were significantly correlated with higher hip adduction angles during the anterior step-down. Reduced activation of the two evaluated muscles that are primarily responsible for controlling knee sagittal movements may lead to decrease joint stability during the movement's descent phase by reducing joint control. This, in turn, could require increased hip adduction as a compensatory mechanism to maintain balance, explaining the relationship. These results were not replicated during the single-leg squat, likely due to the tasks' different characteristics (e.g., contralateral limb position) (Rabello et al. 2022). We did not find studies that evaluated these muscles in populations known for increased hip adduction during step-downs or the biceps femoris muscle during similar tasks. However, two studies evaluated the vastus medialis during the single-leg squat but did not present consistent results (Mirzaie et al. 2019; Rodrigues et al. 2022c). Mirzaie et al. (2019) found increased vastus medialis activation in healthy participants in comparison to patients with patellofemoral pain, while Rodrigues et al. (2022c) did not find any differences for the same population. Gluteus medius is the most commonly evaluated muscle in similar correlation studies. We did not find significant correlations with hip adduction, a result which is supported by Hollman et al. (2009, 2014) in two studies but oppose those found by Neamatallah et al. (2020), who reported a correlation between greater gluteus medius amplitude and greater hip adduction angles in healthy females.

The onset value gives information about the instant when the muscle “turns-on”. Although there are some discrepancies regarding the determination of onset, (Morey-Klapsing et al. 2004) the metric has been considered relevant for patellofemoral pain (Alsaleh et al. 2021) and anterior cruciate ligament rupture patients (Theisen et al. 2016) in meta-analysis containing various tasks. We found that delayed onset times of tibialis anterior, gluteus medius and ipsilateral oblique were associated with higher knee abduction angles only during the anterior step-down. These findings suggest that earlier activation of muscles that act on the frontal plane of the distal (ankle) and proximal segments (hip and trunk) may be helpful in preventing excessive frontal plane knee motion. This could occur because these muscles can act by controlling the center of mass and the base of support from an earlier time and consequently require less compensation by the knee joint. Although tibialis anterior and ipsilateral oblique onset have not been compared between people that specifically present greater knee abduction in similar tasks, our results for the gluteus medius are in line with Crossley et al. (2011), who found a delayed gluteus medius activation in poor performers during a step-down (originally called a single-leg squat in the study). However, a meta-analysis with patellofemoral pain patients did not find delayed activation of this muscle in similar tasks (Rodrigues et al. 2022a). Delayed vastus medialis activation was associated with increased hip adduction, possibly due to impairments in balance resulting from lower activation of this muscle that controls eccentric knee flexion at the very start of the movement. Greater instability at the knee joint could lead to excessive center of mass movement, which would require re-establishment of its position on top of the base of support that could be achieved by hip frontal plane movement. This result does not necessarily agree with previous patellofemoral pain studies that found no delayed vastus medialis onset in this group (Brindle et al. 2003; Earl et al. 2005), however, as our results indicate, this correlation is task-dependent and the aforementioned studies used stair-stepping or lateral step-down tasks.

Both the EMG and kinematic data presented substantial variability between participants; however, they are in line with what is found in the literature (Earl et al. 2005; Nakagawa et al. 2015; Hatfield et al. 2017; Han et al. 2018; Orozco-Chavez and Mendez-Rebolledo 2018; Mirzaie et al. 2019). EMG_{AMP} is highly dependent on the normalization task (Burden 2010), but similar studies have also found, for example, gluteus medius activation close to 20% (Han et al. 2018; Mirzaie et al. 2019), vastus medialis close to 50% (Hatfield et al. 2017) and external obliques close to 15% (Nakagawa et al. 2015) of MVIC in similar tasks. Our activity onset values are more difficult to compare with the current literature, as only few studies actually report their values and are limited to the vastus medialis, gluteus medius and biceps femoris (Earl et al. 2005; Han et al. 2018; Orozco-Chavez and Mendez-Rebolledo 2018). Mean onset values ranged from 103 ms to 417 ms on average depending on the muscle and task involved and from 2 ms to 931 ms individually recorded values, which is in line with the literature that often presents very high standard deviation values (Earl et al. 2005; Han et al. 2018; Orozco-Chavez and Mendez-Rebolledo 2018). Finally, our knee abduction and hip adduction values ranged from -6° to 11° and from -7° to 16°, respectively. Although higher values are proposed to increase the strain on the knee joint, there is no cut-off value that puts someone in greater risk of injury, making it difficult to determine if a certain activation amplitude or onset may suggest that someone is likely to get injured.

This study showed that some muscle activation metrics from proximal, distal and local joints (i.e., hip, ankle and knee when referring to the knee joint) are associated with injury-related kinematics. However, for the correct usage and interpretation of EMG measurements in the context of injuries, it is important to take into account the limitations of each metric. Although EMG amplitude provides a quantitative measure of how active a muscle is (typically in relation to a maximal contraction), it does not represent the amount of torque that is being generated given that torque is also dependent on muscle architecture and joint angles (Hug et al. 2015). For this reason, lower and higher amplitudes can be interpreted as lack of neural drive leading to lower torque generation or as increased neural drive due to a compensatory strategy for reduced torque production capacity, respectively (Rodrigues et al. 2022a). Activation onset is also influenced by the criteria adopted to consider a muscle to be activated (e.g., 2, 3 or 5 standard deviations above a resting activation) (Rodrigues et al. 2022a) and can often not be identified in some trials (from 21% to 49% of trials in this study). It would be good practice for future studies to report the percentage of trials where the onset was actually present. Another common barrier is the muscle crosstalk can influence the results, as the activity is registered under skin electrodes that are not exclusively targeting the muscle of interest (Konrad 2005). Finally, both EMG and kinematic data provide high-frequency signals (typically above 500Hz and 60Hz, respectively), which are often reduced to a single number to represent it, losing possibly important data in the process (Pataky 2010b). Nonetheless, EMG remains essential for providing insights into how the neuromuscular system controls kinematics. Therefore, within the context of their limitations, specific measurements can help in understanding the presence of injury-related patterns in certain populations and contribute to the development of evaluation tools and rehabilitation and prevention protocols. For example, adopting earlier activation of distal and proximal muscles may lead to reduced knee abduction angles and greater and earlier activation of the knee extensors as well as greater activation of the knee flexors may lead to reduced hip adduction angles due to improved stability.

In this study, we evaluated two commonly used muscle activation metrics in muscles acting on distal, proximal, and local joints to the knee and hip. Along with the use of two different functional tasks, the multi-joint consideration is a strength of the study. There are some limitations that must be mentioned: (i) due to its sensitive position, we did not evaluate the gluteus maximus muscle, which has been proposed to influence knee valgus (Hollman et al. 2009, 2014). (ii) due to the limited number of available probes, we had to choose only seven out of the several muscles that act on the ankle, knee, hip and trunk. We decided to evaluate one prime mover of each joints' frontal plane movement plus the biceps femoris and the vastus medialis, which are two of the most important muscles that act on the knee joint. Other muscles acting on the other planes and different frontal plane prime movers might have displayed different associations; (iii) because of equipment limitations, MVICs were performed against manual resistance. Although similar techniques are used in the literature it is possible that participants did not achieve their maximal activation due to the instability of the resistance (Chamorro et al. 2017; Lyons et al. 2017). (iv) despite careful preparation and probe positioning, some trials from a few subjects were not correctly recorded and had to be excluded (v) we only evaluated women given that these tasks are frequently used in the context of patellofemoral pain and anterior cruciate ligament injuries, both of which affect women at a higher rate (Boling et al. 2010; Stanley et al. 2016). Therefore, care should be taken when applying our findings to different populations; (vi) because we sought to verify the relationship between kinematics with muscles that cross different joints, in two different metrics and in two different tasks, we ran a large number of correlations. Multiple comparisons might elevate the risk of type I error, particularly with reduced sample sizes, however, we chose not to adjust the significance level in order to avoid type II error. Another reason for not making this adjustment is that similar studies with multiple correlations also chose not to do so (Hollman et al. 2009; Neamatallah et al. 2020), thus, our choice was also made in order to facilitate comparisons. However, as there is a risk for type I error, we encourage the readers to consider the sample size, ρ and p-values when observing our data; (vii) we chose to analyze the data at 60° of knee flexion in order to standardize the two tasks and facilitate literature comparison, however, results could have been different if we had selected a different joint angle or had analyzed the whole time-series. Finally, there are other EMG metrics that can be used to describe muscle activity that were not evaluated in this study: activation duration was not possible to calculate due to the difficulty in identifying an offset (i.e., when the muscle turns-off) in a high number of trials due to the characteristics of the chosen tasks (Rodrigues et al. 2022a) and metrics in the frequency domain, such as median frequency and spectral analysis, were not evaluated due to issues regarding their validity and interpretation (Farina et al. 2004; Beaulieu et al. 2008; Enoka 2008), although it has also been used in studies with similar tasks (Leporace et al. 2011; Rodrigues et al. 2022b).

7. CONCLUSION

We found that earlier onset of tibialis anterior, gluteus medius and ipsilateral external oblique were correlated with reduced knee abduction angles and that increased vastus medialis and biceps femoris activation amplitude and earlier vastus medialis onset correlated with increased hip adduction angles during the anterior step-down. The same results were not replicated during a single-leg squat, indicating a task-dependency effect. These findings provide insights into the relationship between muscle activation metrics and kinematics during functional task, allowing for the development of evidence-based hypothesis and inferences regarding injury-related outcomes, ultimately being helpful in establishing evaluation tools and rehabilitation and prevention programs.

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STUDY 3 - RUNNING STIFFNESS AND SPATIOTEMPORAL PARAMETERS ARE SIMILAR BETWEEN NON-RUNNERS AND RUNNERS WITH DIFFERENT EXPERIENCE LEVELS

1. CONTEXT

In the beginning of the second year during the PhD, I conducted a visiting period at Indiana University Bloomington, USA, under the supervision of Dr. Allison Gruber. The goal of the visit was to study running biomechanics, particularly as it relates to injuries, as it is by far the most studied topic in movement biomechanics. This visiting period went on from January 2022 until June 2022.

During this period, I focused on the effects of running experience on running mechanics and developed a project that would allow me to study it for myself, in both practice and theory. Given the short time to start a study, instead of developing a protocol from scratch, we decided to adopt the protocol that was being used by Dr. Gauri Desai, a PhD of Dr. Gruber's. In her study, she was studying the effects of running experience, age and body composition on shock absorption during running. However, she only had an experienced and a novice group. Therefore, because we also valued the importance of having truly inexperienced participants when comparing experience levels, we decided to add and collect this group during my visit.

This study deals with aim 4 of the present thesis. The focus on running stiffness as a measurement was to highlight the importance of choice of metric when evaluating biomechanics during functional tasks. In addition, the group comparison also deals with the group classification issue that can affect studies where comparisons are made.

A modified version of the following text was submitted for publication at the Journal of Sports Sciences on December 11th, 2023 and is currently under review.

2. ABSTRACT

Background: Spatiotemporal parameters and leg and joint stiffness are measurements that represent the fundamental dynamics of running. Therefore, these measurements may effectively differentiate the gait of less- from more-experienced runners, possibly addressing differing injury rates between populations. This study compared stiffness and spatiotemporal parameters between runners with different experience levels, including a group with no previous running experience.

Methods: Healthy physically active participants (mean±1 standard deviation: age= 22.1±3.6y) were divided into three groups, according to running experience: experienced (running>1-year, 14-48 km/week; n=23, 9 female), novice (running<1-year, 5-21 km/week; n=15, 4 female) and non-runners (no running for the past 5 years; n = 17, 7 female). Three-dimensional motion capture and force plates measured gait mechanics during overground running at 3.35 m/s. Knee, ankle, and three-dimensional leg stiffness, contact time, flight time, and step length were calculated and compared between groups using an independent-measures ANCOVA (covariate = sex).

Results: None of the biomechanical dependent variables were significantly different between groups (leg: p=0.652, Hedges' g=0.09-0.17; ankle: p=0.439, g=0.07-0.19; knee: p=0.153, g=0.13-0.29; contact time: p=0.592, g=0.06-0.24; flight time: p=0.513, g=0.03-0.40; step length: p=0.107, g=0.26-0.61).

Conclusions: Stiffness and spatiotemporal measurements are not influenced by a runner's experience level. These results suggest that running is a fundamental motor pattern that is not influenced overall by motor practice. Therefore, running gait may not differentially affect injury rates between experience levels.

3. INTRODUCTION

Running related injuries are one of the main reasons why people stop running, limiting its positive health benefits (Fokkema et al. 2019). Less-experienced runners incur 17.8 injuries per 1000 hours of running compared with 7.7 injuries among more-experienced runners (Videbæk et al. 2015). There could be two main justifications for this discrepancy. One is that less-experienced runners have not obtained the beneficial musculoskeletal adaptations related to training due to their limited lifetime exposure to running. Therefore, the tissues of less-experienced runners may be more susceptible to injury due to having low resistance to mechanical stress and strain (Agresta et al. 2018). The second possibility is that less-experienced runners may utilize running gait patterns that increase their injury risk compared with more-experienced runners. That is, less experienced runners could have a propensity to run with "poor" mechanics that increases mechanical stress on musculoskeletal tissues, thereby increasing the risk for injuries. Identifying biomechanical differences between experience levels would allow researchers, clinicians, and athletic trainers to target those gait mechanics to reduce injury incidence in less-experienced runners.

Previous studies comparing the biomechanics of experienced runners included novice runners who possess some, albeit limited, running experience (Maas et al. 2018; Agresta et al. 2019). These studies are inconclusive regarding group differences (Schmitz et al. 2014; Boey et al. 2017; Gómez-Molina et al. 2017; Agresta et al. 2018, 2019; Maas et al. 2018; Mo et al. 2020). Inconsistent definitions of running experience may contribute to conflicting findings because these studies use different measures of running exposure (e.g., years running, lifetime mileage) and different thresholds for these exposures to define experience groups. Although one study included a truly inexperienced group (i.e., no running experience) (Schmitz et al. 2014), most studies comparing experience groups define novice runners as having up to 6–24 months of running exposure (Agresta et al. 2019; Mo and Chow 2019; Mo et al. 2020). However, biomechanical adaptations due to motor learning may occur within the first 10-weeks of training (Moore et al. 2012). Therefore, in previous studies comparing gait between experience levels, the novice and less-experienced runners included may have already developed similar gait patterns as their more-experienced counterparts. Given that running is an inherent gait mode in which all individuals have the motor program to perform, evaluating the truly inexperienced is necessary to understand the influence of experience on running biomechanics.

Running kinematic and kinetic measurements (e.g., joint angles, joint moments, and ground reaction forces) have been compared across running experience levels. However, these measurements, when examined in isolation, provide limited information about a person's running mechanics because human locomotion requires a complex interaction of multiple components of the musculoskeletal system. One measurement that represents the integration of the musculoskeletal system during locomotion is stiffness, specifically stiffness of the leg and lower-extremity joints. Stiffness regulates how the musculoskeletal tissues engage with the external environment during the stance phase of gait (McMahon and Cheng 1990; Farley and González 1996; Ferris and Farley 1997) and provides insight into how the body absorbs the forces experienced during stance (Butler et al. 2003). Stiffness also influences spatiotemporal metrics, which are considered the final output of gait (Ferris et al. 1998; Agresta et al. 2019). Spatiotemporal metrics have been compared between runners with different experience, albeit with inconsistent findings (Gómez-Molina et al. 2017; Agresta et al. 2018), however, the influence of experience level on stiffness has yet to be studied.

Leg and joint stiffness calculations represent “quasi-stiffness” as they do not reflect the individual stiffnesses of each tissue (tendons, ligaments, bone, cartilage and muscles) (Zatsiorsky and Latash 1993). Leg stiffness models the lower limb as a linear “spring” and considers the change in leg length during stance due to peak ground reaction force (Butler et al. 2003). Given that human movement occurs through a combination of joint rotations, leg stiffness depends on the stiffness of lower-extremity joints. Joint stiffness is an angular metric that refers to the ratio of the change in joint moment to the change in joint angle. Ankle and knee joint stiffness are evaluated most because these joints undergo flexion after initial contact, representing spring compression. The hip may not be a valid measure of joint stiffness because it does not reliably flex after contact in all runners. Because stiffness represents the complex interaction of different components of the musculoskeletal system with the environment and dictates the final output of gait (McMahon and Cheng 1990; Farley and González 1996; Ferris and Farley 1997), stiffness measurements may be the best choice to evaluate whether the gait pattern between runners with varying experience levels is inherently different. Identifying whether running gait is learned or an ingrained component of the motor cortex is required to support (or reject) the hypothesis that group-based differences in running mechanics underlie their differing rates of running-related injury. Therefore, this study aimed to compare leg, ankle, and knee stiffness between three running experience levels (experienced, novices and non-runners). Additionally, spatiotemporal metrics were examined because they are the outcome of multi-joint motions and dictated by leg and joint stiffness. We hypothesized that these stiffness dependent variables would differ between groups and that the differences would be more distinct between the experienced and non-runners groups.

4. METHODS

Experimental Overview

After signing the informed consent form, participants changed into form-fitting clothes and shoes provided by the researchers (One X CrossFit Cushion 3.0; Reebok) to standardize the effects of footwear on leg and joint stiffness. Participants then warmed up by running at a self-selected speed on an indoor track or treadmill for approximately five minutes. Lower-limb markers were applied following published standards (Hamill et al. 2014b). Afterwards, a standing calibration was performed and then participants performed the running trials. Spatiotemporal metrics (step length, flight time, and contact time) and knee, ankle, and leg stiffness were calculated and compared between groups.

Participants

A sample size estimate for a fixed effects, omnibus, one-way ANOVA determined that 14 participants per group (total 42) would be required to obtain large effects (f=0.5) with 0.80 power and significance at 0.05 (G*Power, Version 3.1.9.7, University of Trier, Germany). Fifty-five participants engaging in regular physical activity and with different levels of running experience were recruited. Participants were recruited primarily through flyers and class announcements at Indiana University Bloomington. All participants were free from injury or were completing usual physical activity for at least eight weeks if an injury was sustained within the 6-months prior to data collection. Participants were excluded if they underwent previous lower-back or lower-limb surgery, had a history of chronic physiological or neurological ailments, or were unable to exercise at a low intensity (slow running) for at least 15 minutes. Approval from the University's ethics committee was obtained, and all participants gave written informed consent before participating.

Participants were divided into three groups based on the number of total years spent running and their average weekly distance in the past three years. The experienced group included participants running for more than one year and ran between 14–48 km/week. The novice group included participants running for less than one year and ran between 5–21 km/week. The non-runner group included participants that had not run consistently within the past five years but took part in physical activity at least twice per week. “Consistent” running was defined as performing a running session more than once a week for at least three consecutive weeks in the past six months, and/or more than four running sessions in a month for two consecutive months in the past five years. A running session was defined as “any time in which you ran exclusively for the purpose of going for a run, either outside or on a treadmill, for the purposes of running for at least 5-minutes and/or 1-mile or longer, and/or run for the purposes of recreation, fitness, pleasure, and/or socially”.

Instrumentation

Forty reflective markers were applied to the participants’ lower limbs (Hamill et al. 2014b). Individual markers were positioned bilaterally on the skin or on the shoe over the following anatomical landmarks: first distal phalanges, first and fifth meta-tarsals, top of the foot (approximately on top of the intermediate cuneiform bone), medial and lateral malleoli, femoral epicondyles, greater trochanters, anterior and posterior-superior iliac spines, and iliac crests. Additionally, rigid clusters of non-colinear markers were positioned bilaterally on the thigh, shanks, and heel. After the standing calibration, the markers on the metatarsals, malleoli, and femoral epicondyles were removed. Kinematic data were collected using a 13-camera motion capture system (240 Hz; Oqus 400, 500; Qualisys AB, Gothenburg, Sweden) synchronized with three force plates (1200 Hz; OR-6-2000, OR-7-1000; AMTI Inc, Watertown, USA) to measure center of pressure and ground reaction forces.

Running protocol

Participants warmed up by running at a self-selected speed on an indoor track or treadmill for approximately five minutes. Non-runners used the track because they were unsure of which speed to select on the treadmill. Participants were instructed to run overground along the laboratory's 18-meter runway at a predetermined speed of 3.35 $m \cdot s^{-1}$ (±5%), which was monitored using two timing gates positioned 6-meter apart and at the center of the runway. The speed was standardized across participants to ensure participants performed the same mechanical task and so between-group differences would be due to inherent gait mechanics, rather than running speed. For the trial to be considered valid, participants had to make full contact with their foot on at least one force plate, make no modification to their gait to target the force plates, and run within 5% of the predetermined speed. The researchers visually assessed participants' gait during all trials to ensure no targeting occurred. A maximum of 60 trials were performed to limit fatigue in the non-runners group. A minimum of 6 and maximum of 20 valid trials were recorded (i.e., 3 – 10 trials per limb). Given the relatively low between-limb asymmetry of leg stiffness in a healthy population (Pappas et al. 2015), valid trials in which either the right or the left foot contacted a force plate were retained for analysis. A sensitivity analysis was performed to confirm that collapsing the data across limbs did not influence the results (Supplemental Material 1).

Data analysis

Reflective markers were tracked with Qualisys Track Manager then gait data were calculated with Visual3D (C-Motion, Germantown, United States) using the hybrid model for the limbs and the composite pelvis model to estimate hip joint center locations. Raw marker and force plate data were low-pass filtered at 12Hz for the calculation of knee and ankle joint angles and moments to remove artifacts due to foot-ground impact (Kristianslund et al. 2012; Derrick et al. 2020). Knee and ankle joint moments were calculated using the proximal segment as the reference segment and normalized by body weight. Ground reaction force (GRF), center of pressure position, knee and ankle angles and moments from initial contact (vertical GRF ≥ 20N) to toe-off (vertical GRF ≤ 20N) were normalized to 101 datapoints then exported to calculate joint stiffness. A custom Visual3D algorithm was used to detect the instant of initial contact and toe-off events from the stance phase on the force plate to the subsequent stance phase of the same limb. Accuracy of the gait events were verified by checking the three-dimensional video recording and were used to calculate spatiotemporal metrics.

Three-dimensional leg stiffness ($kN \cdot m^{-1}$) was calculated using the multiplanar method (Liew et al. 2017; Kuzmeski et al. 2021):

$$3D\ Leg\ Stiffness = \frac{Fmax - 3D}{\Delta L3D}$$

Fmax-3D is the resultant force vector, calculated from the three GRF components then projected in alignment with the leg. ΔL3D is the change is leg length considering all three planes, from initial contact to the instant of maximum resultant force (Liew et al. 2017; Kuzmeski et al. 2021).

Both ankle and knee angular stiffness ($N \cdot m \cdot deg^{-1}$) were calculated as the slope of the best fit line in a joint angle – moment plot from initial contact until the end of the absorption phase (Figure 1) (Hamill et al. 2014a). In order for the knee and ankle joint stiffness to be calculated during the same portion of stance, the end of the absorption phase was defined as the maximum ankle dorsiflexion angle for both joints (Hamill et al. 2009, 2014a; Gruber et al. 2021).

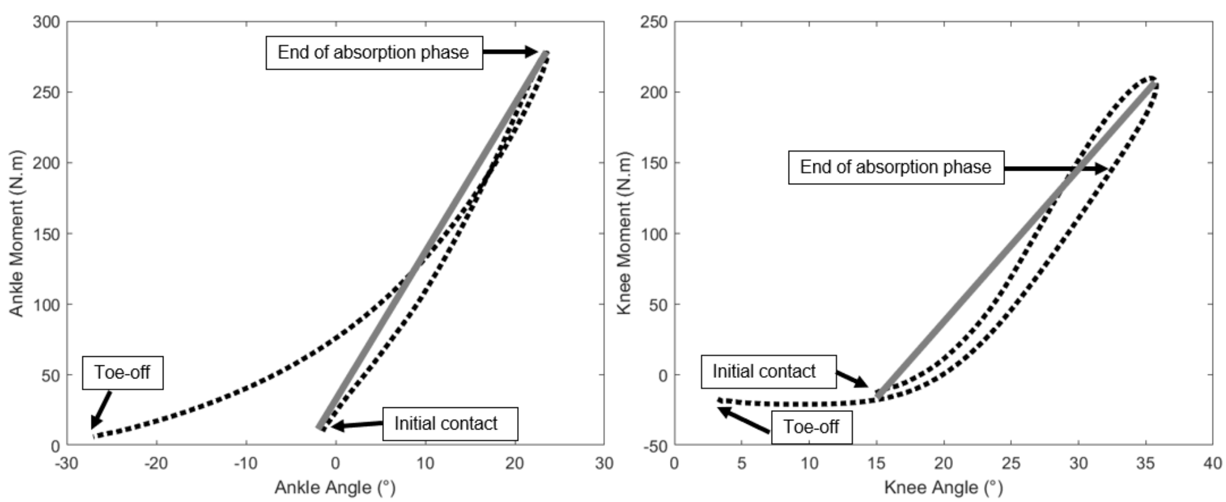

Figure 1. Example of the joint moment against joint angle plots used to calculate the joint stiffness. The slope of the line from the moment of initial contact until the end of the absorption phase (i.e., maximum ankle dorsiflexion for both joints) was considered the joint stiffness.

Flight time was defined as the duration from toe-off to the instant of initial contact with the subsequent foot. Step length was calculated by the distance between the antero-posterior position of the distal heel marker at toe-off and the position of the contralateral limb's heel marker at initial contact of the following step. Contact time was calculated as the time interval from the initial contact to toe-off for each stance phase.

Each metric was averaged across all trials, including both the right and left sides, for each participant. All calculations were conducted using custom-written MATLAB program (Version R2021b; MathWorks Inc, United States).

Statistical analysis

All dependent variables were normally distributed, as assessed by the Shapiro-Wilk test and Q-Q plots. Independent measures analysis of variance was used to compare anthropometrics between groups. An independent measures analysis of covariance (ANCOVA) and pairwise Hedges' *g* effect sizes (Lakens 2013) was performed to compare the ankle, knee and leg stiffness, flight time, contact time, and step length between the three groups (experienced, novice, non-runner) while controlling for sex as the covariate. Sex was included as a covariate given that the distribution of males and females varied between groups and running stiffness may differ between sexes (Brown et al. 2021). When significant differences between experience groups were observed, pairwise comparisons with a Bonferroni correction were further calculated.

5. RESULTS

Anthropometric characteristics were similar between the groups (Table 1). The number of valid trials recorded were (mean ± 1 standard deviation) 21 ± 4.7, 20.4 ± 3.7, and 17.6 ± 5.7 for the experienced, novice, and non-runner groups, respectively.

Table 1. Participant characteristics of the three experience groups (mean ± 1 standard deviation).

Group	Experienced	Novice	Non-runner	p-value
N (Females)	23 (9)	15 (4)	17 (7)	
Age (years)	22.3 ± 3.8	22.6 ± 4.0	21.4 ± 2.8	.580
Height (m)	1.72 ± 0.09	1.75 ± 0.06	1.73 ± 0.07	.412
Weight (kg)	68.2 ± 9.7	72.5 kg ± 9.0	75.4 ± 9.7	.065
Weekly distance (km)	19.4 ± 10.7	15.2 ± 5.0		.164
Experience (years)	7.4 ± 5.0	0.6 ± 0.4		<0.001

The Non-runner group included five participants taking part in activities that require intermittent running: basketball (2-10 hours/week, four participants), soccer (2 hours/week, one participant, American football (2 hours/week, one participant), and ultimate frisbee (1 hour/week, one participant).

There were no statistical differences between the three groups for leg, ankle, or knee stiffness ($p \geq 0.382$; Figure 2). Pairwise Hedges' g effect sizes between groups ranged from 0.09–0.17 for leg stiffness, 0.07–0.19 for ankle stiffness, and 0.13–0.29 for knee stiffness. There were no statistical differences for contact time, flight time, or step length ($p \geq 0.107$; Figure 3). Pairwise Hedges' g effect sizes between groups ranged from 0.06–0.24 for contact time, 0.03–0.40 for flight time, and 0.26– 0.61 for step length.

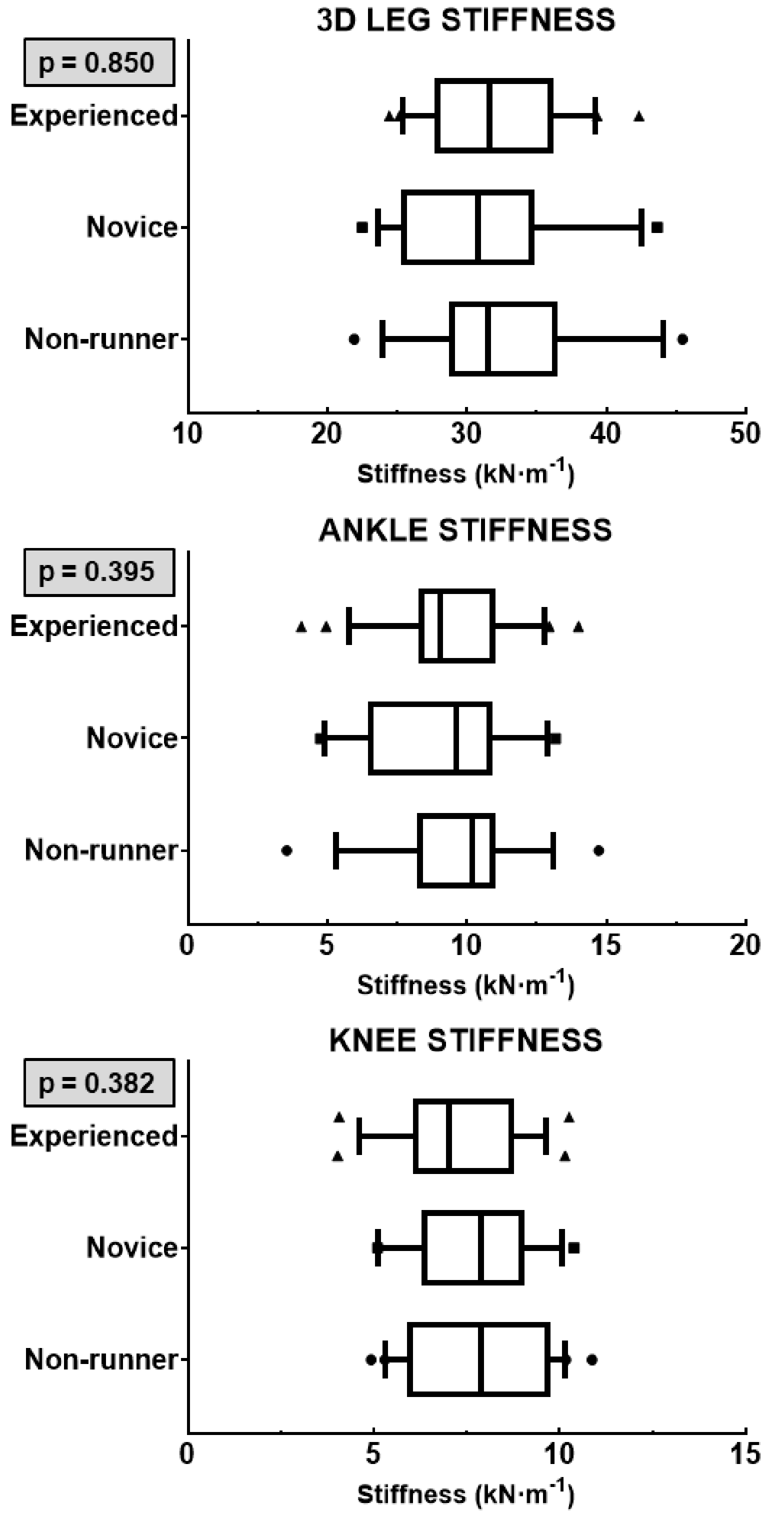

Figure 2. Leg, ankle, and knee stiffness among the three groups. Boxplots show values for 10^{th}, 25^{th}, 50^{th}, 75^{th} and 90^{th} percentile and symbols represent participant means outside the 10th – 90th range with triangles indicating experienced runners, squares indicating novice runners, and circles representing non-runners.

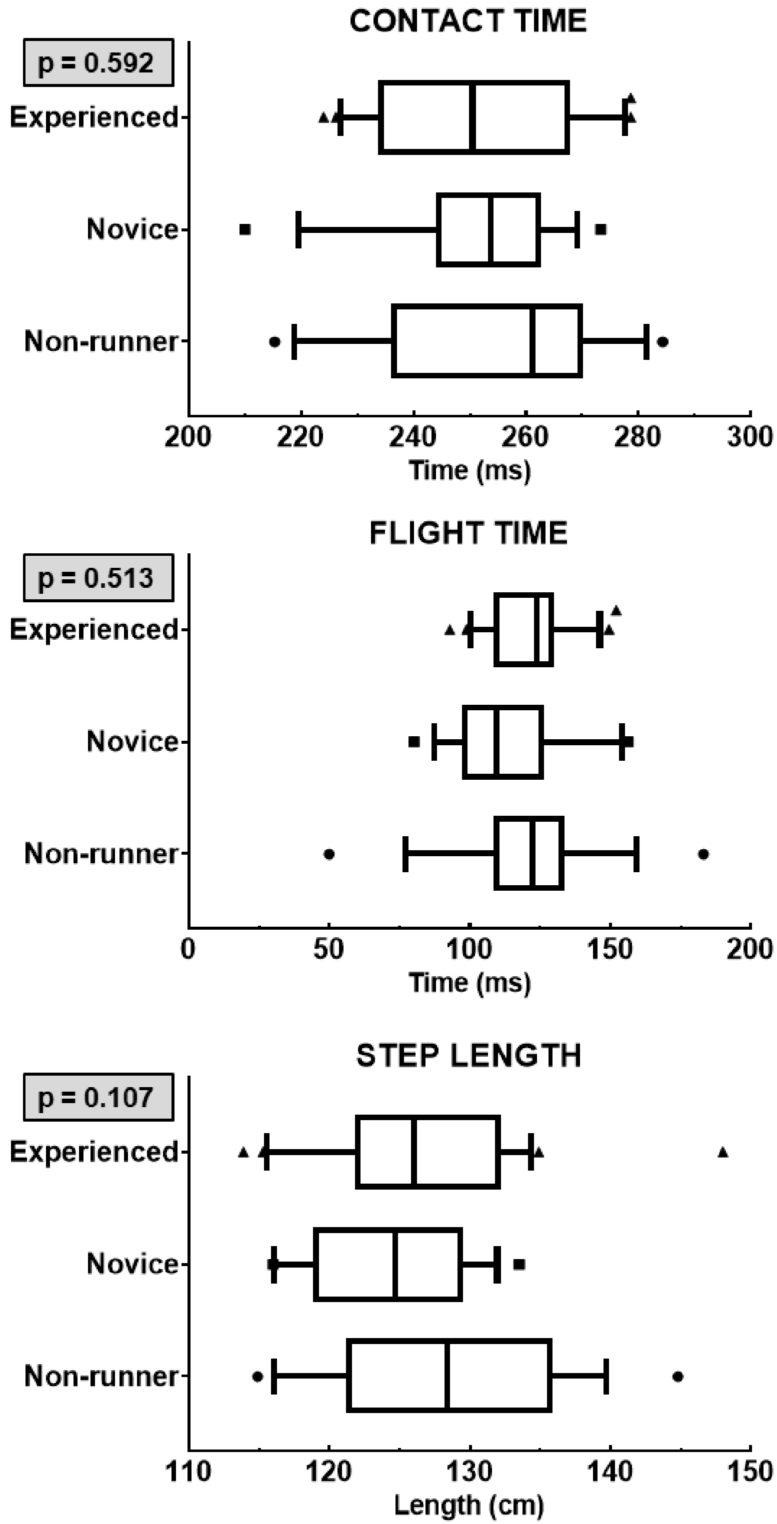

Figure 3. Contact time, flight time, and step length among the three groups. Boxplots show values for 10th, 25th, 50th, 75th and 90th percentile and symbols represent individual values outside the 10th – 90th range with triangles indicating experienced runners, squares indicating novice runners, and circles representing non-runners.

6. DISCUSSION

Some stiffness and spatiotemporal metrics have been associated with running-related injuries in individual studies (Messier et al. 2018; Malisoux et al. 2022). Examining gait characteristics that differ between less-experienced versus more-experienced runners may explain why less-experienced runners have a greater risk for injury than more-experienced runners. However, the findings from previous studies evaluating the effects of experience on running biomechanics are conflicting, which may be due to including less-experienced participants with some level of running experience and by evaluating dependent variables that focus on a single component of running mechanics (Schmitz et al. 2014; Boey et al. 2017; Gómez-Molina et al. 2017; Agresta et al. 2018, 2019; Mo et al. 2020). Examining gait kinematics and kinetics in isolation may limit information about a person's running mechanics. In this study, we addressed these limitations by comparing leg and joint stiffness and spatiotemporal parameters between a group with no running experience, a novice group, and experienced group. Contrary to our hypothesis, we found no significant differences between groups for any of the metrics evaluated.

We defined the inclusion criteria so that one group consisted of participants inexperienced to running training because, after initiating a running program, these individuals are more likely to suffer a running-related injury and abandon running (Fokkema et al. 2019). The non-significant differences found in the present study support the possibility that either individuals truly inexperienced with running may have an embedded motor program for running that is not altered with training alone or stiffness and spatiotemporal parameters specifically are not significantly modified or adapted with training. Other factors (e.g., coaching) may explain the differences between groups found in some studies. Thus, adaptations that improve running performance in the absence of coaching or other instruction may be neurophysiological and not biomechanical. Although not the aim of this study, these results suggest that other factors (such as optimized training/rest protocols and greater tolerance to repetitive loads) are likely responsible for the greater injury rate in less- experienced runners (Agresta et al. 2018).

Stiffness and spatiotemporal metrics were selected for this study as they result from complex interactions by the central nervous system and the musculoskeletal system, represents the response of the musculoskeletal system to the external environment, and encompass the kinematic, kinetic, and motor strategies for absorbing GRFs (McMahon and Cheng 1990; Farley and González 1996; Ferris and Farley 1997). Greater knee stiffness has been associated with running-related injuries (Messier et al. 2018), possibly due to increased musculoskeletal loading on bony structures (Butler et al. 2003). However, insufficient stiffness may lead to excessive joint motion and a greater risk of soft tissue injuries (Butler et al. 2003). The potential for this non-linear relationship with stiffness and injury was observed in a previous study, but only weak or non-significant associations with injury were found (Davis and Gruber 2021). In our study, we found no significant differences in leg, ankle, or knee stiffness between the groups. Therefore, our results do not support the hypothesis that there are biomechanical gait differences between differently experienced runners. Taken together with previous running experience studies (Schmitz et al. 2014; Boey et al. 2017; Agresta et al. 2018), it may be appropriate to reject a gait-related rationale for greater injury rates in less-experienced runners.

Changes in motor patterns can occur within 10-weeks of practice (Moore et al. 2012). For this reason, evaluating a group of non-runners is necessary to determine if differing gait mechanics explain different injury rates between novice and experienced runners. Our results are in-line with Schmitz et al. (2014) who included a similarly defined group of non-runners and found no significant differences in hip kinematics or vertical GRF metrics. Our non-runners group included 5/17 participants who performed activities involving intermittent, multidirectional running for 1–10 hours/week. This subgroup of non-runners were within the 5–95 percentiles of the non-runners group for each metric examined (two exceptions, where the values were with 4% of the closest participant), which further supports that their inclusion did not meaningfully influence the results.

One possible justification for the lack of significant differences between people with different levels of experience is that distinct differences in stiffness may not emerge when running in a typical environment. Instead, distinct differences in stiffness may emerge only when examining stiffness adaptations to different running conditions. For example, runners alter leg and joint stiffness when running at different speeds (Arampatzis et al. 1999; Brughelli and Cronin 2008; Kuzmeski et al. 2021), on different surfaces (Ferris et al. 1998, 1999), when barefoot (Sinclair et al. 2016), or with minimalist, traditional, or maximalist shoes (Borgia and Becker 2019; Gruber et al. 2021). Changes in stiffness under various conditions may influence both performance and injuries (Butler et al. 2003; Messier et al. 2018). The ability to adjust stiffness under various conditions may only develop with sufficient exposure to different running conditions, thus stiffness may be less adaptable in non- and less-experienced runners than more-experienced runners. Because adjustments in stiffness are required for activities other than running (e.g., jumping, landing, hopping, etc.), past experiences in other physical activities influence the ability to adjust leg and joint stiffness under different conditions.

There are several methods to calculate leg stiffness, depending on whether GRF data are available and on how many planes of movements are measured (Kuzmeski et al. 2021). Leg stiffness magnitudes vary with each method, ranging from approximately 10–40 $kN \cdot m^{-1}$ when running at a speed similar to the present study (3.5 $m \cdot s^{-1}$) (Liew et al. 2017; Kuzmeski et al. 2021). Leg stiffness may be overestimated when not accounting for all three dimensions (Liew et al. 2017). The present study adopted the multiplanar method which accounts for the motion and forces occurring in all three planes, and may better approximate the important contribution of non-sagittal aspects to injury mechanisms than other methods (Liew et al. 2017; Willwacher et al. 2022). No previous study compared leg stiffness between running experience groups, so the influence of calculation method is necessary to consider only when comparing our results to previous studies examining leg stiffness between groups or conditions other than running experience.

None of the three spatiotemporal metrics measured were significantly different between experience groups. These findings are generally in-line with existing literature comparing spatiotemporal metrics between running experience groups (Gómez-Molina et al. 2017; Agresta et al. 2018; Mo et al. 2020; Quan et al. 2021). These studies examined groups with a range of running experience, including no specific running training (undefined) to those with a minimum of five years of practice. Only one study observed differences in any spatiotemporal metric, finding longer step lengths in less-experienced runners than more-experienced runners (Gómez-Molina et al. 2017). No previous study examined spatiotemporal metrics between truly non-runners.

Our participants were assessed while performing a discontinuous running task (i.e., single trials of 18-meters) at a fixed speed (3.35 $m \cdot s^{-1}$). Given the well-documented influence of running speed on spatiotemporal parameters (Fukuchi et al. 2017; Hollis et al. 2021) and leg stiffness (Arampatzis et al. 1999; Kuzmeski et al. 2021), we chose to standardize speed to ensure all groups were performing the same mechanical task. A continuous running task (e.g., track, outdoors) may reveal group differences if stiffness varies between lab and free-living environments, but caution comparing environments is given that stiffness changes with running surface (Ferris et al. 1998, 1999).

The number of participants in each group are in-line with other studies investigating the effects of experience on biomechanics (Boey et al. 2017; Gómez-Molina et al. 2017; Maas et al. 2018). Our study was powered at .80 to identify large effect sizes, therefore, given that the effect sizes for the dependent variables were trivial to small, it is unlikely that a larger sample would present significant differences.

It is important to note that this study and all previously published studies were cross-sectional. To identify if biomechanics change as running experience develops, and its influence on injuries, a prospective study which follows participants from when they start running until they are considered experienced are necessary.

7. CONCLUSION

Leg, knee, and ankle stiffness may provide more comprehensive insights of gait patterns than isolated kinematic or kinetic measurements given that stiffness represents the complex interaction between different components of the musculoskeletal system and the environment and dictates spatiotemporal metrics, the final output of gait. Therefore, the lack of differences in leg and joint stiffnesses, contact time, flight time, and step length between non-runners, novice runners, and experienced runners supports a hypothesis that running is an inherent motor program that is not altered significantly with running exposure. Therefore, running gait may not differentially affect injury rates between experience levels.

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9. SUPPLEMENTAL MATERIAL

Supplemental material 1. Between group comparisons for the dependent variables where right and left sides were calculated separately and subsequently pooled in the main results.

Variable - Side	Experienced	Novice	Non-runner	p-value
Leg stiffness ($kN \cdot m^{-1}$) - Right	32.8 ± 4.5	32.1 ± 6.7	31.9 ± 5.0	0.834
Leg stiffness ($kN \cdot m^{-1}$) - Left	31.4 ± 5.5	31.1 ± 6.9	32.8 ± 4.5	0.555
Leg stiffness ($kN \cdot m^{-1}$) - Pooled	32.1 ± 4.9	31.6 ± 6.6	32.7 ± 6.6	0.850
Ankle stiffness (N.m/deg) - Right	9.4 ± 2.3	9.2 ± 3.0	9.8 ± 2.8	0.428
Ankle stiffness (N.m/deg) - Left	9.6 ± 2.5	9.1 ± 2.5	9.6 ± 2.7	0.388
Ankle stiffness (N.m/deg) - Pooled	9.5 ± 2.3	9.2 ± 2.7	9.7 ± 2.7	0.395
Knee stiffness (N.m/deg) - Right	7.4 ± 1.9	7.7 ± 1.7	7.8 ± 1.8	0.646
Knee stiffness (N.m/deg) - Left	6.9 ± 1.5	7.2 ± 1.5	7.5 ± 1.9	0.249
Knee stiffness (N.m/deg) - Pooled	7.2 ± 1.7	7.5 ± 1.5	7.7 ± 1.8	0.382
Contact time (ms) – Right	251.0 ± 17.8	249.6 ± 18.5	255.3 ± 20.9	0.534
Contact time (ms) - Left	251.0 ± 21.2	250.1 ± 16.8	259.9 ± 23.1	0.681
Contact time (ms) - Pooled	251.0 ± 18.7	249.8 ± 17.3	254.6 ± 21.8	0.592

STUDY 4 – RELIABILITY OF FORCE-DERIVED METRICS DURING FUNCTIONAL TASKS IN HEALTHY AND INJURED PARTICIPANTS

1. CONTEXT

During my final year in the PhD, I conducted a second visiting period at Queen Mary University of London, UK, under the supervision of Prof. Dylan Morrissey. This visit lasted from January 2023 until September 2023. The aim of this visit was to acquire knowledge and experience which is more applied to clinical populations, as the focus of my studies have been on the relationship between biomechanics and injuries. Because Dr. Morrissey is a practicing physiotherapist and has most of his research directly applied to an injured population, his laboratory was a logical choice.

During the visit, I started developing a project together with Dr. Morrissey's PhD student Merve Bodur. This multiple-year project has the aim of investigating whether there are measurements and metrics that can predict the recovery outcome of an injured patient. We developed a protocol that included measurements related to patient history, clinical evaluation, pain, patient reported outcomes through questionnaires, imaging, strength measurements, gait analysis and functional tasks. Prior to conducting the prospective study, we felt it was necessary to also conduct a feasibility and reliability study of the evaluation protocol. The data regarding the reliability of functional tasks were used for a stand-alone study that is included in the present thesis.

This study deals with aim 5 of the present thesis, focusing on the choice of metrics and on the reliability of said metrics, representing an important factor to be taken into account when using functional tasks for the evaluation of healthy and pathological people.

The reliability and feasibility project is still in progress and only the first fifteen participants were included in the following study for logistic reasons. Once a higher number of participants has been evaluated, the analysis will be updated and only then it will be submitted for publication.

2. ABSTRACT

Background: Ground reaction force-derived metrics can provide important information about movement during different functional tasks. In particular, asymmetry assessment is commonly used in clinical evaluation of patient status – being a key driver of diagnostic decisions and functional status judgements. Evaluating the reliability of these metrics, particularly in a sample containing injured participants, could allow clinicians to make informed decisions regarding patient status, and therefore diagnosis and progression. The objective of this study was to evaluate the reliability of several kinetic metrics during seven functional tasks with a progressively more complex challenge when performed by participants with and without lower limb injuries.

Methods: Fifteen participants (seven injured) performed seven functional tasks on two force plates following a progressive loading order: single-leg balance task, double-leg squat, single-leg squat, double-leg countermovement jump, single-leg countermovement jump, double-leg landing, and single-leg landing. The same tasks were collected in two different sessions with and interval between 2 and 13 days. Force-derived metrics were extracted, and asymmetry was calculated when data were available separately for each side. Reliability was assessed for all combined, injured, and non-injured participants by intraclass correlation coefficients (ICCs), standard error of measurement (SEM) and SEM relative to the mean (for the non-asymmetry metrics; %SEM). ICCs were classified as poor (<0.500), moderate (0.500-0.750), good (0.750-0.900) and excellent (>0.900).

Results: Double-leg squat and countermovement jump were the most reliable tasks, with all but one metric presenting moderate to excellent ICCs, regardless of injury status (peak landing force asymmetry ICC = 0.470). SEMs for asymmetry metrics ranged from 3.2 to

10.8%, while for the non-asymmetry metrics it ranged from 5.3 to 20.7% of the grand mean. Double-leg landing and the four single-leg tasks presented poor or at the lower end of moderate ICCs for all metrics evaluated, with exceptions depending on the group assessed. ICCs for comparisons including all participants ranged from 0.000 to 0.594 and SEMs from 2.5 to 41.4%. There was a tendency for the injured group to present less reliable values, but this was dependent on the task and metric.

Conclusions: Given that reliability of force-based metrics, particularly asymmetry, was considered poor in several cases, caution should be taken when using their results for clinical decision-making.

3. INTRODUCTION

Functional tasks are a popular tool to assess athletes and physically active individuals inside the clinic, the field and research laboratories (Nakagawa et al. 2013; Glaviano et al. 2020; Hetsroni et al. 2020; Tan et al. 2020), as it difficult to readily conduct sport-specific evaluations, despite the current investment in technology that seeks to make them available. Many different tasks and metrics have been used depending on why and who is being evaluated (Nakagawa et al. 2013; Glaviano et al. 2020; Hetsroni et al. 2020; Tan et al. 2020) as they each have different characteristics. The load the body structures need to absorb, the forces required to execute the movement and the mechanisms that are utilized (e.g., concentric, eccentric, or isometric contractions, using the stretch-shorten cycle, etc.) differ markedly between tasks (Tanikawa et al. 2013; Donohue et al. 2015). In clinical practice, a patient might only be asked to perform low-load tasks such as balance and squats at the early stages of recovery (only after being able to bear weight in some cases) and move on to more demanding tasks such jumps and landings as they progress in their recovery process.

Although some valuable information can be obtained by qualitative assessment (Padua et al. 2009; Crossley et al. 2011), having quantifiable results allows for more precise decision-making, as comparisons between limbs and over time are made easier. Currently, most of the literature focuses on kinematics (e.g., joint angles) or kinematic-dependent measurements (e.g., joint moments) to quantify how participants move during functional tasks (Glaviano et al. 2020; Tan et al. 2020). However, there is important information that can be obtained with isolated force measurements (Ueno et al. 2020; Jeon et al. 2021). The main advantages of focusing the assessment on kinetics is that force plates are typically easier to use because they require less preparation and post-processing time than traditional motion capture systems. In fact, there are systems available commercially, such as the VALD ForceDecks (VALD Performance, Brisbane, Australia) that prioritize usability and instantaneous results, despite only recording one axis.

One of the many types of metrics that can be extracted from ground reaction force measurements is limb asymmetry, which can be calculated for any metric from which there is data available for both limbs separately. Injuries may result in altered movement mechanics due to a conscious choice to protect the injured limb or an involuntary loss of strength or avoidance of pain (Dai et al. 2014; Ithurburn et al. 2015; Emamvirdi et al. 2023). In addition, asymmetry during habitual or athletic movements may disproportionally load one side, leading to accumulation of micro-traumas and possibly injuries (Helme et al. 2021). Comparing the injured to the non-injured limb may be more useful than a comparison to normative ranges during functional tasks. Therefore, asymmetry is a valuable option to assess patient status or who is at greater risk of suffering an injury, being a key driver of diagnostic decisions and functional status judgements, consequently governing which aspects the rehabilitation/prevention program should prioritize.

A patient may obtain different results when performing the same task without there being any true change in their condition, because factors such as instrumentation error, tester error and participant learning effects, fatigue or mood can all lead to changes in the measurements (Weir 2005). Therefore, to be able to correctly use the results obtained, it is necessary to identify if the change in results between assessments is due to true participant change or due to measurement error. This quantification of reliability can be conducted by evaluating a sample two or more times and using statistical tests such as intraclass correlation coefficients (ICC) and standard error of measurement (SEM) to obtain a relative and absolute index of reliability, respectively (Weir 2005; Malfait et al. 2014). Only by knowing how reliable a metric is, can a clinician make informed decisions on patient status and consequent next steps on the rehabilitation process.

There are several studies that have evaluated the reliability of different measurements and metrics during functional tasks (Meshkati et al. 2011; Alenezi et al. 2014; Malfait et al. 2014; Byrne et al. 2021), however, the study of reliability of ground reaction force-derived metrics has been limited to mainly jumping and balance tasks (Meshkati et al. 2011; Malfait et al. 2014; Heishman et al. 2020). In addition, despite the important application of these tasks to clinical populations, most studies have only evaluated young and healthy participants (Meshkati et al. 2011; Alenezi et al. 2014; Malfait et al. 2014; Byrne et al. 2021), who likely present lower variability in their movements and may present far different ranges in the metrics evaluated (Baida et al. 2018). The movement strategies of injured participants can be affected by the amount of pain or stiffness they are experiencing on the day, which can vary between days depending on many factors and consequently be less reliable. Therefore, knowing the measurement error will also allow for the identification of the true participant variability. With this study we aimed to verify if force-derived measurements can be reliably used for patient and healthy populations. To achieve this, the objective this study was to evaluate the reliability of several kinetic metrics during seven functional tasks with a progressive load when performed by participants with and without lower limb injuries.

4. METHODS

Participants

Recruitment for this study was conducted through communication with the university students and staff, in person community outreach and word of mouth. Participants were included if they were aged 18 years or older, had no systemic inflammatory disease affecting joints and muscle and were able to independently bestow consent. For the injured subgroup, participants needed to show a hip, knee or ankle injury diagnosed by a relevant clinician, while for the non-injured subgroup, participants had no injury for the past six months. The study was approved by the university's ethics committee (QMREC2018/48/111) and before taking part, all participants gave independent and informed consent.

Experimental overview

Prior to the start of the study, we conducted two Public and Patient Involvement events (one with four clinicians and another with four patients with lower-limb injuries) in order to listen to their views, and adapt as needed, on a larger project where the reliability and feasibility of a battery of tests (history, patient reported outcomes, imaging, pain, clinical examination, strength, gait assessment and functional tasks) was investigated. As this manuscript focuses on the functional tasks, the results from the remaining domains will be reported elsewhere. To assess functional task reliability, participants visited the laboratory in two different days with 2 to 13 days in between and the reliability of the results obtained in each day was compared (Figure 1).

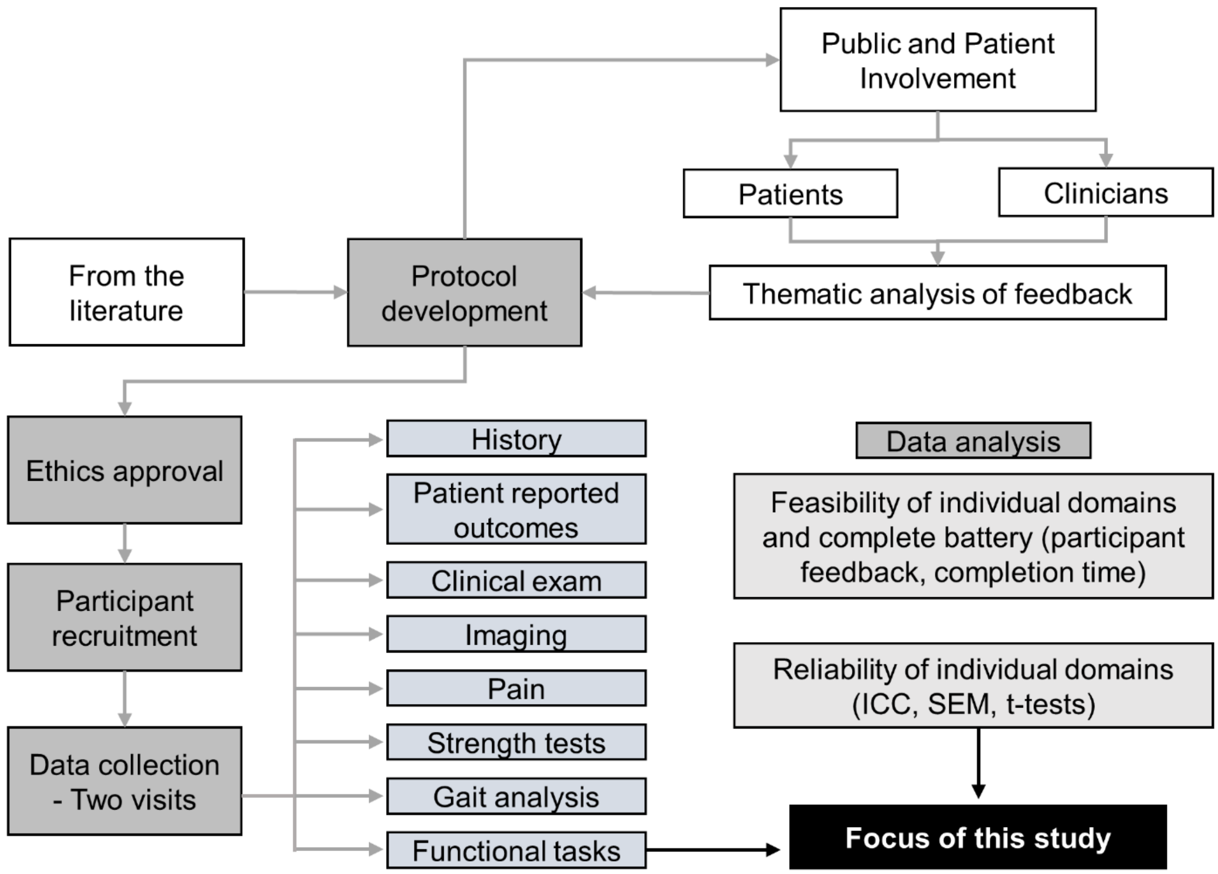


Figure 1. Flowchart detailing the main project conducted and the focus of the present study.

Functional tasks

Seven functional tasks were performed in a graded loading order (Figure 1). Participants started with the single-leg balance task ($SL_{BALANCE}$), followed by the double-leg squat (DL_{SQUAT}), single-leg squat (SL_{SQUAT}), double-leg countermovement jump (DL_{CMJ}), single-leg countermovement jump (SL_{CMJ}), double-leg landing (DL_{LAND}) and single-leg landing (SL_{LAND}). All tasks were executed while participants kept the hands on their hips and three trials were completed (six total trials for single-leg tasks). For all single-leg tasks, participants were asked to keep the contralateral knee flexed at 90° and the contralateral thigh perpendicular to the ground. For the $SL_{BALANCE}$, participants were instructed to balance on one foot for 20 seconds, staying as still as possible. During DL_{SQUAT} and SL_{SQUAT}, participants were instructed to squat into a comfortable position at a rhythm of one second down and one second up. For DL_{CMJ} and SL_{CMJ}, participants were instructed to jump as high as possible and land roughly on the same spot. For DL_{LAND} and SL_{LAND}, participants jumped from a box 30 cm high and positioned at a distance equivalent to their lower limb length and were instructed to maintain their balance for three seconds following foot contact. Although participants were asked to perform all tasks, a few were unable or did not feel comfortable with some tasks and were therefore excluded from the analysis of that task. Similarly, a few participants were only able to complete one or two trials for some tasks, which were included in the analysis.

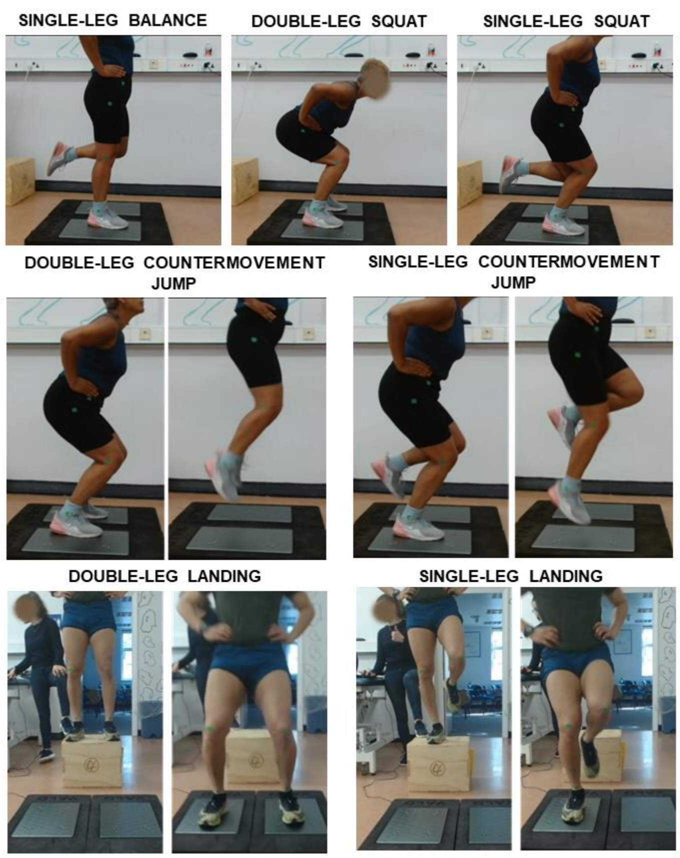


Figure 2. Tasks performed by the participants.

Instrumentation

The VALD ForceDecks dual-plate systems (FDlite, VALD Performance, Brisbane, Australia) was used in this study. It consists of two connected single-axis force plates with a sampling frequency of 1000 Hz. Based on the force measurements, the system can detect each trial and immediately calculate a range of metrics for each functional task.

Data analysis

<u>Validation of the software</u>

The automatic event detection for the system in all tasks was checked using the raw force plate recording extracted to a text file, the individual trial graphs produced by the ForceDecks software and video recording of the participants performing the tasks. Using a plot produced with the raw recordings, one analyst tagged the moment (in milliseconds) in which each repetition started and finished. When the events were not easily detected by viewing the raw recordings, the analyst used the video recording to determine each event. Once all moments were defined, they were compared to the events automatically determined by the software. If the manual and automatic events matched, the automatic detection was considered correct. This was done for a sample of five participants and all trials were judged to be correct.

In addition, the metrics automatically calculated by the software for the double-leg countermovement jump were also validated. Using the raw data for each participant, a custom-built MatLab script, based on the script published by Harry (2021), was used to re-calculate the same metrics extracted from the VALD ForceDecks software.

For each task two to seven metrics were selected to represent different aspects of the movement (Table 1). The calculated values for each metric were extracted for all valid trials of both sides. Because the system detects the values for the left and right trials, the data were manually rearranged to represent the injured and non-injured sides (for injured participants) and non-dominant and dominant (for non-injured participants). The dominant side was determined by having the participants kick a ball three times (van Melick et al. 2017) and was merged with the non-injured side from the injured participants as it can be considered the "good" leg.

For the metrics that were not separated by side (e.g., jump height for the DL_{CMJ}), only one value was obtained for each trial. For the metrics that were extracted for each side, the asymmetry was calculated with Equations 1-2, depending on the participant injury status. With these equations, negative values indicate that the non-injured/dominant side had a higher value than the injured/non-dominant side and positive values indicate the opposite.

$$\frac{(Injured\ limb - Non\ injured\ limb) * 100}{Non\ injured\ limb}$$

Equation 1. Asymmetry calculation for injured participants

$$\frac{(Non\ dominant\ limb - Dominant\ limb) * 100}{Dominant\ limb}$$

Equation 2. Asymmetry calculation for non-injured participants

Table 1. Metrics selected for the evaluation of all tasks. Unit of measurement is % unless otherwise specified.

SINGLE-LEG BALANCE
Asymmetry of center of pressure range in the Anterior-Posterior direction
Asymmetry of center of pressure range in the Medio-lateral direction
Asymmetry of the mean velocity of the center of pressure
Asymmetry of the total excursion of the center of pressure
DOUBLE-LEG SQUAT
Concentric peak power (W)
Maximum negative displacement – Squat depth (cm)
Asymmetry of eccentric peak force
Asymmetry of concentric peak force
SINGLE-LEG SQUAT
Asymmetry of concentric mean power
Asymmetry of eccentric mean power
Asymmetry of concentric peak power
Asymmetry of maximum negative displacement – Squat depth
DOUBLE-LEG COUNTERMOVEMENT JUMP
Jump Height – Flight time (cm)
Countermovement depth (cm)
Peak landing power (W)
Asymmetry of eccentric mean force
Asymmetry of concentric mean force
Asymmetry of peak landing force
SINGLE-LEG COUNTERMOVEMENT JUMP
Asymmetry of jump height – Flight time
Asymmetry of countermovement depth
Asymmetry of concentric mean power normalized by bodyweight
Asymmetry of concentric mean force
Asymmetry of peak landing force
Asymmetry of peak landing power
DOUBLE-LEG LANDING
Time to stabilization (s) – From the moment of foot contact until stabilization
Asymmetry of peak drop landing force
SINGLE-LEG LANDING
Asymmetry of time to stabilization – From the moment of foot contact until stabilization
Asymmetry of peak drop landing force

Statistical analysis

For all tasks and all metrics, statistical analysis was performed for all participants as well as for injured and non-injured group separately. A paired-samples t-test was used to identify eventual systematic differences between the results obtained in day 1 and in day 2. Relative reliability was calculated using a two-way random, absolute agreement, single measurements intraclass correlation coefficient ($ICC_{2,1}$) and its 95% confidence interval (Weir 2005). ICCs lower than 0.5 were considered poor, between 0.5-0.75 moderate, between 0.75-0.90 good and above 0.90 excellent (Koo and Li 2016). It is important to note that because of the limited sample, particularly in the stratified calculations, ICC calculation were invalid due to low between subject variance and were reported as 0. The SEM was estimated to evaluate the absolute index of reliability by using the square root of the mean square error term from an ANOVA and are reported in the same measurement unit as the metric (Weir 2005). To facilitate interpretation, for the non-asymmetry variables, the SEM was reported both as the absolute value and as the percentage of the pooled mean from day 1 and day 2. All analysis were conducted using Jamovi (version 2.3.28, Jamovi, Sydney, Australia).

5. RESULTS

Fifteen participants took part in the study, with seven presenting lower limb injuries. Validation of the VALD ForceDecks calculated metrics showed excellent ICCs between the software and the custom-made script for all but one metric, which showed good ICCs (ICC = 0.891; Supplementary Material 1). Figures 3-9 show the results of the paired sample t-tests and the reliability analysis.

During $SL_{BALANCE}$, centre of pressure ranges in the anteroposterior direction was significantly different between days when considering pooled participants ($p = 0.024$) and in the mediolateral direction for the pooled ($p = 0.006$) and non-injured groups ($p = 0.003$). $SL_{BALANCE}$ showed poor ICC values for all metrics when all participants were pooled and when evaluating injured participants alone. The non-injured participants presented moderate to good ICCs for all metrics but the mediolateral centre of pressure range asymmetry. SEM ranged from 12 to 35% and tended to be lower for the non-injured group (Figure 3).

During the DL_{SQUAT}, no metric presented significant differences between days for all groups ($p > 0.220$). ICC values for the pooled and non-injured participants were good to excellent, while for the injured group they were moderate to good. For the asymmetry metrics, the SEM ranged from 2 to 5% and was lower for the non-injured group. For the other metrics, SEM ranged from 10 to 20% of the pooled mean values and was substantially lower for the non-injured group only for concentric peak power (Figure 4).

Table 2. Individual participant characteristics and tasks that were not completed.

Number	Age (years)	Sex	BMI (kg/m^2)	Physical activity (days/week)	Injury	Tasks not completed
			Non-injured			
01	23	M	25.1	1	None	SL_{LAND}
02	28	M	26.8	6	None	
03	22	M	25.7	5	None	
04	22	F	22.4	3	None	
05	30	F	20.2	2	None	
06	21	M	20.1	3	None	
07	69	F	33.8	0	None	$SL_{BALANCE}$ DL_{LAND} SL_{LAND}
08	68	F	25.8	7	None	$SL_{BALANCE}$ DL_{LAND} SL_{LAND}
Mean (SD)	37.9 (17.9)	4M, 4F	26.1 (5.6)	4.0 (1.8)		
			Injured			
09	25	M	14.5	4	Accessory navicular syndrome	$SL_{BALANCE}$ SL_{SQUAT} SL_{CMJ} SL_{LAND}
10	27	F	26.0	5	Anterior Talo-Fibular Ligament	
11	28	M	29.1	1	Lateral Meniscus	
12	60	F	26.4	4	Plantar fasciitis	SL_{LAND}
13	28	F	36.7	4	Plantar fasciitis	SL_{LAND}
14	60	F	32.8	3	Achilles tendinitis	$SL_{BALANCE}$ DL_{LAND} SL_{LAND}
15	37	F	27.0	7	Lateral ankle sprain	
Mean (SD)	35.4 (20.7)	2M, 5F	25.0 (4.4)	3.4 (2.4)		
			All participants			
Mean (SD)	36.5 (17.9)	6M, 9F	5.6	3.7 (2.1)		

For SL_{SQUAT}, no metric presented significant differences between days for all groups ($p > 0.050$). ICC values were poor to moderate for all metrics when considering all participants and just the non-injured ones. With the exception of eccentric mean power, which showed poor ICCs for all groups, values for the injured group were moderate to good (>0.70 for all). SEM followed the same trend, ranging from 5 to 19%, with the non-injured group presenting higher values for all metrics but eccentric mean power (Figure 5).

During DL_{CMJ}, no metric presented significant differences between days in all groups ($p > 0.060$). DL_{CMJ} showed moderate to good ICC values for all metrics but peak landing force and eccentric mean force asymmetry, which presented poor or good values, depending on the group. Jump height and countermovement depth consistently presented good to excellent ICCs (>0.88). For asymmetry metrics, SEM ranged from 3 to 12% and showed no clear difference between injured and non-injured participants. For the other metrics, SEM ranged from 3 to 21% of the grand mean and there was also no detectable influence of injury status (Figure 6).

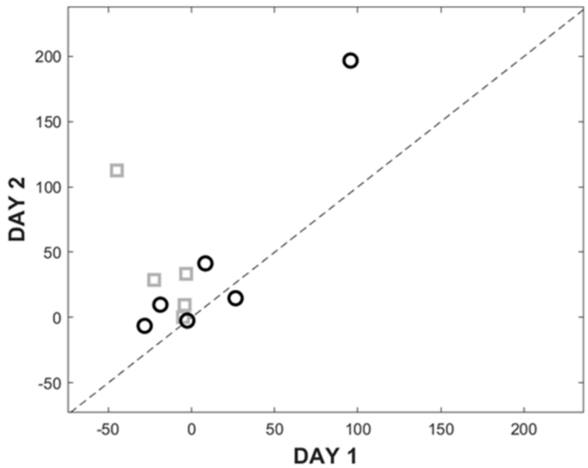
During SL_{CMJ}, only concentric mean power asymmetry presented significantly differences between days when considering pooled participants ($p = 0.022$), with no other differences found for all metrics and groups ($p > 0.061$). All asymmetry metrics presented poor ICC values. SEM values were lower than 10% for asymmetry of concentric mean power and force, as well and peak landing force asymmetry and were as high as 25% for the remaining metrics. There were no clear trends regarding differences in SEM between the groups (Figure 7).

For DL_{LAND}, no metric presented significant differences between days for all groups ($p > 0.107$). ICC values for all participants were poor for both metrics. When looking at the isolated participants, the injured group presented moderated ICC for time to stabilization and poor ICC for peak landing force asymmetry. The opposite was found for the non-injured group. SEMs followed the same pattern, ranging from 10 to 24% and being higher for the injured group in peak landing force asymmetry and for the non-injured group in time to stabilization (Figure 8).

Finally, during the SL_{LAND} both metrics did not present significant differences between days for all groups ($p > 0.90$). ICC values were poor for time to stabilization asymmetry, independent of group. Peak landing force asymmetry presented moderate ICCs for all participants or just injured and poor (ICC = 0.475) for the non-injured group. SEM were lower for the injured group, ranging from approximately 3 to 9% for peak landing force and from 15 to 41% for time to stabilization. It is important to note that only two injured participants were included in the analysis of this test (Figure 9).

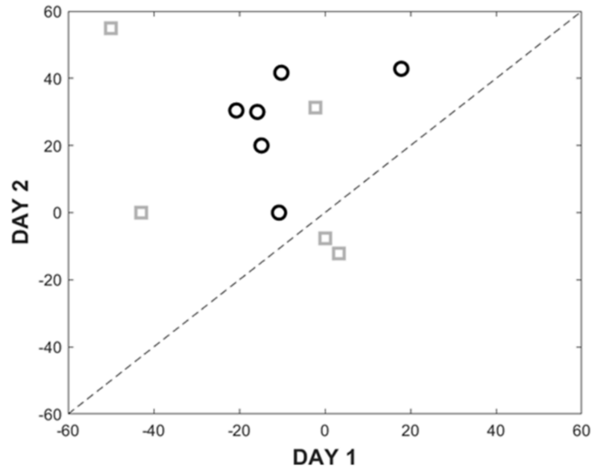
SINGLE-LEG BALANCE

COP RANGE – ANTEROPOSTERIOR (%ASY)



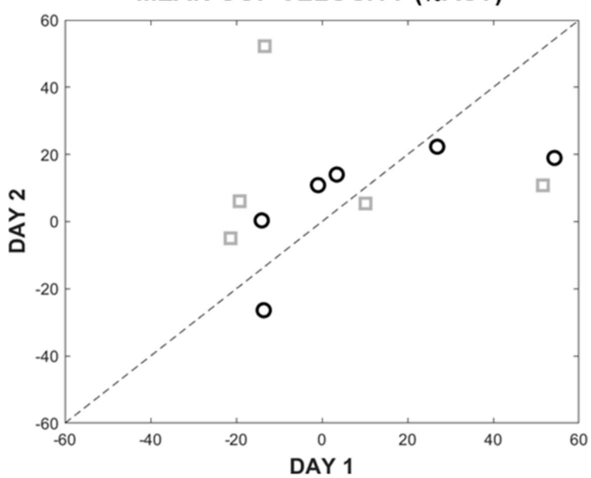
	Day1	Day2	p	ICC	95% CI	SEM
All	0.1	39.9	.024	.418	-.023-.744	34.9
Injured	-16.0	36.9	.126	0	-.265-.573	34.0
Non-injured	13.5	42.3	.133	.752	.225-.944	27.9

COP RANGE – MEDIOLATERAL (%ASY)



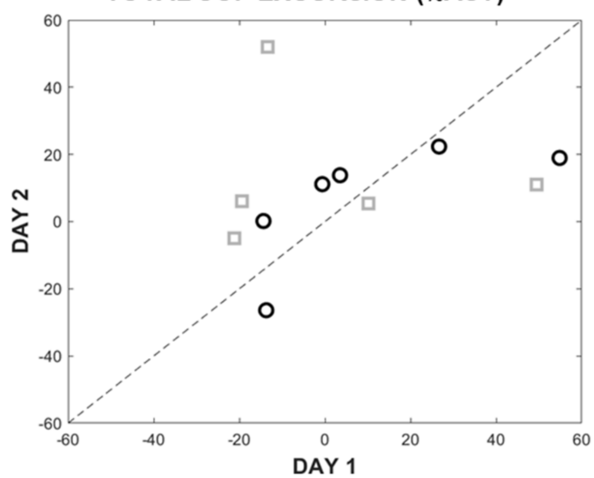
	Day1	Day2	p	ICC	95% CI	SEM
All	-13.3	21	.006	0	-.176-.308	21.2
Injured	-18.4	13.3	.215	0	-.403-.647	27.3
Non-injured	-9.1	27.5	.003	.101	-.063-.490	11.6

MEAN COP VELOCITY (%ASY)



	Day1	Day2	p	ICC	95% CI	SEM
All	5.8	10.0	.649	.265	-.254-.671	20.0
Injured	1.6	13.9	.520	0	-.729-.729	26.1
Non-injured	9.4	6.7	.748	.685	.096-.927	12.5

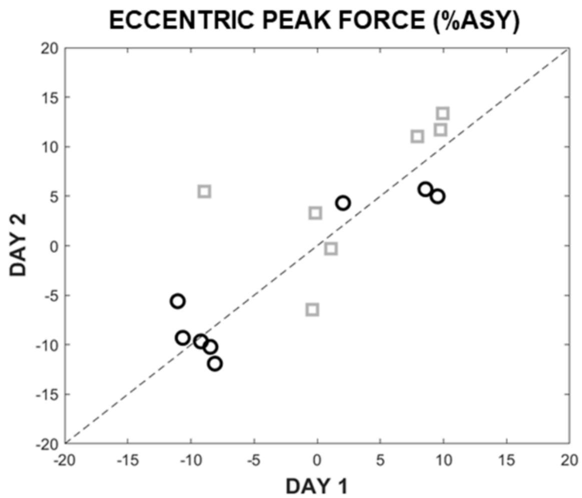
TOTAL COP EXCURSION (%ASY)



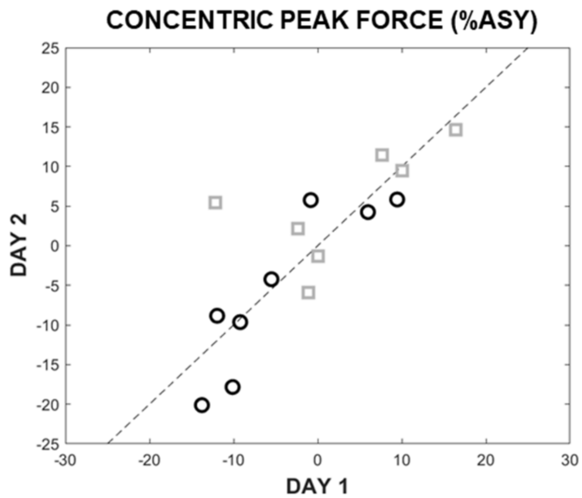
	Day1	Day2	p	ICC	95% CI	SEM
All	5.6	10.0	.631	.273	-.246-.676	19.7
Injured	1.1	13.9	.496	.000	-.729-.729	25.6
Non-injured	9.4	6.7	.745	.684	.093-.927	12.7

Figure 3. Reliability results for the single-leg balance task. Circles represent non-injured (N = 6) and squares represent injured (N = 5) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

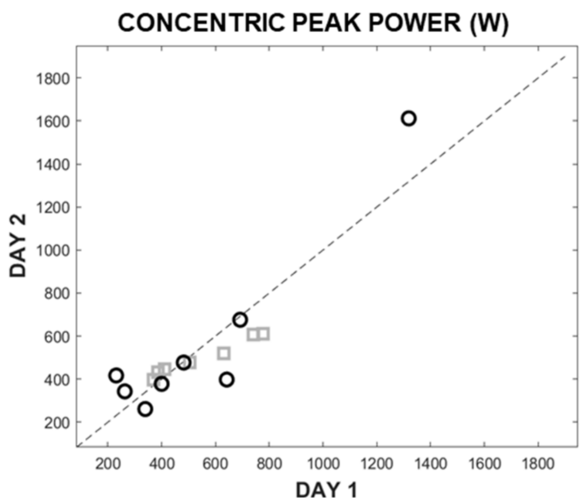
DOUBLE-LEG SQUAT



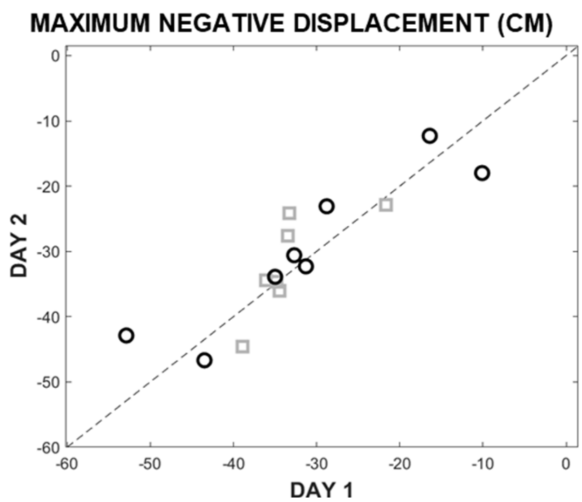
	Day1	Day2	p	ICC	95% CI	SEM
All	-0.5	0.4	.731	.829	.630-.927	3.5
Injured	2.8	5.4	.298	.599	.030-.887	4.4
Non-injured	-3.4	-4.0	.664	.922	.752-.979	2.3



	Day1	Day2	p	ICC	95% CI	SEM
All	-1.2	-0.6	.731	.814	.601-.920	4.3
Injured	2.6	5.1	.412	.626	.055-.896	5.2
Non-injured	-4.5	-5.6	.535	.881	.637-.967	3.3



	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	545	537	.810	.902	.778-.959	92	17.1
Injured	545	499	.220	.758	.316-.936	63	19.7
Non-injured	546	570	.689	.923	.755-.979	110	12.2

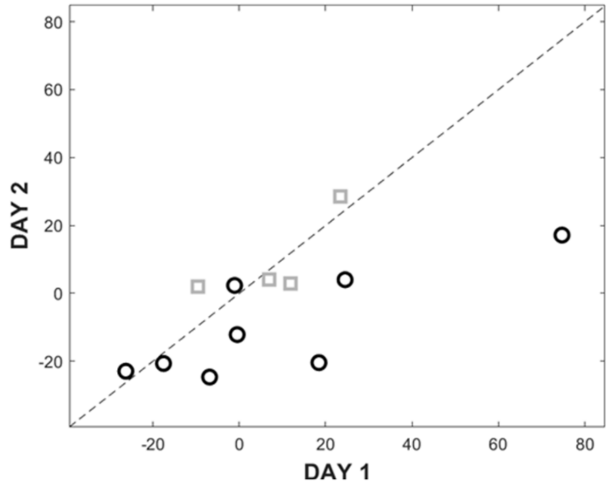


	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	-32.2	-31.0	.346	.873	.718-.947	3.6	11.3
Injured	-33.3	-32.1	.550	.743	.272-.932	3.4	10.3
Non-injured	-31.3	-30.0	.508	.912	.722-.976	3.8	12.3

Figure 4.Reliability results for the double-leg squat task. Circles represent non-injured (N = 8) and squares represent injured (N = 7) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

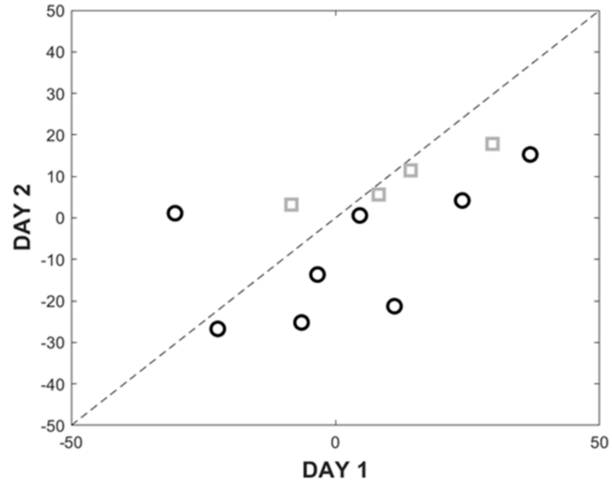
SINGLE-LEG SQUAT

CONCENTRIC MEAN POWER (%ASY)



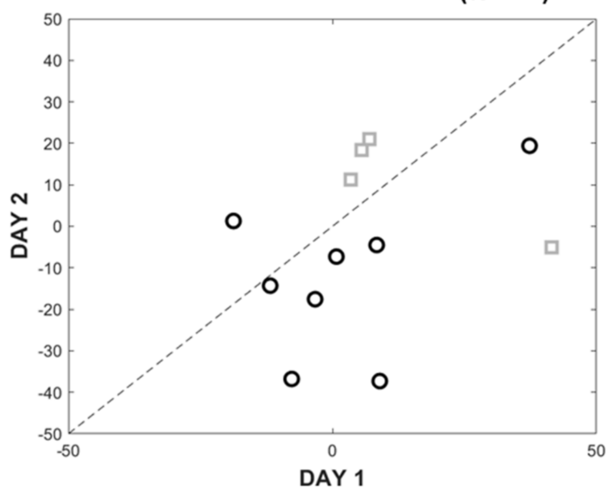
	Day1	Day2	p	ICC	95% CI	SEM
All	8.4	-3.1	.072	.503	.111-.767	13.9
Injured	8.6	7.5	.803	.759	.248-.946	5.5
Non-injured	8.2	-9.7	.050	.525	.001-.840	15.1

CONCENTRIC PEAK POWER (%ASY)



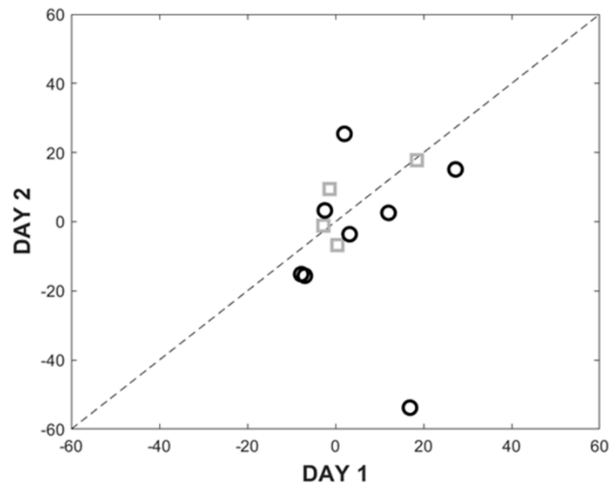
	Day1	Day2	p	ICC	95% CI	SEM
All	4.8	-2.4	.167	.514	.127-.773	11.6
Injured	9.8	6.8	.781	.710	.145-.934	5.9
Non-injured	1.8	-8.2	.186	.469	-.066-.818	13.6

ECCENTRIC MEAN POWER (%ASY)



	Day1	Day2	p	ICC	95% CI	SEM
All	5.4	-3.7	.141	.260	-.152-.615	15.5
Injured	11.2	9.8	.855	0	-.669-.669	13.6
Non-injured	1.7	-12.1	.082	.355	-.126-.754	13.7

MAXIMUM NEGATIVE DISPLACEMENT (%ASY)

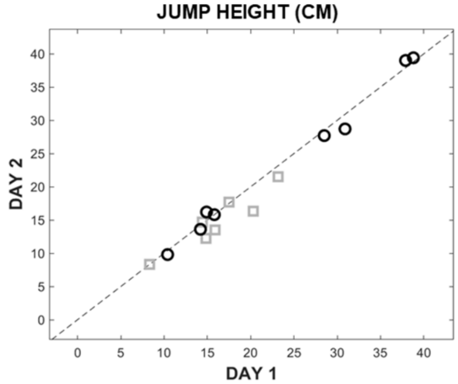


	Day1	Day2	p	ICC	95% CI	SEM
All	4.9	-1.4	.319	.064	-.401-.506	15.7
Injured	3.7	4.6	.780	.747	-.140-.955	4.6
Non-injured	5.5	-5.2	.294	.009	-.544-.580	19.0

Figure 5. Reliability results for the single-leg squat task. Circles represent non-injured (N = 8) and squares represent injured (N = 5) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

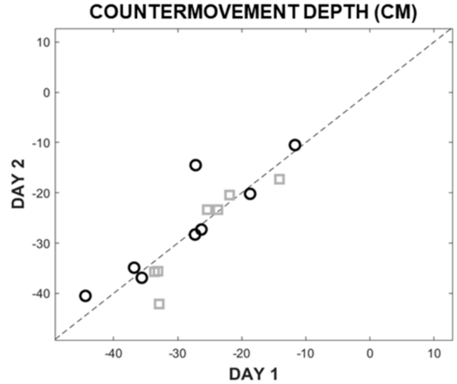
DOUBLE-LEG COUNTERMOVEMENT JUMP

JUMP HEIGHT (CM)



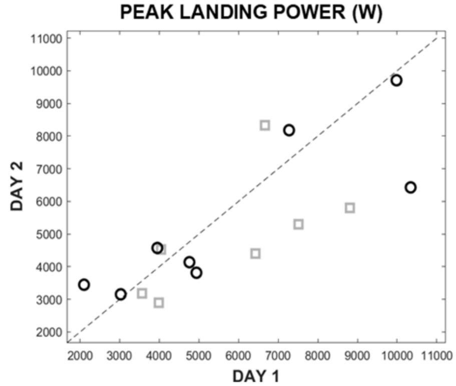
	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	20.4	19.6	.073	.985	.961-.994	1.1	5.3
Injured	16.3	14.9	.060	.893	.513-.975	1.2	7.5
Non-injured	23.9	23.8	.723	.996	.984-.999	0.8	3.2

COUNTERMOVEMENT DEPTH (CM)



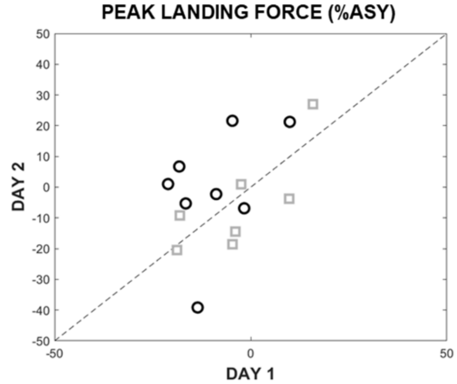
	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	-27.5	-27.4	.926	.886	.743-.952	3.2	11.5
Injured	-26.4	-28.3	.238	.888	.625-.972	2.7	9.8
Non-injured	-28.5	-26.6	.307	.896	.679-.971	3.4	12.2

PEAK LANDING POWER (W)



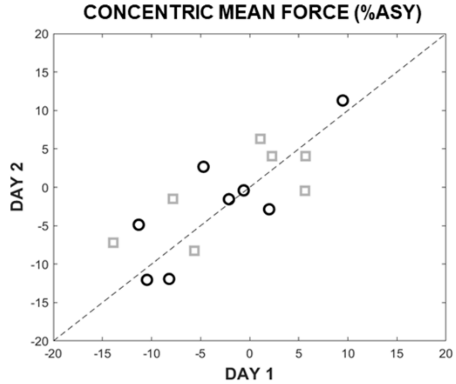
	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	5825	5191	.150	.747	.480-.889	1141	20.7
Injured	5857	4918	.182	.596	.053-.884	1165	21.6
Non-injured	5797	5430	.548	.836	.524-.953	1120	20.0

PEAK LANDING FORCE (%ASY)



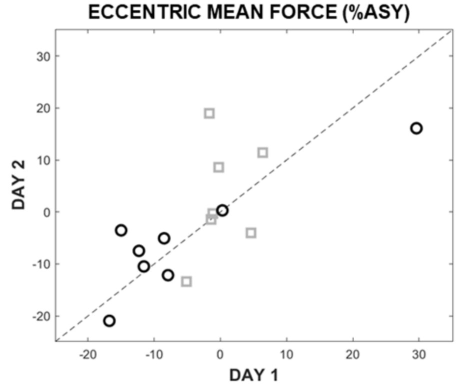
	Day1	Day2	p	ICC	95% CI	SEM
All	-6.5	-2.8	.365	.470	.065-.745	10.8
Injured	-3.2	-5.5	.583	.765	.319-.939	7.1
Non-injured	-9.3	-0.4	.191	.326	-.214-.750	12.4

CONCENTRIC MEAN FORCE (%ASY)



	Day1	Day2	p	ICC	95% CI	SEM
All	-2.6	-1.5	.376	.783	.544-.906	3.2
Injured	-1.8	-0.4	.489	.734	.253-.930	3.4
Non-injured	-3.2	-2.5	.629	.836	.523-.953	3.0

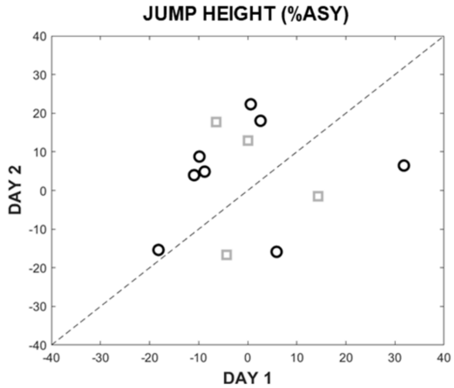
ECCENTRIC MEAN FORCE (%ASY)



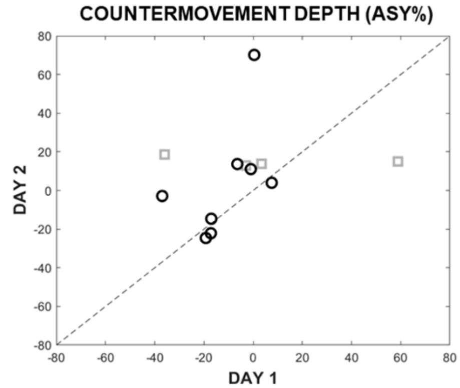
	Day1	Day2	p	ICC	95% CI	SEM
All	-2.7	-1.6	.611	.721	.435-.876	5.9
Injured	0.2	2.8	.518	.255	-.408-.754	6.9
Non-injured	-5.3	-5.4	.959	.858	.577-.960	4.9

Figure 6. Reliability results for the double-leg countermovement task. Circles represent non-injured (N = 8) and squares represent injured (N = 7) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

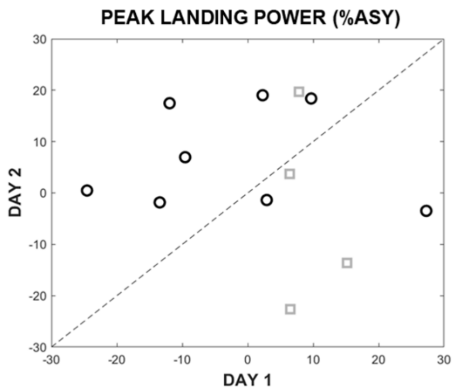
SINGLE-LEG COUNTERMOVEMENT JUMP



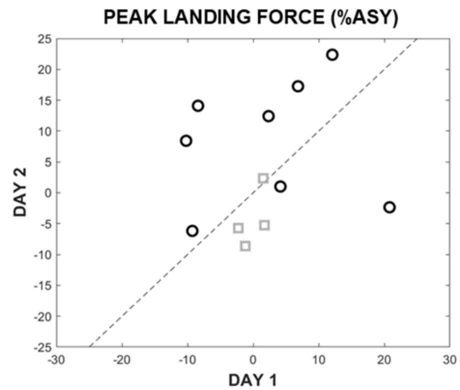
	Day1	Day2	p	ICC	95% CI	SEM
All	-0.3	3.8	.449	.131	-.362-.570	12.5
Injured	0.9	3.1	.835	0	-.805-.805	11.9
Non-injured	-0.8	4.1	.470	.231	-.386-.714	12.7



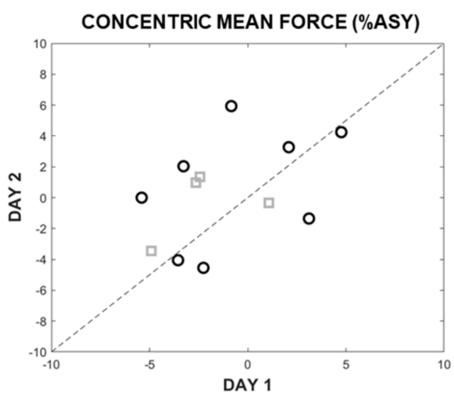
	Day1	Day2	p	ICC	95% CI	SEM
All	-5.5	7.9	.146	.261	-.179-.641	21.0
Injured	5.9	15.1	.680	.059	-.769-.824	25.6
Non-injured	-11.2	4.3	.134	.349	-.160-.757	18.3



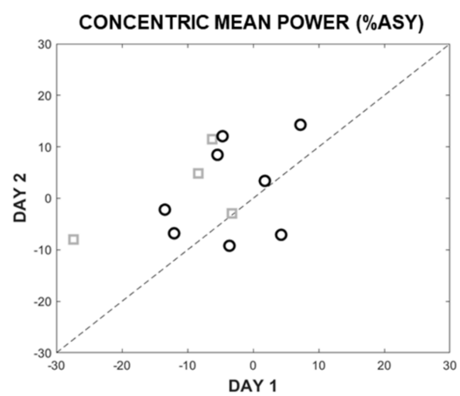
	Day1	Day2	p	ICC	95% CI	SEM
All	1.5	3.6	.749	0	-.476-.476	13.6
Injured	9.0	-3.2	.316	0	-.633-.782	13.6
Non-injured	-2.2	7.0	.219	0	-.498-.557	13.4



	Day1	Day2	p	ICC	95% CI	SEM
All	1.5	4.2	.485	.179	-.316-.602	8.8
Injured	0.0	-4.3	.133	.275	-.196-.847	2.7
Non-injured	2.3	8.4	.267	.087	-.467-.625	10.1



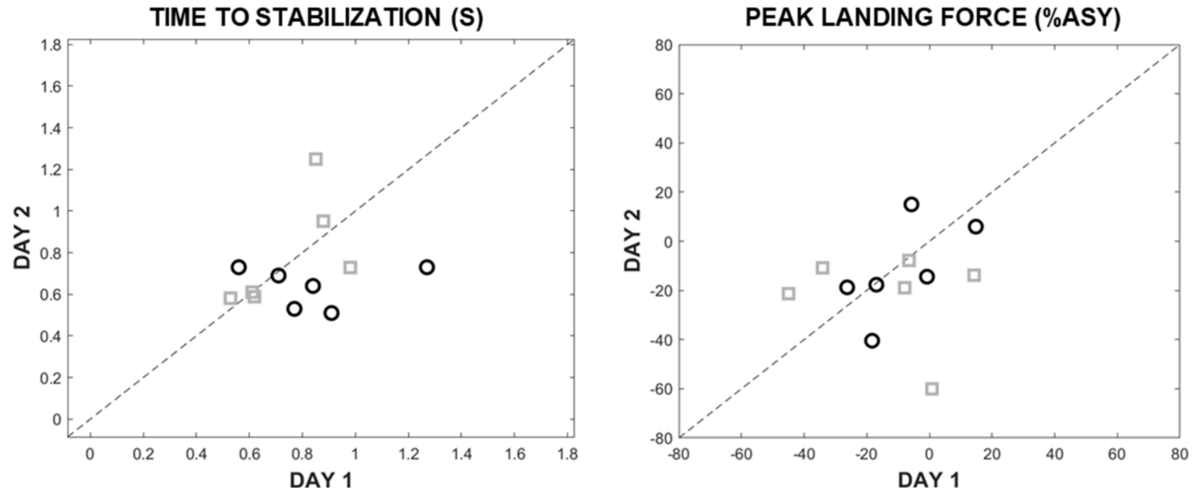
	Day1	Day2	p	ICC	95% CI	SEM
All	-1.2	0.3	.158	.410	-.032-.733	2.5
Injured	-2.2	-0.4	.224	.389	-.285-.898	1.7
Non-injured	-0.7	0.7	.375	.412	-.193-.800	2.9



	Day1	Day2	p	ICC	95% CI	SEM
All	-5.9	1.5	.022	.307	-.081-.655	6.9
Injured	-11.3	1.4	.061	.354	-.122-.868	6.1
Non-injured	-3.2	1.6	.197	.308	-.234-.741	6.8

Figure 7. Reliability results for the single-leg countermovement jump task. Circles represent non-injured (N = 8) and squares represent injured (N = 8) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

DOUBLE-LEG LANDING

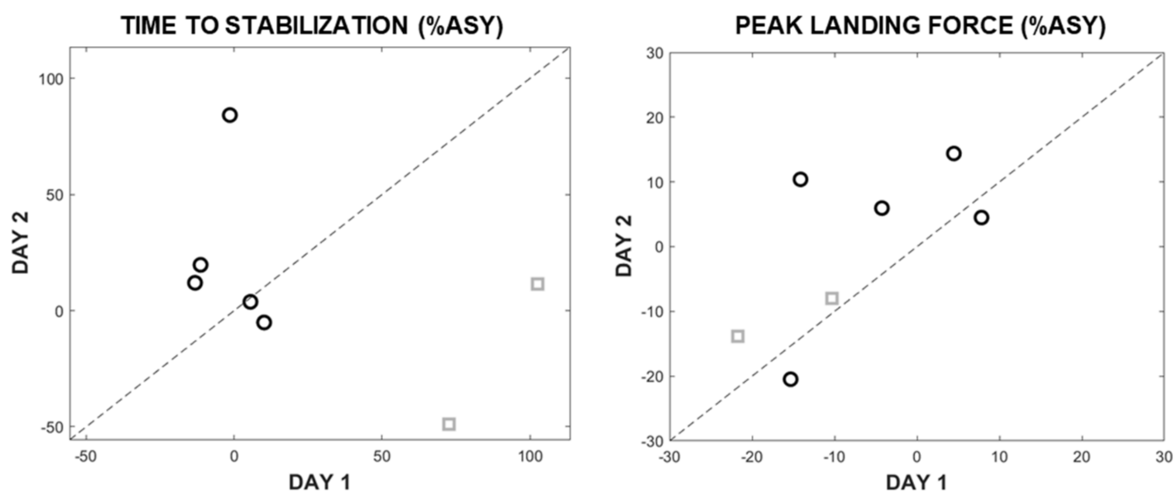


	Day1	Day2	p	ICC	95% CI	SEM	SEM%
All	0.79	0.71	.290	.233	-.252-.634	0.18	24.2
Injured	0.75	0.79	.661	.621	-.017-.910	0.13	17.0
Non-injured	0.84	0.64	.107	.023	-.358-.596	0.11	24.3

	Day1	Day2	p	ICC	95% CI	SEM
All	-11.0	-16.9	.416	.170	-.326-.596	17.0
Injured	-13.1	-22.1	.524	0	-.669-.669	20.3
Non-injured	-8.9	-11.7	.669	.650	.033-.918	10.2

Figure 8. Reliability results for the double-leg landing task. Circles represent non-injured (N = 6) and squares represent injured (N = 6) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

SINGLE-LEG LANDING



	Day1	Day2	p	ICC	95% CI	SEM
All	23.5	11.0	.662	0	-.621-.621	41.4
Injured	87.6	-18.7	.090	.134	-.017-.969	15.2
Non-injured	-2.1	22.9	.224	0	-.505-.682	26.1

	Day1	Day2	p	ICC	95% CI	SEM
All	-7.6	-1.0	.132	.594	.005-.883	7.1
Injured	-16.0	-11.0	.316	.669	-.159-.998	2.7
Non-injured	-4.3	3.0	.251	.475	-.211-.888	8.5

Figure 9. Reliability results for the single-leg landing task. Circles represent non-injured (N = 5) and squares represent injured (N = 2) participants. Grey dashed line represents perfect between day agreement. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

6. DISCUSSION

Understanding the reliability of the metrics is fundamental to be able to correctly interpret the data and, therefore, make informed decisions about rehabilitation and injury prevention protocols. In this study, we found that reliability depended on the task, injury status and type of metric (calculated asymmetry or not). Double-leg squats and countermovement jump seemed to provide the highest reliability statistics, while all single-leg tests, as well as double-leg landing, had few metrics that did not present poor ICCs. Although there was a tendency for the injured group to present lower reliability scores, this was task and metric-dependent, with some metrics showing the opposite results.

One important limitation of this study must be discussed prior to the interpretation of the findings. Although similar to other studies (Alenezi et al. 2014; Malfait et al. 2014; Byrne et al. 2021; Miner et al. 2022), our sample was likely not numerous enough to adequately power the comparisons, particularly when stratifying by injury status and during tasks where some participants did not feel comfortable performing. This resulted in the ICCs' confidence interval possibly spanning the four interpretation categories (poor, moderate, good and excellent) for some metrics. In addition, ICC calculations are conducted using the difference between between-subjects and within-subjects variability, so a reduced sample may result in larger within- than between-subjects values, resulting in negative ICC values, which have no real meaning and are therefore considered 0 in some statistical packages (Weir 2005). Although rarely found in reliability studies, Borg et al., (2022) provided a table for researchers to determine the adequate sample size based on the estimated true ICC, the number of repeated evaluations and the probability of obtaining a certain precision (e.g., CI spanning ± 0.1, 0.2, etc.). With a 50% probability of obtained CI spanning 0.2, two evaluations and an estimated true ICC of 0.90, the required sample size would be 20 participants and these values increase as the probability increases and the precision and true ICC decrease. However, our current sample of 15 participants when considering all participants and the injured and non-injured subgroups are already useful to identify trends in the reliability of force-derived metrics.

Although it is possible to obtain individual values for each side in most metrics, it may be difficult to make clinical decisions based on those results, as there is no guide to identify if a patient's injured limb is performing as well as it should be (or how limited it is) due to the high variability between people for most metrics. For this reason, the performance of the non-injured limb arises as a preferred comparison for the injured one, as it is expected there not to be important differences between them in healthy individuals. While in single-leg tasks this comparison is made between trials executed with each limb, in double-leg tasks it is also possible to assess compensations within a single movement. However, asymmetry metrics seem to be more sensitive to statistical reliability tests, as changes in either side can have an important effect on the calculated asymmetry and result in worse ICCs and SEMs, whereas a change on a specific side would be required to reduce reliability scores when considering each side individually. In our study, all the metrics that were not evaluated as asymmetry consistently presented moderate to excellent reliability, with the exception of time to stabilization during the DL_{LAND}. Although reliability of asymmetry metrics may not necessarily represent errors from the instrument or issues with the task performance (which are common factors influencing reliability), the fact remains that these are important metrics in clinical practice and the possible reduction in reliability due to the asymmetry calculation is just another factor that needs to be considered. Therefore, it is advisable to also look at the individual side results to make clinical decisions, instead of only considering asymmetry. For example, if a clinician finds that the injured limb is performing 10% worse than the non-injured after a rehabilitation protocol (from a 20% asymmetry prior), it can be that the non-injured side worsened and not necessarily that there was an improvement in the injured side. In this example, looking only at asymmetry could have led to an erroneous conclusion and consequently an error in planning. To further illustrate this point, Supplementary Material 2 shows the separated reliability statistics of the "good" and "bad" sides for the metrics for which asymmetry was calculated. The results show that for most cases, the reliability of the separated sides was much higher than those of the calculated asymmetry.

In this study, there were clear differences in reliability between tasks. DL_{SQUAT} and DL_{CMJ} were considered overall the easiest tasks to perform, which seemed to lead to the high reliability of these tasks, in comparison with the ones that were considered most difficult (SL_{CMJ}, DL_{LAND} and SL_{LAND}), whose reliability was mostly poor. The effect of task difficulty in reliability is also supported by the direct comparison between the single- and double-leg variations of the same tasks, where there was a clear lower reliability in the single-leg variations of the squat and countermovement jump. It is possible that between-day variability in the most difficult tasks is due to participants being more familiar with the task on the second day and/or that they decide to adopt different movement strategies as they are still searching for their optimal performance. Thus, good task familiarization could lead to improved reliability. For all tasks, we sought to extract metrics that represented different aspects of movement and each metric seemed to be affected differently by the between-day variability in how participants perform the task. Within the asymmetry metrics, there didn't seem to be a consistent type of metric that always resulted in high or low reliability. Therefore, it is advisable to verify the reliability of all metrics, considering the phase (eccentric, concentric or landing), the type (e.g., power or force) and the method of discretization (e.g., peak or mean).

Reliability of force measurements have been investigated previously, albeit with samples containing only healthy participants and using different metrics (Clark et al. 2010; Alenezi et al. 2014; Baltich et al. 2014; Schwartz et al. 2017; Heishman et al. 2020; Byrne et al. 2021). Studies have found good to excellent ICCs for several single-leg balance metrics (Clark et al. 2010; Baltich et al. 2014), excellent ICCs for vertical ground reaction force during single-leg squats (Alenezi et al. 2014), good for force and stabilization time in double and single-leg landings (Schwartz et al. 2017; Byrne et al. 2021) and good to excellent reliability in several force-derived metrics in countermovement jumps (Heishman et al. 2020). These findings strengthen our hypothesis that asymmetry metrics may not be as reliable as individual values for each side, despite its greater clinical application.

Because injured participants are usually the target of these assessments and most reliability literature is focused on healthy individuals, we decided to also include patients that had lower-limb injuries in our sample (Meshkati et al. 2011; Alenezi et al. 2014; Malfait et al. 2014; Byrne et al. 2021). We expected they would present higher variability in task performance (Baida et al. 2018) which would be reflected in lower reliability scores. Although we did not directly compare the injured and non-injured groups due to a small sample, there were some important takeaways from the observation of the reliability scores stratified by injury status. During the first two tasks ($SL_{BALANCE}$ and DL_{SQUAT}), the injured group generally presented higher ICCs and SEMs. However, during the SL_{SQUAT}, the non-injured group was the one that presented lower reliability scores. In the SL_{SQUAT}, the non-injured group squatted deeper than the injured group (22 vs 19 cm on average), which was not seen on the DL_{SQUAT} (31 vs 33 cm on average). As greater squat depths lead to lower balance, the variability in the execution of the task due to balance requirements may explain these differences (Talarico et al. 2019). Similarly, the injured group also showed less balance in the $SL_{BALANCE}$ task (770 vs 649 mm in total excursion), indicating there was more need to regain balance, which is more likely to differ between days than when the task is executed with less movement. As the load continued to progress, differences were not as clear, being metric-dependent and likely being more heavily influenced by the number of participants that were able to complete the task. Although we were not able to statistically show in our results, it seems likely that the reliability of a given metric is affected by injury status, given that it can affect the overall performance as well as require specific compensations to mitigate pain, which, in turn may vary depending on the day of assessment.

A few other limitations should be mentioned. There are countless metrics that can be extracted from each task, so our findings for the selected metrics are not guaranteed to be replicated in others (although several metrics would be correlated because they all come from the same ground reaction force measurement). Finally, our sample of injured participants contained mainly ankle and foot injuries. A larger variation of injury location and characteristic may give a more comprehensive view of what to expect when evaluating different patients in clinical settings.

7. CONCLUSION

Reliability of force-derived metrics during tasks with increasing loads is dependent on the task, the metric and the participants' injury status. Within the limitations of the current sample size, we found that double-leg squats and countermovement jumps presented the best reliability statistics while landings and single-leg jumps presented poor reliability statistics. Although asymmetry metrics are commonly used in clinical practice, in most instances they did not present high reliability scores. Finally, injured participants may display lower reliability values than non-injured ones, however this trend is task dependent. Given that reliability of force-based metrics, particularly asymmetry, was considered poor in several cases, caution should be taken when using their results for clinical decision-making.

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9. SUPPLEMENTAL MATERIALS

Supplementary Material 1 – Comparison between the Double-Leg Countermovement Jump metrics automatically calculated by the VALD ForceDecks software and re-calculated using the raw data. Statistics were calculated by comparing the values from each of the three trials and by comparing the mean of the three trials. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

	Day 1	Day 2	ICC	LOWER 95% CI	UPPER 95% CI	SEM
JUMP HEIGHT (FLIGHT TIME) [CM]						
Individual trials	20.1	21.3	0.982	0.923	0.993	0.9
Means	19.5	20.9	0.978	0.884	0.993	1.1
COUNTERMOVEMENT DEPTH [CM]						
Individual trials	-29.1	-29.5	0.912	0.855	0.948	2.5
Means	-28.7	-29.3	0.891	0.746	0.956	2.7
PEAK LANDING POWER [W]						
Individual trials	5476.2	5587.7	0.965	0.941	0.979	436.6
Means	5517.8	5731.4	0.944	0.866	0.978	552.8
ECCENTRIC MEAN FORCE (RIGHT) [N]						
Individual trials	373.6	373.7	0.986	0.976	0.992	9.5
Means	371.5	372.8	0.992	0.980	0.997	7.1
ECCENTRIC MEAN FORCE (LEFT) [N]						
Individual trials	355.2	354.3	0.988	0.980	0.993	13.3
Means	354.3	352.1	0.993	0.983	0.997	7.2
CONCENTRIC MEAN FORCE (RIGHT) [N]						
Individual trials	598.4	601.1	0.998	0.997	0.999	6.5
Means	593.7	594.5	0.998	0.994	0.999	7.9
CONCENTRIC MEAN FORCE (LEFT) [N]						
Individual trials	575.2	578.2	0.998	0.997	0.999	6.8
Means	569.0	569.6	0.996	0.990	0.998	10.1
PEAK LANDING FORCE (RIGHT) [N]						
Individual trials	1842.8	1875.4	0.930	0.884	0.959	168.1
Means	1843.1	1873.4	0.977	0.942	0.991	93.5
PEAK LANDING FORCE (LEFT) [N]						
Individual trials	1766.5	1816.4	0.940	0.899	0.964	154.3
Means	1771.5	1817.8	0.976	0.939	0.991	83.6

Supplementary Material 2 – Reliability statistics for the asymmetry metrics and for their corresponding “bad” (injured or non-dominant) and “good” (non-injured or dominant) sides. Red = poor; Yellow = moderate; Blue = good and Green = Excellent reliability.

SINGLE-LEG BALANCE								
	Day 1	**Day 2**	**p**	**ICC**	**Lower 95% CI**	**Upper 95% CI**	**SEM**	**SEM%**
COP RANGE - ANTERIOPOSTERIOR								
Asymmetry (%)	0.1	39.9	0.024	0.418	-0.023	0.744	34.9	
“Bad” side (mm)	39.8	44.1	0.325	0.708	0.344	0.890	9.7	23.2
“Good” side (mm)	45.1	37.8	0.336	0.305	-0.211	0.694	16.9	40.8
COP RANGE - MEDIOLATERAL								
Asymmetry (%)	-13.3	21.0	0.006	0.000	-0.176	0.308	21.2	
“Bad” side (mm)	33.5	35.5	0.186	0.900	0.735	0.965	3.5	10.0
“Good” side (mm)	36.2	26.5	0.065	0.322	-0.101	0.684	11.0	35.1
MEAN COP VELOCITY								
Asymmetry (%)	5.8	10.0	0.649	0.265	-0.254	0.671	20.0	
“Bad” side (mm/s)	37.5	34.4	0.203	0.802	0.524	0.928	5.4	14.9
“Good” side (mm/s)	68.2	50.7	0.152	0.877	0.677	0.957	26.5	44.6
TOTAL COP EXCURSION								
Asymmetry (%)	5.6	10.0	0.631	0.273	-0.246	0.676	19.7	
“Bad” side (mm)	706.3	666.9	0.365	0.902	0.743	0.965	96.9	14.1
“Good” side (mm)	791.5	660.9	0.035	0.644	0.162	0.873	117.7	16.2

DOUBLE-LEG SQUAT								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
ECCENTRIC PEAK FORCE								
Asymmetry (%)	-0.5	0.4	0.731	0.829	0.630	0.927	3.5	
“Bad” side (N)	465.4	475.8	0.340	0.918	0.812	0.966	28.7	6.1
“Good” side (N)	469.5	476.1	0.562	0.913	0.802	0.964	29.6	6.3
CONCENTRIC PEAK FORCE								
Asymmetry (%)	-1.2	-0.6	0.731	0.814	0.601	0.920	4.3	
“Bad” side (N)	468.5	479.4	0.269	0.939	0.859	0.975	25.9	5.5
“Good” side (N)	476.4	485.4	0.351	0.943	0.868	0.977	25.4	5.3

SINGLE-LEG SQUAT								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
CONCENTRIC MEAN POWER								
Asymmetry (%)	8.4	-3.1	0.072	0.503	0.111	0.767	13.9	
“Bad” side (W)	217.4	219.2	0.864	0.857	0.677	0.942	38.0	17.4
“Good” side (W)	210.5	228.1	0.333	0.812	0.587	0.922	46.3	21.1
CONCENTRIC PEAK POWER								
Asymmetry (%)	4.8	-2.4	0.167	0.514	0.127	0.773	11.6	
“Bad” side (W)	341.9	360.0	0.637	0.808	0.579	0.920	99.3	28.3
“Good” side (W)	348.7	361.9	0.725	0.837	0.636	0.933	93.8	26.4
ECCENTRIC MEAN POWER								
Asymmetry (%)	5.4	-3.7	0.141	0.260	-0.152	0.615	15.5	
“Bad” side (W)	169.4	178.2	0.728	0.879	0.722	0.951	28.3	16.3
“Good” side (W)	161.0	184.4	0.214	0.822	0.605	0.926	34.3	19.9
MAXIMUM NEGATIVE DISPLACEMENT								
Asymmetry (%)	4.9	-1.4	0.319	0.064	-0.401	0.506	15.7	
“Bad” side (cm)	-20.9	-20.7	0.954	0.785	0.535	0.910	3.8	-18.0
“Good” side (cm)	-20.7	-20.4	0.310	0.950	0.880	0.980	1.8	-8.6

DOUBLE-LEG COUNTERMOVEMENT JUMP								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
PEAK LANDING FORCE								
Asymmetry (%)	-6.5	-2.8	0.365	0.470	0.065	0.745	10.8	
“Bad” side (N)	1891.5	1828.7	0.383	0.872	0.716	0.946	189.8	10.2
“Good” side (N)	1750.1	1749.1	0.992	0.743	0.474	0.887	248.5	14.2
CONCENTRIC MEAN FORCE								
Asymmetry (%)	-2.6	-1.5	0.376	0.783	0.544	0.906	3.2	
“Bad” side (N)	574.7	583.5	0.413	0.966	0.921	0.986	28.3	4.9
“Good” side (N)	591.3	593.3	0.887	0.944	0.870	0.977	36.7	6.2
ECCENTRIC MEAN FORCE								
Asymmetry (%)	-2.7	-1.6	0.611	0.721	0.435	0.876	5.9	
“Bad” side (N)	357.6	360.5	0.532	0.976	0.943	0.990	12.3	3.4
“Good” side (N)	370.1	368.0	0.634	0.981	0.955	0.992	11.3	3.1

SINGLE-LEG COUNTERMOVEMENT JUMP								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
JUMP HEIGHT								
Asymmetry (%)	-0.3	3.8	0.449	0.131	-0.362	0.570	12.5	
“Bad” side (cm)	7.9	7.4	0.290	0.926	0.824	0.970	1.1	14.5
“Good” side (cm)	7.6	6.8	0.107	0.877	0.706	0.950	1.3	17.9
COUNTERMOVEMENT DEPTH								
Asymmetry (%)	-5.5	7.9	0.146	0.261	-0.179	0.641	21.0	
“Bad” side (cm)	-14.5	-15.1	0.717	0.698	0.382	0.871	4.1	-27.8
“Good” side (cm)	-14.8	-13.9	0.576	0.641	0.291	0.842	3.9	-27.3
PEAK LANDING POWER								
Asymmetry (%)	1.5	3.6	0.749	0.000	-0.476	0.476	13.6	
“Bad” side (W)	2647.0	2499.8	0.407	0.755	0.481	0.896	345.7	13.4
“Good” side (W)	2641.6	2483.6	0.248	0.787	0.541	0.911	345.8	13.5
PEAK LANDING FORCE								
Asymmetry (%)	1.5	4.2	0.485	0.179	-0.316	0.602	8.8	
“Bad” side (N)	2280.8	2165.6	0.264	0.716	0.417	0.878	174.8	7.86
“Good” side (N)	2278.4	2173.0	0.169	0.786	0.537	0.91	191.7	8.61
CONCENTRIC MEAN FORCE								
Asymmetry (%)	-1.2	0.3	0.158	0.410	-0.032	0.733	2.5	
“Bad” side (N)	947.5	949.8	0.84	0.963	0.91	0.985	37.9	4.00
“Good” side (N)	974.4	960.4	0.354	0.962	0.908	0.985	38.7	4.00
CONCENTRIC MEAN POWER								
Asymmetry (%)	-5.9	1.5	0.022	0.307	-0.081	0.655	6.9	
“Bad” side (W/kg)	9.1	9.0	0.557	0.865	0.693	0.945	1.1	11.6
“Good” side (W/kg)	9.5	8.8	0.090	0.849	0.643	0.939	1.1	11.5

DOUBLE-LEG LANDING								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
PEAK LANDING FORCE								
Asymmetry (%)	-11	-16.9	0.416	0.170	-0.326	0.596	17.0	-
“Bad” side (N)	2523.8	2158.8	0.205	0.836	0.594	0.941	42.8	1.8
“Good” side (N)	2804.3	2651.6	0.335	0.827	0.574	0.938	333.1	12.2

SINGLE-LEG LANDING								
	Day 1	**Day 2**	**p**	**ICC**	**LOWER**	**UPPER**	**SEM**	**SEM%**
TIME TO STABILIZATION								
Asymmetry (%)	23.5	11.0	0.662	0.000	-0.621	0.621	41.4	
“Bad” side (s)	6.0	1.0	0.729	1.000	0.998	1.000	0.3	8.5
“Good” side (s)	0.8	0.9	0.604	0.706	0.247	0.912	0.2	24.5
PEAK LANDING FORCE								
Asymmetry (%)	-7.6	-1.0	0.132	0.594	0.005	0.883	7.1	
Asymmetry (%)	3024.1	3297.3	0.179	0.986	0.949	0.996	153.0	4.9
“Bad” side (N)	3803.1	3364.5	0.004	0.872	0.158	0.970	207.0	5.8

THESIS CONCLUSION

Functional tasks are useful tools to evaluate how both athletes and patients move, observing how someone might be progressing with a rehabilitation program and possibly identifying who is at risk of injury. Having quantifiable results also allows clinicians, athletic trainers and coaches to develop more precise rehabilitation and injury prevention strategies. However, the correct interpretation of these results obtained with functional tasks depends on several factors, such as task variations, choice of metric and reliability of the measurements. This thesis sought to add to the knowledge concerning these factors, particularly about the practical application of current and future findings by highlighting concerns regarding the use of functional task-based results without the adequate understanding of the context.

Four experimental studies were conducted in order to address five specific aims. Aims 1 and 2 were achieved through the first study, finding that both task type and movement speed can influence several metrics commonly used to assess movement kinematics, albeit with small absolute difference in degrees. Aim 3 was achieved through the second study, finding that the relationship between muscle activation metrics and kinematics during functional task is muscle, metric and task dependent. Aim 4 was achieved through the third study, finding that stiffness and spatiotemporal parameters were not able to discriminate between people with different levels of running experience, suggesting that the increased injury rate in less-experienced runners is likely not explained by different gait patterns. Finally, aim 5 was achieved through the fourth study, finding that the reliability of force-derived metrics during tasks with increasing loads is dependent on the task, the metric and the participants' injury status and that several metrics were not sufficiently reliable.

There are limitations to this thesis that need to be considered. Besides the individual study limitations, which are mentioned within their text, two main points need to be considered. Firstly, although several relevant examples were mentioned, the thesis was not able to identify and/or quantify the studies that have failed to include nuance in the comparison to other studies that have used different task variations or metrics. In addition, it was also unable to identify how many studies did choose to use different task variations for their assessment and how they compare to those that elected to use only one. Accomplishing these steps might result in an even better understanding of the magnitude of the problem within the sports and exercise medicine literature. Secondly, each study included in this thesis was able to address a small component of the influence of different factors on results of functional tasks. However, there is an infinite number of combinations of task variations, instrumentations, measurements, metrics and populations available, with each combination possibly resulting in different findings. In the studies included in the thesis, we chose to focus on dependent variables and tasks that have been previously associated with injuries. This approach was chosen in order to improve the relevance of the studies. However, there are certainly other combinations of tasks and dependent variables that would be relevant for different conditions and also have clinical importance. In particular, the use of whole time-series data as opposed to discretization could provide more detailed, albeit also more complex, information regarding the use of functional task biomechanics.

Finally, it is important to note that the studies included in the thesis follow the trend in the literature to consider each dependent variable as independent. However, the body works as a kinetic chain and the movement in one joint/segment almost certainly affects the others, suggesting that they may not be independent. More complex analysis in future studies should take this dependance into account, which can be helped by the implementation of machine learning and artificial intelligence into the biomechanics literature. Nonetheless, the choice of which metric to use as a dependent variable for a given study will remain difficult. Typically, studies choose metrics that have been found to be statistically significant in other studies or that have been shown to be relevant for a particular condition. There are other approaches that can be taken, including conducting regression analysis and finding the source of variability in a given task or by conducting smaller pilot studies where the effect sizes will dictate the choice of dependent variable.

Despite the limitations, taken together, the thesis findings support the idea that results are highly dependent on many components that need to be taken into account when using functional tasks for evaluations. Therefore, future research should: (1) report in detail all the components of the task employed and (2) disclaim that their findings are not necessarily going to be reflected if using other task variations or metrics and (3) be careful when comparing their findings with similar studies that may have differed on task variations or metrics.

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