



UNIVERSITÀ
DEGLI STUDI
DELL'AQUILA



DISCAB
Dipartimento di Scienze
Cliniche Applicate
e Biotechnologiche

UNIVERSITY OF L'AQUILA

Department of Biotechnological and Applied Clinical Sciences

Doctoral Program in Experimental Medicine

Curriculum in “Basic and Clinical Neurosciences”

XXXVIII cycle

***Kinematic and kinetic laboratory assessment in
healthy and special populations***

MEDF-01/B — Methods and Didactics of Physical Activities

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Academic Year 2024/2025

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Abstract

Traditional biomechanical assessments often rely on single-parameter or maximal outcome measures, which may limit the ability to describe neuromuscular control strategies, coordination patterns, and functional adaptations. This thesis hypothesized that assessment approaches using dynamic tasks, bilateral execution, and the integration of kinetic, kinematic, and electromyographic variables provide a more sensitive and functionally meaningful evaluation of human movement in clinical and athletic populations.

A series of experimental studies was conducted in individuals with autism spectrum disorder (ASD), persons with multiple sclerosis (PwMS), athletes after anterior cruciate ligament reconstruction (ACL-R), rhythmic gymnastics athletes, volleyball players, and longitudinal case studies. Postural control was assessed using static and dynamic balance tasks under unilateral and bilateral conditions, with synchronized center of pressure and surface electromyography recordings. Locomotion was evaluated through angle–angle diagrams synchronized with EMG to examine inter-joint coordination and neuromuscular organization. Neuromuscular performance was assessed via force–time profile analysis during strength and jump tasks, including unilateral and bilateral maximal voluntary isometric contractions (MVC), handgrip assessments, and bilateral Drop Vertical Jump evaluations. Reliability was quantified using intraclass correlation coefficients (ICC), and sensitivity and specificity using Receiver Operating Characteristic (ROC) analysis, alongside inferential statistics for between-group and task-related differences.

Dynamic balance tasks, especially under bilateral conditions, were more sensitive than static or unilateral assessments in detecting postural and neuromuscular alterations ($p < 0.05$), with high reliability in bilateral dynamic conditions ($ICC = 0.86–0.95$). Angle–angle diagrams synchronized with EMG showed high sensitivity and specificity in identifying gait coordination alterations in ASD and PwMS ($ROC\ AUC = 94.4\%$), revealing changes not captured by conventional kinematic metrics ($p < 0.05$). Longitudinal observations confirmed that these integrated measures detect gradual functional adaptations. Force–time profile analysis demonstrated that bilateral MVC tests were more sensitive and reliable ($ICC = 0.89–0.94$) than unilateral tests in detecting subtle neuromuscular alterations ($p < 0.05$) and highlighted phase-specific asymmetries and compensatory strategies during drop vertical jump and strength tasks ($p < 0.05$).

These findings support integrated, task-specific biomechanical assessments as

methodological proposals for accurately describing and interpreting neuromuscular strategies, alterations, and adaptations, beyond traditional measures. Such approaches provide flexible, context-dependent frameworks applicable across a continuum from clinical to healthy populations, offering a solid basis for individualized functional assessment, rehabilitation planning, performance monitoring, and future developments in biomechanical assessment.

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CHAPTER 1 – Biomechanical Assessment of Human Movement

1.1 Human movement as expression of the neuromuscular system

Human movement represents the functional expression of the interaction between the nervous system and the musculoskeletal system, modulated by the biomechanical constraints of the organism and by the specific demands of the motor task. It cannot be interpreted as a simple mechanical output, but rather as an emergent result of control, coordination, and adaptation processes involving multiple levels of the neuromuscular system (Latash, 2008; Winter, 2009).

From a motor control perspective, the central nervous system organizes muscle activation by considering both the mechanical properties of body segments and environmental conditions. Movement therefore reflects the interaction between neural control and musculoskeletal mechanics, which can adapt over time through neuroplastic processes induced by training or rehabilitation (Hortobágyi et al., 2021; Shumway-Cook and Woollacott, 2007). Within this framework, movement is inherently variable, and such variability represents a physiological characteristic of the system rather than a source of error (Bernstein, 1967).

According to the constraints-led approach, movement emerges from the interaction between individual characteristics, task demands, and environmental conditions, resulting in context-dependent motor strategies (Newell, 1986). Movement variability thus assumes a functional role, allowing the neuromuscular system to maintain performance and adapt to different operational conditions. Alterations in these adaptive strategies have been associated with pathological conditions or reduced motor control capacity (Stergiou and Decker, 2011). From a neuromechanical perspective, movement quality depends not only on the magnitude of force production but also on its temporal modulation and intermuscular coordination. Variables such as the rate of force development provide a relevant example of the integration between neural and mechanical components in determining motor performance (Maffiuletti et al., 2016). Considering human movement as an integrated expression of the neuromuscular system is important for biomechanical assessment. The analysis of motor behavior allows indirect insight into the functioning of control mechanisms, the strategies adopted by the individual, and their capacity to adapt to different tasks and contexts, providing meaningful information in both clinical and sports settings (McGill, 2016; McGinnis, 2020).

1.2 Role of movement assessment in clinical and sport settings

Human movement assessment is a fundamental tool for studying neuromuscular function. Movement reflects the interaction between neural control, muscle-tendon properties, and mechanical constraints of the body and environment (Enoka, 2015; Winter, 2009). Analysis of movement allows insight into motor control strategies, intersegmental coordination, and the system's ability to adapt to different task demands (Latash and Singh, 2024).

In clinical settings, movement assessment helps identify functional alterations that may not appear in standard clinical tests. Altered motor patterns, compensatory strategies, and neuromuscular asymmetries can persist even without pain, serving as sensitive indicators of motor system dysfunction (McGill, 2016; Shumway-Cook and Woollacott, 2007). Biomechanical evaluation quantifies functional recovery and supports clinical decisions on rehabilitation progression and return to daily or sport activities (McGinnis, 2020; Shumway-Cook and Woollacott, 2007).

In sports, movement assessment is critical for optimizing performance (McGinnis, 2020). Athletic ability depends not only on strength or endurance, but also on movement quality and neuromuscular organization. Biomechanical analysis identifies technical inefficiencies, functional asymmetries, and movement strategies that may increase injury risk (Bahr and Krosshaug, 2005; McGinnis, 2020). Quantifying these aspects supports training optimization and injury prevention (Lees, 2002).

Movement evaluation can be acute or longitudinal. Acute assessments help with functional screening and short-term decisions. Longitudinal assessments track adaptations from training or rehabilitation, revealing dose–response relationships over time (Enoka, 2015; Winter, 2009). Integrating kinematic, kinetic, and electromyographic data provides a more complete and functionally relevant description of neuromuscular control (Latash and Singh, 2024; McGill, 2016; Winter, 2009).

Biomechanical movement assessment is thus a key tool to understand neuromuscular function and support clinical, and sports decisions based on objective evidence. Integrated approaches overcome the limits of traditional evaluations and allow more sensitive, specific, and applicable assessments.

1.3 Evolution of biomechanical assessment over the last 100 years

The scientific investigation of neuromuscular performance began with the early physiological studies on muscle mechanics. Wallace O. Fenn first demonstrated that the energy released by a contracting muscle increases when the muscle performs mechanical work, a phenomenon later known as the *Fenn effect* (Fenn, 1924). A few years later, A.V. Hill described the characteristic relationship between muscle force and shortening velocity, showing that the force produced by a muscle decreases as contraction velocity increases (Hill, 1938). Together, these foundational studies established the physiological basis for understanding muscle mechanical behavior and provided the conceptual framework for later biomechanical analyses of human movement.

The first systematic photographic study of motion was conducted by Eadweard Muybridge in 1887, who captured sequential images of human and animal movement (Muybridge, 1979). This early work introduced the concept of quantifying movement, paving the way for modern instrumented biomechanical analysis.

In the 1970s and 1980s, the field adopted instrumented measurements, including motion capture systems, force platforms, and surface electromyography. These tools enabled quantitative assessment of joint angles, muscle activity, and ground reaction forces (Sutherland, 2005). Integration of kinetic and energy analyses provided deeper understanding of both normal and pathological gait patterns.

During the 2000s and 2010s, biomechanical evaluation became widely applied in clinical and sports contexts. Modern approaches combine kinematic, kinetic, and electromyographic data to create a multi-dimensional view of movement (Payton et al., 2018). Clinicians can quantify functional deficits, monitor rehabilitation progress, and guide return-to-activity decisions (Hamill et al., 2021; Sutherland, 2005). In sports, these methods allow identification of asymmetries, optimization of technique, and injury prevention (Payton et al., 2018).

The evolution from observational to integrated instrumented assessment highlights the importance of combining multiple data streams to better describe human movement. These methods can be applied both for immediate, single-session evaluations and for longitudinal monitoring of adaptations over time, providing insights into acute responses and chronic changes in the neuromuscular system.

1.4 Acute and chronic assessment: monitoring the dose–response relationship

The assessment of human movement can be applied to study both short-term changes and long-term adaptations. Acute assessments provide information on the current organization of movement and neuromuscular control, whereas chronic assessments allow monitoring how movement evolves over repeated exposures to training or rehabilitation (Enoka, 2015; Latash and Singh, 2024). Repeated measurements are essential to observe how the neuromuscular system responds to interventions.

Acute assessments refer to measurements performed within a single testing session. They describe the current functional state of the neuromuscular system. Common examples include athletic tasks, such as vertical jump tests (e.g., drop vertical jump) (Di Giminiani et al., 2020; Di Giminiani and Petricola, 2016; La Greca et al., 2024; Passacantando et al., 2025) or running performance (Ferber and Macdonald, 2014), and fundamental movement tasks, such as postural balance and gait analysis (Di Giminiani et al., 2024, 2022; La Greca et al., 2025a, 2025b; Winter, 1995). These evaluations can identify functional deficits, asymmetries, or movement inefficiencies and support short-term decisions in clinical and sport settings, such as adjusting training load or modifying rehabilitation strategies.

Chronic assessments are aimed at monitoring changes over time, capturing neuromuscular and functional adaptations induced by repeated exposure to training or rehabilitation stimuli (Enoka, 2015; Hortobágyi et al., 2022). They are important to evaluate intervention effectiveness and to explore dose–response relationships between the applied stimulus and functional adaptations. Within this framework, repeated movement assessments allow the observation of how changes in exercise or rehabilitation dose are reflected in movement organization over time. Rather than focusing only on isolated performance outcomes, biomechanical evaluation captures functional adaptations as gradual changes in motor strategies and neuromuscular coordination. In this sense, movement-based measures provide a useful approach to explore dose–response relationships in both clinical and sport settings.

Even simple tools can provide valuable information. In our previous longitudinal study, investigating the effects of different training programs on posterior kinetic chain flexibility on older adults, repeated sit-and-reach tests were used to monitor changes over time (La Greca et al., 2022). Although this test does not provide detailed kinematic data, it allowed consistent monitoring of functional adaptations and helped interpret individual responses to the exercise dose.

This shows that simple assessment tools, when used systematically, can support decisions about intervention progression and exercise dosing.

Alongside simple tools, chronic assessments can integrate kinematic, kinetic, and electromyographic measures to provide a more complete picture of neuromuscular adaptations. A five-year case report on a person with multiple sclerosis demonstrated that repeated integrated biomechanical assessments capture long-term changes in locomotion and postural control, providing useful information to guide and adjust interventions (La Greca et al., 2025b). Longitudinal biomechanical evaluation thus supports evidence-informed decision-making in both clinical and applied contexts (Enoka, 2015).

Acute and chronic assessments are complementary. Acute measures describe the system's behavior at a specific time point, while chronic assessments show how movement adapts over repeated exposures. Together, they provide a clear and practical framework to study neuromuscular function and apply biomechanical assessment in clinical and sport settings.

Building on these repeated assessments, more detailed evaluations can integrate kinematic, kinetic, and electromyographic measures. This integrated approach allows a deeper understanding of neuromuscular control and prepares the ground for exploring the combined analysis of movement variables, which will be discussed in the following section.

1.5 Integration of kinematics, kinetics, and electromyography

Biomechanical assessment of human movement requires the analysis of multiple components that jointly describe how movement is produced and controlled. Human motion emerges from the interaction between neural control, muscle force production, and mechanical constraints imposed by the body and the environment (Enoka, 2015). For this reason, the integration of kinematic, kinetic, and electromyographic information represents a core methodological approach in modern movement analysis. Importantly, these measurements are most informative when collected in a synchronized manner, allowing the temporal alignment of kinematic, kinetic, and physiological signals. Synchronization enables researchers to relate muscle activation patterns to the mechanical forces produced and the resulting movement, providing a more accurate representation of neuromuscular control during functional tasks.

Kinematic analysis focuses on the description of movement geometry. It includes the

measurement of joint angles, angular velocities, accelerations, and segment trajectories. Kinematics allows researchers and clinicians to examine intersegmental coordination, movement timing, and overall movement quality (McGinnis, 2020; Winter, 2009). Through kinematic variables, it is possible to identify altered movement patterns, compensatory strategies, or changes in motor coordination associated with pathology, fatigue, or training adaptations (Enoka, 2015).

Kinetic analysis addresses the forces that generate and constrain movement. Typical kinetic variables include ground reaction forces, joint moments, and mechanical power. These variables contribute to the understanding of how mechanical work is distributed across joints during locomotion and other functional tasks (Hortobágyi et al., 2016; Zatsiorsky, 2002). In both clinical and sport contexts, kinetic measures are essential for understanding movement efficiency, mechanical loading, and potential risk factors related to injury or re-injury (Payton & Burden, 2018; Sutherland, 2005).

Surface electromyography (EMG) adds information on the neural drive to the muscles. EMG analysis allows the evaluation of muscle activation timing, amplitude, and coordination patterns during movement tasks. These measures help to identify neuromuscular strategies, co-contraction levels, and compensatory activation patterns that may not be evident from kinematic or kinetic data alone (De Luca, 1997; Merletti and Parker, 2004). EMG therefore represents a key link between neural control and mechanical output (Enoka, 2015; Latash and Singh, 2024).

The integration of kinematics, kinetics, and EMG enables a more complete interpretation of movement behavior. While each approach provides valuable information on its own, their combination allows movement to be analyzed from a complementary perspective. This multimodal assessment improves the sensitivity and interpretability of biomechanical evaluations and enhances their applicability in both clinical and sport settings.

Such an integrated framework underlies the methodological approach adopted throughout this thesis. By combining kinematic, kinetic, and electromyographic data, it becomes possible to capture the complexity of human movement and to support functional assessment across different tasks, populations, and contexts.

CHAPTER 2 – Limitations of Currently Used Assessment Methods

2.1 Sensitivity, specificity and discriminative capacity of assessment tests

Traditional assessment tests, widely used in both field and clinical settings, such as the sit-and-reach, Timed Up & Go, Sit-to-Stand, Functional Reach, or 6-Minute Walk Test (American College of Sports Medicine et al., 2026), provide simple and quick assessments of functional performance. However, when evaluating functional performance, several methodological properties determine the usefulness of an assessment test. In both clinical evaluation and sport performance analysis, a test should demonstrate validity, objectivity, reproducibility, and specificity to provide meaningful and interpretable results. Validity refers to the ability of a test to measure the characteristic it intends to assess, while objectivity describes the extent to which results are independent of the evaluator. Reproducibility (or reliability) reflects the consistency of the measurement when repeated under similar conditions.

In addition to these properties, test specificity plays a fundamental role in movement assessment. According to the principles proposed by Bosco (1999), an evaluation test should reflect the biomechanical and neuromuscular characteristics of the movement or performance being investigated. When the task used in the assessment differs substantially from the real movement demands, the ecological and practical value of the results may be reduced. This concept is particularly relevant in sport performance analysis, where task-specific tests allow a more accurate interpretation of neuromuscular strategies and functional capacity.

However, recent studies indicate that these tests may fail to detect subtle functional changes or compensatory strategies compared to more precise assessments or when specific functional information is needed (Coni et al., 2019; Gafner et al., 2021; Hulleck et al., 2022; Mancini and Horak, 2010). They are therefore less capable of capturing specific alterations or changes, particularly in populations with partial or compensated deficits. Their limited sensitivity and specificity may reduce the ability to detect meaningful differences across individuals or overtime.

Even advanced technological assessments, such as 3D motion capture, electromyography (EMG), and ground reaction force measurements, may be less effective if the protocol or

outcome measure is not carefully selected for the specific deficit or movement characteristic of interest (Enoka, 2015). For example, in postural control, fractal dimension analysis of body sway has shown higher sensitivity and specificity compared to traditional sway area measures (Masedu et al., 2013). Similarly, in locomotion, our recent study on young adults with autism spectrum disorder (ASD) showed that angle–angle diagrams could discriminate motor pattern differences between ASD and typically developing individuals more effectively than traditional angle–time analysis, with high sensitivity and specificity in ROC analysis (Di Giminiani et al., 2024).

When selecting the most appropriate test, it is important to consider the neurophysiological basis of movement, the body region assessed, and the type of movement involved (Latash and Singh, 2024). Movement is complex and individualized, and the same test may provide different insights in healthy individuals compared to those with neurological or musculoskeletal conditions. Unilateral and bilateral assessments of lower limb function, for example, may reveal distinct neurophysiological adaptations (Latash and Singh, 2024). In our studies, individuals post- anterior cruciate ligaments (ACL) reconstruction showed neuromuscular alterations in bilateral tasks (maximal voluntary isometric contractions and postural balance) that were not always detectable in conventional unilateral testing (Di Giminiani et al., 2023; La Greca et al., 2025a). Similarly, the combination of angle–angle diagrams and sEMG analyses highlighted functional differences between healthy individuals and those with multiple sclerosis (Di Giminiani et al., 2022) or ASD (Di Giminiani et al., 2024).

These observations underline that subtle, multifactorial, or compensated deficits require assessment tools that are sensitive, specific, and carefully targeted. Selecting the right test and outcome measure is essential to accurately evaluate functional status and detect meaningful changes over time, both in clinical and sports contexts.

2.2 Clinical and sport transferability of movement assessment

Selecting assessment tests that are both specific and transferable is essential in clinical and sport settings. Generic tests can provide global information about physical function, but they often lack relevance for the specific movement or task of interest, limiting their practical applicability (Sutherland, 2005). For an assessment to be transferable, it must generate information that can be directly used to guide practical decisions, such as exercise selection, load progression, or return-to-activity planning. Movement assessment

should therefore focus on outcomes that are meaningful for the context, including key elements of athletic performance, movement coordination, and task-specific motor demands, to preserve ecological validity (Payton et al., 2018). Because neuromuscular strategies are highly task- and context-dependent, aligning the assessment with real functional demands allows a more accurate interpretation of movement behavior and adaptations (Enoka, 2015). Greater specificity in test selection ultimately improves the transferability of results to real-world clinical and sport practice.

As highlighted in the previous section, non-targeted assessments, even when widely used or technologically advanced, may fail to detect subtle functional changes or compensatory strategies. When tests are not calibrated to the specific deficit or functional goal, their ability to inform clinical or sport-related decisions is limited.

Targeted assessments allow movement analysis to be more closely aligned with performance-related demands, offering insights that are potentially more transferable to training and injury prevention contexts. In our research, drop vertical jump tasks have been adopted as an example of sport-specific assessment to explore neuromuscular control and inter-limb asymmetries in athletes. In rhythmic gymnasts, comparing dominant and non-dominant limbs highlighted asymmetry patterns associated with the technical demands of the sport, suggesting how task-specific testing may inform training strategies (Passacantando et al., 2025). Similarly, in volleyball players, drop jump assessments were used to examine landing mechanics during actions resembling competitive play, illustrating how ecologically valid tasks can support performance-oriented evaluation (La Greca et al., 2024).

A similar rationale applies in clinical settings, where task specificity can influence the clinical relevance of assessment outcomes. In individuals following ACL reconstruction, bilateral tasks have been explored as an alternative to conventional unilateral tests to investigate persistent neuromuscular alterations, due to the natural bipedal control of the lower limbs. Assessments involving bilateral maximal voluntary isometric contractions and postural balance tasks illustrate how selecting functionally demanding tasks may reveal residual deficits that are not always apparent in isolated testing conditions (Di Giminiani et al., 2023; La Greca et al., 2025a). These examples emphasize that the choice of assessment task plays a central role in determining the extent to which movement evaluations can inform clinical decision-making.

Transferability does not necessarily depend on the complexity of the measurement system. Targeted assessments can also be implemented using simple tools, provided that

the task and outcome are carefully designed. Repeated monitoring of function, using either simple or advanced methods, allows clinicians and practitioners to track changes over time and adjust interventions based on individual responses.

Overall, movement assessments become clinically and sport-relevant when they are designed around specific tasks, populations, and functional goals. Such an approach enhances sensitivity and practical relevance, ensuring that assessment outcomes can effectively support decision-making in both rehabilitation and performance settings, and setting the stage for discussing the limitations of single-parameter evaluations in the following section.

2.3 Limitations of single-parameter assessment approaches

Assessing movement using a single parameter provides a limited perspective on human motor control. Human movement is inherently multidimensional, involving complex interactions between kinematics, kinetics, and neuromuscular activation (Enoka, 2015). Evaluating only one variable, such as range of motion, peak force, or reaction time, cannot fully capture these interactions and may overlook subtle compensatory strategies (Winter, 2009).

This limitation is particularly evident in populations with partial or compensated deficits. In such cases, single-variable assessments may fail to reveal functionally relevant alterations in movement patterns, limiting their utility for clinical or sports decision-making (Latash and Singh, 2024). Relying on isolated measures risks underestimating the complexity of motor control and may provide incomplete information regarding performance, recovery, or residual functional deficits.

Even when advanced measurement tools are available, such as 3D motion capture, electromyography, or force platforms, focusing on a single outcome can obscure the broader picture of neuromuscular behavior and intersegmental coordination (Enoka, 2015; Winter, 2009). Integrating multiple parameters allows for a more accurate interpretation of movement, increases sensitivity to subtle deficits, and enhances the relevance of assessment outcomes for both clinical and sport applications (Latash and Singh, 2024).

In conclusion, single-parameter assessments are limited in their ability to describe the full complexity of human movement. Recognizing this limitation highlights the need for multi-dimensional, targeted evaluation strategies that can provide actionable information,

preparing the rationale for the next section, which discusses the integration of specific biomechanical and neuromuscular measures.

2.4 Need for integrated assessment approaches

Human movement is a complex and adaptive process that cannot be adequately described by a single variable. As discussed in the previous section, mono-parametric assessments provide only a partial representation of motor behavior and often fail to capture coordination strategies and functional adaptations. Movement emerges from the interaction of multiple systems acting together, rather than from isolated mechanical outputs (Winter, 2009; Enoka, 2015). This limitation highlights the need for integrated assessment approaches.

Each biomechanical domain provides complementary information about movement. Kinematic variables describe how movement is performed, including joint coordination, ranges of motion, and temporal patterns (Cavanagh, 1987; Winter, 1995). Kinetic measures quantify the forces and loads acting on the body, offering insight into mechanical demands and load distribution (Winter, 2009; Zatsiorsky, 2002). Electromyographic signals reflect how the neuromuscular system organizes muscle activation to achieve task goals (Enoka, 2015; Latash and Singh, 2024). When considered in isolation, none of these parameters is sufficient to explain motor strategies, compensatory mechanisms, or functional adaptations. Their integration allows movement to be interpreted as a coordinated system rather than as the sum of independent variables (Latash and Singh, 2024).

From a functional perspective, integrating biomechanical parameters improves both sensitivity and interpretability. Combined analyses make it possible to detect subtle deficits that may not affect global performance outcomes but still influence movement quality and control (Payton et al., 2018). In clinical contexts, this approach supports the identification of residual impairments even when task completion appears restored. In sport settings, it enables the simultaneous evaluation of movement quality, mechanical loading, and neuromuscular control, which is essential for informed training and injury prevention decisions.

Importantly, integration should not be viewed as a purely technological choice but as a methodological one. The selection of multiple parameters must be driven by the functional goal of the assessment and the specific characteristics of the population under

investigation (Enoka, 2015; Payton et al., 2018). Within this framework, integrated assessment approaches provide the rationale for combining targeted biomechanical measures to improve the relevance and applicability of movement evaluation.

In summary, integrating biomechanical parameters is necessary to overcome the limitations of traditional mono-parametric assessments and to achieve a more complete understanding of human movement. This perspective closes the conceptual framework of the present chapter and provides a direct rationale for the methodological choices adopted in the following sections of the thesis.

2.5 Implications for research and applied practice

The previous chapters have shown that movement assessment requires specificity, sensitivity, and an integrated approach. Generic or single-parameter tests provide limited insight and may fail to capture subtle adaptations or compensatory strategies.

For research, this underscores the need to develop and validate reliable assessments that are carefully tailored to the functional deficit or performance aspect of interest. Combining kinematic, kinetic, and neuromuscular measures allows researchers to understand movement as a coordinated system rather than isolated variables, revealing subtle strategies that would otherwise remain undetected.

In applied practice, targeted and task-specific assessments guide practical decision-making. Clinicians and coaches can identify residual deficits, monitor progression, and adjust interventions more effectively. Even simple tools become valuable when used within a clear, reproducible framework that reflects real-world demands. Ecologically valid tasks increase the transferability of results, improving the relevance of assessments in rehabilitation, injury prevention, and performance contexts.

Overall, research and practice benefit from approaches that are sensitive, specific, and integrated, setting the stage for the rationale and objectives of the thesis, which will be detailed in the following section.

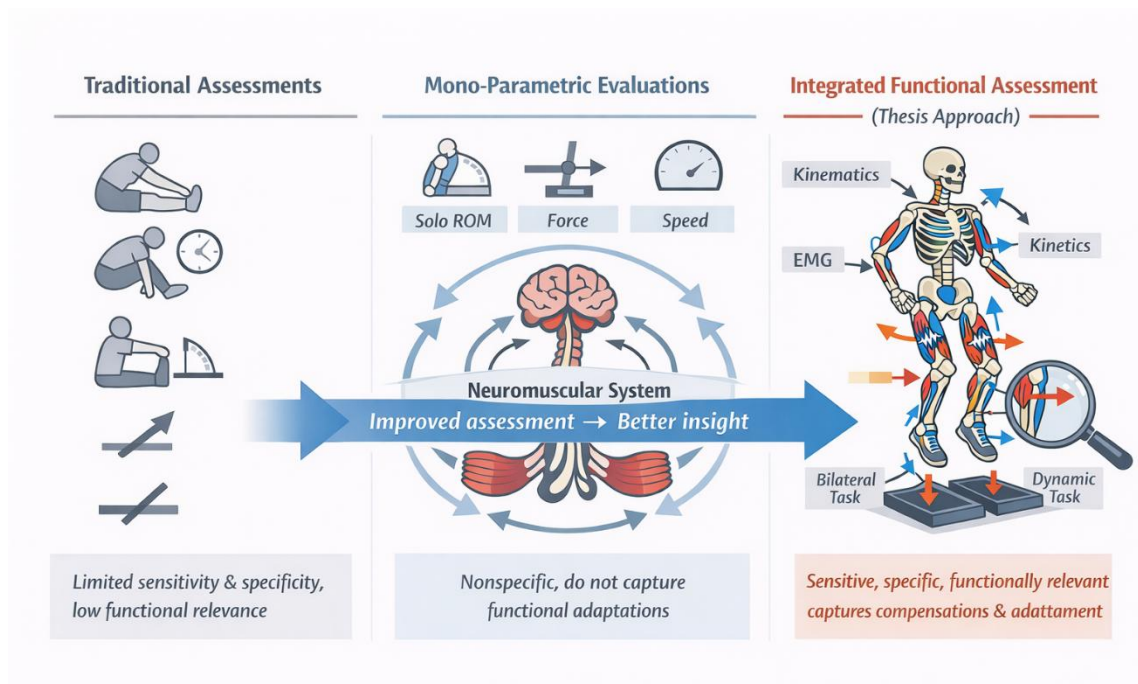


Figure 2.1. Rationale for integrated biomechanical assessment. The schematic compares three approaches to human movement evaluation. Traditional assessments (left) rely on simple, static tests (e.g., sit-and-reach, Timed Up & Go, single-joint measures) and provide limited sensitivity, specificity, and functional relevance. Mono-parametric evaluations (center) assess single parameters such as range of motion (ROM), force, or speed, but remain nonspecific and do not capture functional adaptations. Integrated functional assessment (right, thesis approach) combines kinematics, kinetics, and electromyography (EMG) during bilateral and dynamic tasks, providing a sensitive, specific, and functionally relevant characterization of movement, including compensations and adaptations. The central neuromuscular system illustrates the interplay between the central nervous system, muscles, and environment, highlighting how improved assessments lead to deeper insight into motor function.

2.6 Rationale and aims of the thesis

Movement assessments in this thesis are designed to improve specificity, sensitivity, and reliability, while ensuring practical relevance to real-world and sport-specific contexts. Previous chapters have shown that traditional or generic assessments, including single-parameter evaluations, often fail to capture the complexity of human movement. They may not detect subtle motor strategies, compensatory mechanisms, or functional adaptations, limiting their usefulness for clinical or performance applications. This underscores the need for integrated assessment approaches that combine kinematic, kinetic, and neuromuscular measures to provide a comprehensive understanding of motor

behavior.

In addition to parameter integration, it is crucial that assessment protocols reflect the characteristics and execution patterns of movements encountered in daily life or sport. Even simple or single-parameter tests are more informative when tasks are similar to functional actions, allowing the assessment under context-relevant conditions. Aligning testing with such conditions improves the interpretability of results and ensures that findings can inform practical decision-making in rehabilitation, performance optimization, and injury prevention.

The primary aim of this thesis is therefore to develop and propose assessment protocols that are both integrated and functionally relevant. The focus is not merely on combining multiple parameters, but also on designing tests that allow reproduction of real-world and sport-specific motor demands. This approach enables a more precise and context-sensitive assessment of performance and functional status. Assessments are thus sensitive, specific, reliable, and applicable, providing insights that can guide interventions and training strategies in both clinical and sport settings.

Movement assessments in this thesis are structured according to the neurophysiological principles of human movement and the characteristics of the task environment. This defines a clear methodological framework that guides the experimental work presented in the following chapters.

The selection of parameters, tasks, and protocols is driven by the need to evaluate motor function under conditions that reproduce, or closely approximate, real-world and sport-specific actions. This approach links theoretical models of movement control with empirical assessment, supporting a rigorous and practice-oriented evaluation of motor behaviour.

CHAPTER 3 – Postural Control Assessment

3.1 Postural Control: Neuromotor and Biomechanical Foundations

Postural control refers to the ability to regulate the position of the body in space in order to maintain stability. It is a fundamental function that allows individuals to remain upright during quiet stance and to control body orientation during movement (Winter, 1995). Posture describes the configuration of body segments, whereas balance is a dynamic process that requires continuous regulation to keep the body's center of mass within the base of support (Latash and Singh, 2024).

From a neurophysiological perspective, postural control emerges from the interaction of multiple neural structures, including spinal circuits, brainstem nuclei, cerebellar pathways, and cortical areas. Many postural adjustments occur automatically, without conscious control, yet they remain adaptable to task demands and environmental conditions (Enoka, 2015). This organization allows posture to be maintained while simultaneously supporting voluntary movements, highlighting the integrated nature of postural and motor control (Latash and Singh, 2024).

Postural stability depends on the continuous integration of sensory information from the somatosensory, vestibular, and visual systems. Somatosensory inputs provide information about body position and support surface characteristics, vestibular inputs signal head orientation and motion relative to gravity, and visual inputs contribute to spatial orientation and motion perception (Winter, 1995). The relative contribution of each sensory system is not fixed but changes according to task constraints and environmental context, reflecting the adaptive capacity of the postural control system (Horak and Nashner, 1986; Latash and Singh, 2024).

Biomechanically, upright stance is often described using the inverted pendulum model, where the body rotates around the ankle joint while maintaining balance over a relatively small base of support (Winter, 1995) (Figure 3.1). Within this framework, postural sway is commonly characterized using two related variables: the center of mass (CoM) and the center of pressure (CoP). The CoM represents the weighted average position of body mass, while the CoP is defined as the point of application of the resultant vertical ground reaction force acting on the body (Winter, 1995; Zatsiorsky, 2002). The nervous system

controls balance by modulating CoP trajectories to regulate CoM motion and prevent loss of stability (Latash and Singh, 2024; Winter, 1995).

Postural responses to perturbations are organized through specific corrective strategies. For small and slow disturbances, balance is primarily restored using an ankle strategy, characterized by distal-to-proximal muscle activation. Larger or faster perturbations require a hip strategy, involving proximal muscle activity to rapidly reposition the CoM (Nashner and McCollum, 1985). These strategies are not rigid patterns, but flexible solutions selected according to task demands, support conditions, and prior experience (Horak and Nashner, 1986). Central programming mechanisms allow the nervous system to adapt postural responses when sensory inputs or support surface properties change (Horak and Nashner, 1986; Latash and Singh, 2024).

Overall, postural control represents a complex, adaptive process that integrates sensory information, neural control, and biomechanical constraints. Understanding these mechanisms provides the theoretical foundation necessary to interpret postural behavior and to justify the assessment approaches discussed in the following sections.

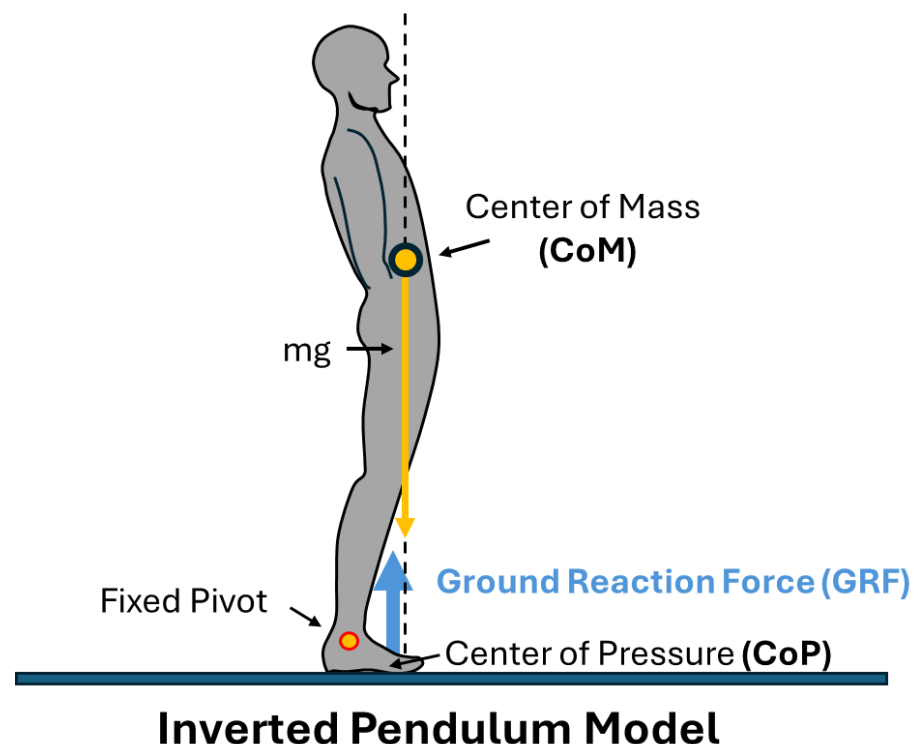


Figure 3.1. Inverted pendulum model of human balance. The human body is modelled as a rigid rod pivoting at the ankle. The Center of Mass (CoM) experiences the downward gravitational force (mg), while the Ground Reaction Force (GRF) acts upward at the Center of Pressure (CoP) on the foot. The relative positions of CoM and CoP illustrate balance control.

3.2 Methods for postural control assessment

Postural control can be assessed using a wide range of methods, spanning from simple clinical tests to instrumented approaches that provide quantitative information. In clinical and sport settings, balance is often evaluated through standardized functional tests, such as single-leg stance tasks, the Y Balance Test, or the Berg Balance Scale (Horak, 2006; Paillard and Noé, 2015). These assessments are designed to capture overall postural stability and functional balance capacity, and they are commonly used for screening, monitoring progression, and identifying general balance deficits (Horak, 2006; Sibley et al., 2015). Their main strengths are practicality, low cost, and ease of implementation across different populations, including healthy individuals, athletes, and patients. However, because they rely primarily on task performance or scaled scores, they provide limited information about the mechanisms underlying postural control and the strategies used to maintain stability (Sibley et al., 2015).

Instrumented assessments allow a more detailed analysis of postural behaviour. Force platforms enable precise measurement of postural sway through the analysis of the center of pressure trajectories, which reflect the neuromuscular actions used to regulate the body's center of mass within the base of support (Horak and Nashner, 1986; Nashner and McCollum, 1985; Winter, 1995). CoP analysis provides quantitative parameters describing sway magnitude, timing, and structure, allowing identification of postural strategies and compensatory behaviours that are not evident in clinical tests. When CoP measures are combined with surface electromyography, postural sway can be directly related to muscle activation patterns and sensory integration processes involved in balance control (Horak and Nashner, 1986; Latash and Singh, 2024).

The selection of postural control assessment methods should be guided by the objectives of the evaluation. Clinical tests are suitable for functional screening and routine monitoring, whereas instrumented and integrated approaches allow a more detailed characterization of postural behaviour and control strategies. Within this framework, different methodological choices offer complementary perspectives on balance control, forming the basis for a focused examination of center of pressure measures and their integration with surface electromyography.

3.2.1 Center of pressure analysis

The analysis of the center of pressure represents the gold standard for the quantitative

assessment of postural control. CoP is measured using force platforms and corresponds to the point of application of the resultant vertical ground reaction force acting on the body (figure 3.2) (Winter, 1995). Because it reflects the continuous interaction between the neuromuscular system and the support surface, CoP analysis provides an objective and reproducible measure of balance control and postural stability (Enoka, 2015; Latash and Singh, 2024; Winter, 1995).

Postural control does not imply maintaining the body in a fixed position but rather regulating the body's center of mass within the base of support. In this context, CoP and CoM represent two distinct but functionally linked variables. The CoM describes the global mechanical state of the body, whereas the CoP reflects the corrective actions generated by the neuromuscular system to control CoM motion. The displacement of the CoP over time is commonly referred to as postural or body sway and represents the observable outcome of balance regulation processes (Winter, 1995).

Different characteristics of the body sway have been studied, including path length, directional components, sway amplitude, the area covered by the sway trajectories, and several spatial and temporal parameters (velocity, acceleration) (Latash and Singh, 2024). These parameters describe how the body maintains equilibrium and how corrective strategies are organized. Importantly, CoP path length can be analysed not only as a total value but also along the antero-posterior (AP) and medio-lateral (ML) directions. This distinction has functional relevance, as AP and ML sway are controlled through partly different biomechanical and neuromuscular mechanisms (Winter et al., 1996).

The control of AP sway is primarily associated with ankle torque modulation and sagittal-plane dynamics, whereas ML sway relies more heavily on hip and trunk control and lateral weight-shifting strategies. Experimental evidence suggests that ML control is often more demanding from a neuromotor perspective and more sensitive to deficits or adaptations, particularly in challenging or pathological conditions (Winter et al., 1996). Neurophysiological studies have further shown that voluntary modulation of sway direction is associated with direction-specific cortical activation, supporting the idea that AP and ML sway reflect distinct control processes (Slobounov et al., 2008).

From a neuromechanical perspective, CoP measures represent the mechanical output of postural control rather than its direct neural organization. CoP trajectories emerge from the combined effects of muscle activation, sensory integration, and automatic corrective responses. For this reason, CoP analysis provides essential but incomplete information when considered in isolation. Its interpretation benefits from integration with measures

that capture the underlying neuromuscular activity, such as surface electromyography. Overall, CoP analysis offers a robust and sensitive method for quantifying body sway and postural strategies. Understanding its biomechanical meaning and directional components is fundamental for interpreting balance behaviour and provides the basis for integrated approaches to postural control assessment,

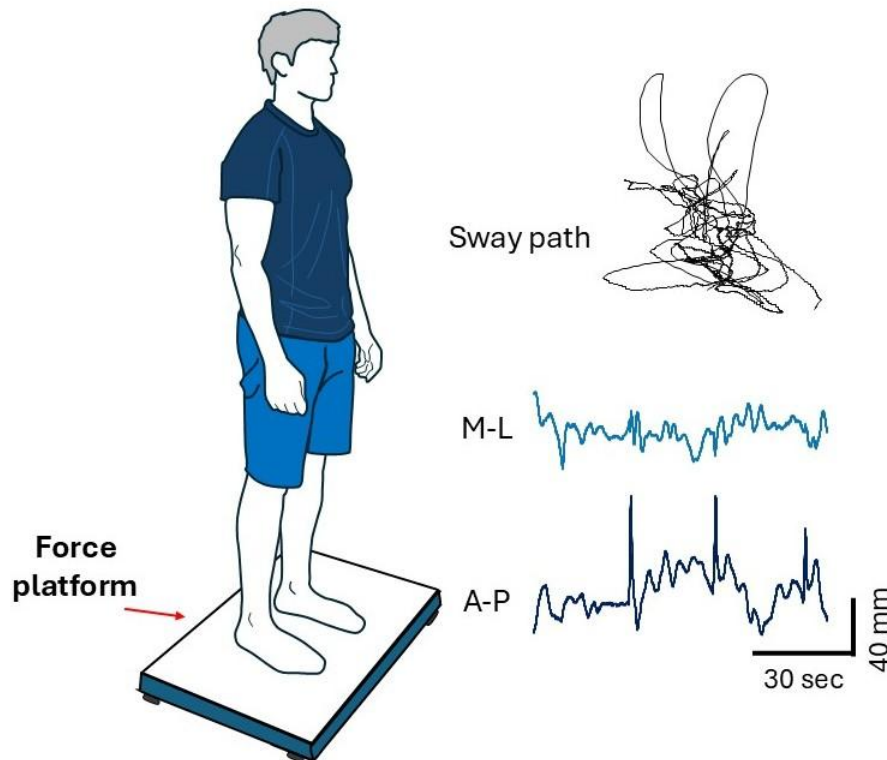


Figure 3.2. Typical body sway assessment setup. Body sway assessment in dynamic upright standing conditions. Total body sway path, medio-lateral (M-L) and anterior-posterior (A-P) displacements of the center of pressure.

3.2.2 Integration with surface electromyography

Center of pressure analysis describes the mechanical outcome of postural control but does not provide direct information on the neuromuscular processes that generate it. The integration of CoP measures with surface electromyography (EMG) allows investigation of the muscle activity underlying postural stabilization, linking mechanical behavior to motor control strategies (Winter, 1995; Latash & Singh, 2024).

Postural control relies on largely automatic motor behaviors, organized through spinal, brainstem, and supraspinal structures. EMG enables the analysis of muscle activation timing, amplitude, and coordination, providing insight into how postural muscles are

modulated to maintain stability or respond to perturbations (Horak, 2006; Nashner and McCollum, 1985; Winter, 1995). This information complements CoP data by clarifying whether observed sway patterns result from efficient control, compensatory activation, or altered neuromuscular organization (Enoka, 2015; Latash and Singh, 2024).

The combined analysis of CoP and EMG is particularly relevant for identifying postural strategies, such as ankle- and hip-dominated control, which may produce similar mechanical outputs but differ substantially at the neuromuscular level (Horak and Nashner, 1986; Nashner and McCollum, 1985). By integrating these measures, postural assessment moves beyond descriptive sway metrics toward a more complete characterization of motor behavior, consistent with contemporary models of movement organization (Latash and Singh, 2024; Winter, 1995).

3.3 Static and dynamic balance

Postural control can be assessed under static or dynamic conditions, each providing complementary information about the neuromuscular system (Winter, 1995). Static balance tasks typically involve maintaining an upright standing position on a stable support surface. They require low-level neuromuscular regulation to keep the body's center of mass within the base of support through small and continuous postural adjustments (Winter, 1995). These tasks are widely used in both clinical and sport settings because of their simplicity, reproducibility, and ability to provide a rapid overview of postural stability (Enoka, 2015; Latash and Singh, 2024). However, static tasks may not fully reveal persistent alterations in neuromuscular control or subtle deficits, which often emerge only when the postural system is challenged by more demanding conditions.

Dynamic balance tasks, by contrast, require continuous postural adjustments in response to movement of the body or the support surface (Winter, 1995). These tasks engage both anticipatory and reactive postural strategies and demand greater sensorimotor integration. They also involve coordinated activation of multiple muscle groups to maintain equilibrium (Horak and Nashner, 1986; Latash and Singh, 2024; Winter, 1995). As a result, dynamic assessments provide more detailed information on functional postural control, allowing the identification of deficits or compensatory strategies that may remain undetected during static testing (Di Giminiani et al., 2024; La Greca et al., 2025a; Latash and Singh, 2024).

In the literature, postural control is most often evaluated using static conditions or through assessments that are not synchronised with neuromuscular activity. This methodological

choice limits the ability to fully interpret the underlying control strategies, particularly during tasks that better reflect real-life or sport-specific demands.

In the studies conducted for this thesis, postural control was assessed under both static and dynamic conditions to determine whether dynamic tasks could highlight neuromuscular and postural control alterations not observable during static assessments. This rationale is based on the premise that the neuromuscular organization required for dynamic tasks is more complex, requiring refined coordination, anticipatory adjustments, and multijoint control (Winter, 2009). Specifically, static and dynamic tasks were used to investigate differences in postural control between special populations and matched healthy controls, as well as to evaluate the effects of targeted interventions. This approach supports the development of assessment protocols that are not only reliable and reproducible, but also functionally relevant to real clinical contexts and sport-specific scenarios.

In the literature, it appears common practice to evaluate postural stability through static tests (Howells et al., 2011; Noamani et al., 2023; Rizzato et al., 2025). Under dynamic conditions, several imbalance systems have been proposed. Some studies employ tilting or translating support surfaces (Nashner and McCollum, 1985; Paterno et al., 2013; Petró et al., 2017; Stensdotter et al., 2016; Travers et al., 2013; Winter, 2009), whereas others involve voluntary movements of body segments during assessment, such as arm displacements, or require participants to relocate their center of pressure toward specific targets using real-time visual feedback (e.g., HUMAC Balance System, Technobody system) (Meiners and Loudon, 2020; Sahoo and Sahu, 2021). The lack of standardized dynamic assessment protocols may partially explain the variability observed across studies.

Winter (1995) employed different types of perturbations, including segment-specific disturbances (e.g., arm, leg, or trunk) and tilting platforms, to investigate the regulation of dynamic balance, often without the participant's prior awareness. There is general agreement that the central nervous system seeks to simplify the control of posture by selecting task-specific strategies depending on the body segment involved in the perturbation (Nashner and McCollum, 1985; Winter, 1995). According to Winter (1995), a critical aspect in defining the experimental setup is whether the selected paradigm accurately represents real-world perturbations.

Moving platforms capable of rapid shifts or tilts are frequently used to assess dynamic balance. However, the extent to which support-surface perturbations reflect real-life or

sport-specific conditions remains uncertain. In daily activities and athletic contexts, individuals are more commonly exposed to trunk perturbations or external forces, such as pushing or physical contact, rather than sudden movements of the support surface. Considering these, the experimental setup adopted in this thesis employs an individualized pendulum system that applies controlled and reliable trunk perturbation (ICC ranging from 0.70 to 0.95) (La Greca et al., 2025a), allowing the investigation of dynamic postural control under conditions that more closely resemble real-life and sport-related challenges.

Methods

Setup for postural balance assessment

The postural balance was evaluated by measuring the body sway during upright standing using the force platform (Muscle-Lab 4000e, Ergotest Technology, Langesund, Norway). White walls were positioned 1.5 meters away from the platform, and a 1-cm red square was positioned at eye level in front of the participant. To guarantee that the participant had normal binocular vision and could comfortably focus on the red dot on the wall, they were permitted to wear spectacles if needed. The body sway was evaluated in different conditions, double-leg static and double-leg dynamic, both with closed eyes (CE) and with open eyes (OE). The participants maintained a fixed gaze on the red square during the OE condition. To guarantee that each participant maintained a consistent position throughout the trials, the force plate was marked with the appropriate toe and heel positions. The forefeet were slightly extra-rotated (about 30°), and the intermalleolar distance (distance between the medial malleoli) was approximately 1.5 cm. For each trial, the body sway was recorded for 30 s while the participants stood as still as possible with their hands held relaxed laterally along their hips. The dynamic conditions are represented by an external stimulus generated by a pendulum system, individualized on the height of the subjects (Figure 3.3A). The pendulum, placed behind the subject, impacted on the dorsal part corresponding to the area between the inferior angles of the two scapulae, generating a slight trunk perturbation. The stimulus was produced three times every 7 s (at 7, 14, and 21 s) during the 30 s of the dynamic trials (Di Giminiani et al., 2024; La Greca et al., 2025a). The mass of the pendulum was about 3.5% of the average mass of the participant. Data from the force plate was collected with a sampling frequency of 100 Hz. Body sway was measured by quantifying the displacement of the CoP (figure 3.3B). Previous study suggested that the total CoP pathway has good reliability (ICC ranged

from 0.79 to 0.91), recommending its use as an assessment tool in individuals with ACL-R (Kouvelioti et al., 2015).

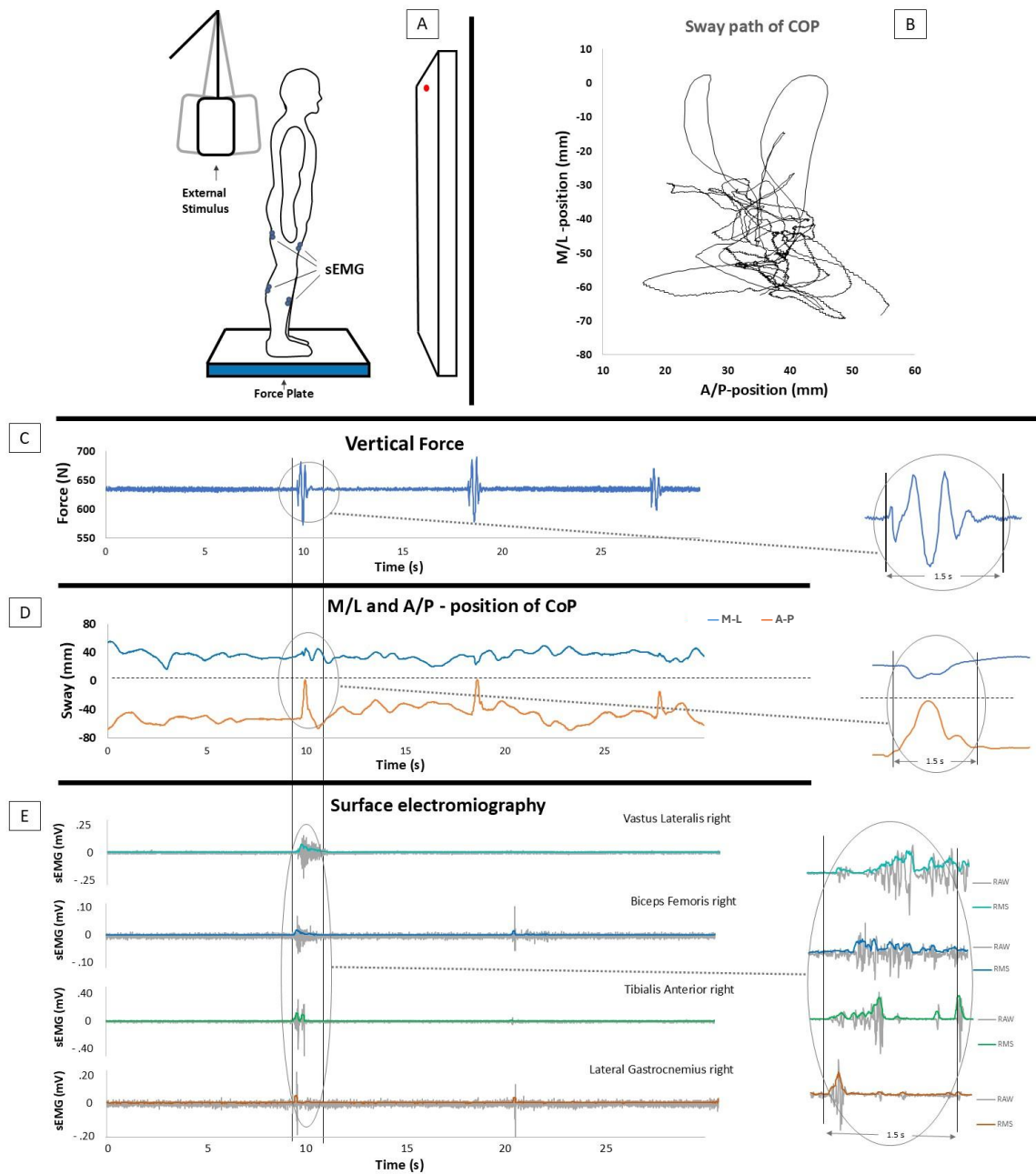


Figure 3.3. Experimental setup for assessment of body sway in dynamic upright standing conditions. A) double-leg dynamic condition; B) body sway path of CoP; C) vertical ground reaction force recording during body sway in dynamic conditions; D) CoP displacement of the anterior–posterior (A-P) and the medio–lateral (M–L) directions, during the dynamic postural assessment; E: sEMG activity during dynamic postural assessment. C–E) examples of the windows to analyses the signals during dynamic conditions (From: La Greca et al., 2025a).

For the offline analysis, the CoP displacements parameter was considered: total length sway path and the M-L and anterior–posterior A-P components of CoP were considered. The total path length was calculated using the following equation:

$$Sway\ path\ length = \sum_{n=1}^{N=1} \sqrt{(FB_{n+1} - FB_n)^2 + (RL_{n+1} - RL_n)^2}$$

where N is the total number of data points in the CoP displacement time series and n is a data point of interest (Rhea et al., 2014).

A nominal resolution of 0.1– 0.2 mm was evaluated in the medio-lateral and anterior-posterior CoP directions (Masedu et al., 2013). The path displacements of the CoP (in millimeters) were analysed throughout the entire 30-second trial in static conditions. Three 1.5-second windows were analysed in dynamic conditions (Figure 3.3C, D, E). Each window was opened when the ground reaction force changed in response to the external stimulus (Figure 3.3D). The body sway was synchronized with the surface electromyography (sEMG) of leg muscles.

Synchronised sEMG activity with the Body Sway

The sEMG activity was detected in the following muscles of both the legs: vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and lateralis gastrocnemius (LG). sEMG was recorded using bipolar electrodes (Ambu Neuroline 720 00-S/25, electrode diameter 45×22 , inter-electrode distance 2.5 cm, Ambu A/S, Baltorpbakken, Denmark). The electrodes were placed on both legs, according to the international surface EMG for non-invasive assessment of muscles (SENIAM) recommendations (Hermens et al., 2000). To minimize the impedance ($<5\text{ k}\Omega$) before the electrodes' application, the skin was shaved, gently abraded with sandpaper (P320) without producing redness, and cleaned with alcohol. The exact location on the muscle belly was marked to ensure correct sensor positioning across the inter-day EMG recordings. In addition, the electrodes and the modules were fixed with elastic strips (Flexa Elast, Pic Solution, Pikdare S.p.A.) to prevent motion artifacts. The wireless data synchronization unit (MuscleLab 6000, Ergotest-innovation, Porsgrunn, Norway) characteristics were as follows: a built-in radio frequency (RF) module ML6RFM02 with 2.4 GHz and 1 mW and a typical wireless range in an open space of 20 m. The modules of the MuscleLab characteristics were as follows: RF characteristics of 2.4 GHz, 1 mW; sample rate of 1.2 kHz; high-pass filter of 20 Hz; bandwidth of 20–500 Hz; input signal range of $\pm 5\text{ mVp-p}$; noise of $14\text{ }\mu\text{Vp-p}$ ($2.2\text{ }\mu\text{V}_{\text{RMS}}$); and resolution of $0.33\text{ }\mu\text{V/bit}$. Utilizing a preamplifier that was situated near the

pickup point and directly connected to the electrodes, the raw sEMG signal was amplified and filtered at 1.2 kHz. The frequencies of 20 Hz and 500 Hz were used to band-pass filter the signals using a fourth-order Butterworth filter. The preamplifier's common mode rejection ratio was 100 dB. In accordance with the standards for reporting sEMG data (International Society of Electrophysiology and Kinesiology, <https://isek.org/resources/>, accessed on 24 July 2025), the EMG signal was converted to a root-mean-square (RMS) signal using a hardware circuit network (frequency response 20–500 kHz, integrating moving average filter with 100 ms width, total error $\pm 0.5\%$). The signal was analysed using the same windows that were used for body sway in the static condition (entire 30 s duration) and the dynamic condition (three windows commenced the perturbation) (Figure 3.3 E).

Statistical analysis

In each of the following studies the XLSTAT 2013.2.07 software (Addinsoft; New York, NY, USA) and/or the Statistics Toolbox implemented in the MATLAB software interface (version 204 R2024b; MathWorks, Natick, MA, USA), were used for statistical analysis. Shapiro-Wilks's W test was used to assess the normality distribution of the variables. Non-parametric tests were implemented due to the absence of a standard Gaussian distribution. To determine the differences between the groups for the anthropometric measured at baseline the Mann-Whitney U test was used.

The dependent variables were compared with the Friedman test for each group during repeated measures over time. A Wilcoxon test for within-group comparisons and by a Mann–Whitney U test for between-group comparisons to determine differences. The Bonferroni correction was used to adjust the p value in relation to the number of contrasts carried out. Statistical significance was set at $\alpha = 0.05$, and the effect size was determined using Hedges' g , which was considered to be small at $g < 0.5$, moderate at $0.5 < g < 0.8$, and large at $g > 0.8$ or was estimated by dividing the Z-statistic by the square root of the sample size ($r = z/\sqrt{N}$). The ES values were small at $r < 0.5$ (Cohen, 2013a).

In each study no significant anthropometric variables differences were found (age, height, body mass, BMI and physical activity level), confirming group that the groups were well-matched.

In the cases report to evaluate the long-term effects of the intervention for each variable the following formula was used:

$$\Delta\% = \left[\frac{\text{follow up} - T0}{T0} \right] * 100$$

In addition, in case of multiple follow-ups, a linear regression model was constructed across the follow-ups, this model examined the trend of the dependent variables over the time. The significance level (α) was set to 0.05.

Study 1.

Locomotion and Postural Control in Young Adults with Autism Spectrum Disorders: A Novel Kinesiological Assessment. (Di Giminiani et al., 2024)

Participants

The study included twenty young adult males: 10 with typical development (TD; age: 23.5 ± 2.5 years; height: 1.79 ± 0.09 m; body mass: 70.1 ± 10.9 kg; body mass index: 22.0 ± 2.6 kg·m⁻²; leg preference: right) and 10 with autism spectrum disorder (ASD; age: 22.4 ± 3.4 years; height: 1.75 ± 0.05 m; body mass: 78.4 ± 10.6 kg; body mass index: 25.2 ± 3.9 kg·m⁻²; leg preference: 9 right, 1 left).

The sample size estimation for the primary outcomes (angle–angle diagram and sEMG synchronised during walking) was computed a priori by means of statistical software for power analysis (G*Power 3.1.9.4, Heinrich Heine-Dusseldorf University, Düsseldorf, Germany). The computation was performed in relation to the study design (*t*-test family for parametric and nonparametric distribution), setting the effect size (ES), and using the protocol for a power analysis (test attributes, large ES [1.8], $\alpha = 0.05$, power $[1 - \beta] = 0.96$, total sample size $n = 20$ participants). The diagnosis of ASD was as follows: 8 with Level 1 and 2 with Level 2 (mean QI = 81.6, standard deviation = 18.8). The inclusion criteria were the absence of any history of injury, fracture, or morphological changes in the lower extremities. Individuals with ASD were recruited from the Reference Regional Centre for Autism (CRRA) in L'Aquila (Italy). They had been diagnosed according to the criteria of the Diagnostic and Statistical Manual of Mental Disorders (American Psychiatric Association, 2013) and the Autism Diagnostic Observation Schedule-2 (ADOS-2) (Lord et al., 2012). Each participant had ADOS-2 scores that were above the cut-off (mean communication and social interaction = 12.2 ± 2.8 ; mean stereotyped behaviours and restricted interests = 0.8 ± 0.8). The experimental procedure was conducted according to the ethical standards of the Declaration of Helsinki. The informed consent was obtained from all the participants and/or their legal guardians. The participants became familiar with the experimental procedure before the measurements (a different day). The measurements lasted about 2 h for each participant, and the protocol was carried out between 10:00 and 13:00 h. At the beginning of the assessment, each participant provided personal, anthropometric, and biometric data. Then, each participant underwent the gait and postural control assessment with synchronised sEMG.

Participants performed static and dynamic balance tasks on a force platform under open-eyes and closed-eyes conditions. Outcome measures included body sway components derived from the center of pressure (CoP) displacement and synchronized surface electromyography (sEMG) activity of the lower limb muscles.

Results

During static balance tasks, no significant differences were observed between the ASD and TD groups in either OE or CE conditions when analysing the medio-lateral and anterior–posterior displacements ($p > 0.05$; Table 3.1). CoP displacement parameters were comparable between groups, indicating similar postural stability under non-perturbed conditions.

In contrast, during dynamic conditions, significant changes in sway path were observed within each group. The ASD group showed a significant decrease in sway path during the OE condition ($p < 0.05$; ES = 0.96 and ES = 1.14), whereas the TD group exhibited a significant decrease in sway path during the CE condition ($p < 0.05$; ES = 1.19 and ES = 1.07) (Table 3.2).

Table 3.1. Body sway values measured during static upright in ASD and TD. Median values (interquartile range) are reported. (From: Di Giminiani et al., 2024).

Variables	ASD	TD	p-Values
DLS-CE-ML	393.1(321.0-447.6)	472.3(410.8-533.5)	0.063
DLS-CE-AP	554.7(482.2-605.4)	540.0(405.9-597.9)	0.796
DLS-OE-ML	294.0(247.7-518.5)	421.3(366.0-545.4)	0.089
DLS-OE-AP	419.7(372.5-491.9)	434.9(356.0-538.9)	0.869

Abbreviations: Double-leg static (DLS); Open Eyes (OE); Closed Eyes (CE); Medio-Lateral (ML); Anterior-Posterior (AP).

Table 3.2. Body sway values measured during dynamic upright standing in ASD and TD. Median values (interquartile range) are reported. (From: Di Giminiani et al., 2024).

Variable	1 st	2 nd	3 rd	Overall p-values
ASD				
DLD-OE-ML	35.9(29.6-43.9)	26.7(21.8-34.7)	28.2(25.9-30.5) §	0.020
DLD-CE-ML	46.4(31.4-53.7)	32.5(25.0-39.0) §§	31.3(27.6-48.6)	0.006
DLD-OE-AP	110.7(86.0-133.4)	100.2(74.5-137.8)	66.6(53.0-89.9) §§§	0.027
DLD-CE-AP	123.9(94.4-142.8)	97.5(66.6-128.9)	99.6(65.7-138.5)	0.497
TD				
DLD-OE-ML	27.9(19.7-45.2)	29.7(18.8-33.7)	26.1(15.6-32.5)	0.301
DLD-CE-ML	45.9(41.6-53.2)	36.8(21.3-46.2)	36.6(18.8-41.3) *	0.020
DLD-OE-AP	62.2(44.9-104.2)	62.8(45.3-81.2)	56.9(46.2-68.2)	0.407
DLD-CE-AP	152.2(92.7-197.7)	93.6(56.8-176.8)	72.3(39.0-131.5) **	0.045

Abbreviations: Double-leg dynamic (DLD); Open Eyes (OE); Closed Eyes (CE); Medio-Lateral (ML); Anterior-Posterior (A-P); 1st External Perturbation; 2nd External Perturbation; 3rd External

Perturbation. §Significant contrast 1st vs. 3rd (P=0.007); §§ Significant contrast 1st vs. 2nd (P=0.002); §§§Significant contrast 1st vs. 3rd (P=0.007); *Significant contrast 1st vs. 3rd (P=0.007); **Significant contrast 1st vs. 3rd (P=0.014).

sEMG activity recorded during static tasks did not differ between ASD and TD participants (Table 3.3). During dynamic tasks, distinct muscle activation patterns emerged between groups. In individuals with TD, the tibialis anterior showed significant responses during both OE and CE conditions ($p < 0.05$; ES = 0.70 and ES = 0.68, respectively). In individuals with ASD, the lateral gastrocnemius was significantly involved during the OE condition ($p < 0.05$; ES = 0.46), whereas the tibialis anterior showed significant activation during the CE condition ($p < 0.05$; ES = 0.21) (Table 3.4). Overall, dynamic conditions were able to detect alterations in postural control and neuromuscular organization in individuals with ASD that were not observable during static assessments, with CoP sway changes corresponding to group-specific muscle activation patterns.

Table 3.3. Surface electromyographic (sEMG) activity of leg muscles synchronized with body sway during static upright standing. Median values (interquartile range) are reported. (From: Di Giminiani et al., 2024)

Variables	ASD	TD	p-Values
Static-CE			
VL	2.8(2.6-6.9)	3.4(2.8-13.2)	0.481
BF	5.8(3.1-12.0)	12.5(7.4-15.9)	0.105
TA	3.6(3.4-9.4)	3.7(3.3-4.6)	0.967
LG	5.2(3.7-6.3)	9.4(6.8-15.0)	0.029*
Static-OE			
VL	2.9(2.6-4.0)	2.9(2.7-13.6)	0.481
BF	4.6(2.7-15.1)	10.9(4.7-19.6)	0.160
TA	3.6(3.4-12.2)	5.0(4.1-10.5)	0.393
LG	4.0(3.1-5.2)	7.5(5.0-14.4)	0.011*

Abbreviations: Open Eyes (OE); Closed Eyes (CE); Vastus Lateralis (VL); Biceps Femoris (BF); Tibialis Anterior (TA); Lateralis Gastrocnemius (LG); *Significant contrast ASD vs. TD ($P < 0.05$)

Table 3.4. Surface electromyographic (sEMG) activity of leg muscles synchronized with body sway during dynamic upright standing. Median values and (interquartile range) are reported. (From: Di Giminiani et al., 2024)

Dynamic-OE	ASD			Overall p-values
	1 st	2 nd	3 rd	
VL	7.0(2.7-18.3)	3.1(2.8-5.7)	2.8(2.7-7.1)	0.614
BF	8.5(7.0-12.6)	9.5(7.2-11.2)	7.4(5.5-9.6)	0.407
TA	7.6(3.5-38.3)	5.6(3.8-14.3)	4.1(3.5-18.7)	0.717
LG	14.4(6.2-44.8)	10.2(5.2-13.1)	8.1(4.4-17.2) §	0.008*

Dynamic-CE				
VL	7.5(2.8-28.8)	3.0(2.8-14.8)	2.9(2.5-10.6)	0.497
BF	14.8(5.0-16.1)	13.2(4.0-22.0)	14.8(4.3-17.5)	0.905
TA	7.6(3.9-56.2)	4.3(3.4-6.3)	3.8(3.6-6.6) *	0.027*
LG	28.3(15.2-41.0)	14.8(10.1-32.3)	15.8(7.2-47.3)	0.407
TD				
Dynamic-OE	1 st	2 nd	3 rd	
VL	3.8(3.0-8.2)	3.4(3.0-5.4)	2.9(2.8-6.9)	0.497
BF	17.6(10.3-28.5)	14.2(8.9-24.9)	17.7(7.6-28.4)	0.584
TA	4.8(3.4-13.8)	3.7(3.1-8.0) ##	4.2(3.2-5.7) #	0.007*
LG	14.2(8.3-34.4)	7.9(5.2-17.5)	14.2(8.1-25.4)	0.061
Dynamic -CE				
VL	10.6(5.4-22.4)	9.4(5.3-28.9)	4.8(3.3-36.7)	0.202
BF	20.8(17.2-46.7)	16.6(13.4-50.5)	19.1(9.5-40.8)	0.082
TA	5.5(3.1-19.4)	4.5(3.0-19.8)	4.0(2.9-11.1) +	0.006*
LG	24.6(15.9-42.5)	26.0(18.3-31.9)	20.3(7.9-34.0)	1.000

Abbreviations: Open Eyes (OE); Closed Eyes (CE); Vastus Lateralis (VL); Biceps Femoris (BF); Tibialis Anterior (TA); Lateralis Gastrocnemius (LG); 1st External Perturbation; 2nd External Perturbation; 3rd External Perturbation. §Significant contrast 1st vs. 3rd (P=0.003); ##Significant contrast 1st vs. 2nd (P=0.004); #Significant contrast 1st vs. 3rd (P=0.014); *Significant contrast 1st vs. 3rd (P=0.007); +Significant contrast 1st vs. 3rd (P=0.002).

Study 2.

An evaluation of postural balance in single-leg and double-leg conditions in patients with reconstruction of the anterior cruciate ligament and healthy controls. (La Greca et al., 2025a)

Participant

This study included athletes who had undergone anterior cruciate ligament reconstruction (ACL-R) with completed rehabilitation and a group of healthy control athletes matched for sex, anthropometric characteristics, and physical activity level. Participants performed static and dynamic balance tasks on a force platform. Anthropometric data are reported in Table 3.5. The following were the requirements for inclusion in this group: full functional recovery with a minimum of 6-month lag and full return to athletic competition. The rehabilitation program was managed in accordance with the recommendations of Beynnon et al. (2005) and was conducted at a national health system (NHS) center that is affiliated with our university. For the control group, the inclusion criteria were as follows: no history of musculoskeletal, ligament, or meniscus injuries to the lower extremities. The exclusion criteria for the two groups were as follows: actual or a history of skeletal muscular or nervous injuries, neuromuscular system pathologies, herniated, arrhythmias, epilepsies, and comorbidities with other disturbances. The experimental procedure was conducted according to the ethical standards of the Declaration of Helsinki.

Table 3.5. Participants characteristics. (From: La Greca et al 2025a)

Variables	Group		P	ES
	ACL-R Group (n = 13)	Health Control Group (n = 14)		
Age (years)	24.5±2.7	25.4±3.4	0.497	0.029
Stature (cm)	178.5±.09	177.4±.06	0.884	0.019
Body Mass (Kg)	77.0±10.4	77.9±14.7	0.990	0.070
BMI (Kg/m ²)	24.1±2,2	24.8±4.1	0.607	0.210
Sex (M/F)	10/3	11/3	0.928	0.040
Sport level	Competitive	Competitive	0.933	0.134
Leg Dominance	Right (n=11) / Left (n=2)	Right (n=12) / Left (n=2)	0.888	0.030
Operated Leg	Right (n=7) / Left (n=6)	N/A	N/A	N/A
Graft type	SGT (n=13)	N/A	N/A	N/A
Event Distribution	No-Contact Mechanism	N/A	N/A	N/A
Post-Operative Period	6 Months-2 Years	N/A	N/A	N/A

Abbreviations: body mass index (BMI), semitendinosus/gracilis tendon (SGT).

Results

During static balance tasks, significant differences were observed in the anterior–posterior (A–P) CoP sway path between groups, with higher values in the ACL-R group under both OE ($p = 0.032$; ES = 1.20; Figure 3.4A) and CE conditions ($p = 0.032$; ES = 1.17; Figure 3.4B) compared with healthy controls.

During dynamic conditions, athletes with ACL-R exhibited significantly greater CoP displacement in the medio-lateral (M–L) sway path under OE conditions compared with controls ($p = 0.012$; ES = 1.39; Figure 3.4C).

No significant differences between-groups were observed in sEMG activity of the analysed muscles during static conditions (Table 3.6). In contrast, during dynamic tasks, sEMG activity of the tibialis anterior was significantly higher in the ACL-R group than in the healthy control group under OE conditions in both legs ($p = 0.040$; ES = 1.62 and $p = 0.028$; ES = 1.82 Figure 3.5, Table 3.7).

Additionally, the biceps femoris muscle in the ACL-R group showed higher EMG activity during OL ($p = 0.040$; ES = 1.05) and N-OL ($p = 0.028$; ES = 0.83) conditions compared with healthy controls (Figure 3.5).

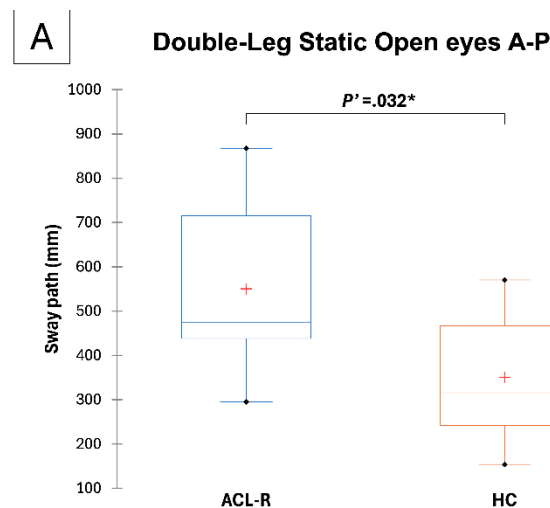


Figure 3.4. cont.

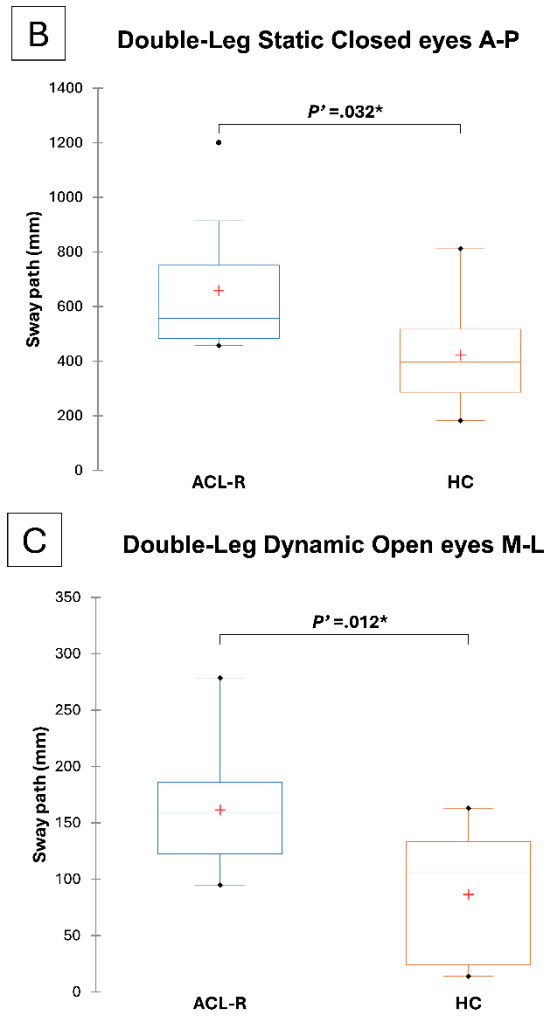


Figure 3.4. Displacement of medio-lateral (M-L) and anterior-posterior (A-P) directions of the centre of pressure (CoP) during double-leg condition. Differences between anterior cruciate ligament reconstruction group (ACL-R) and healthy control group (HC). $^* = P < 0.05$. (From: La Greca et al., 2025a)

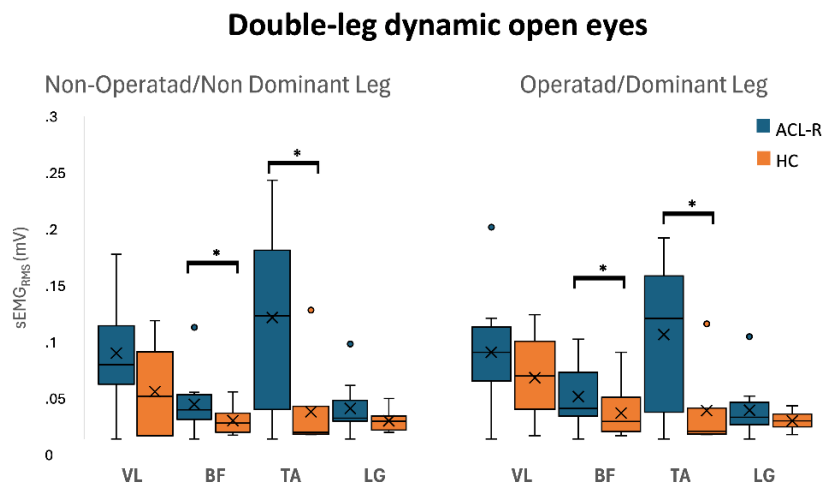
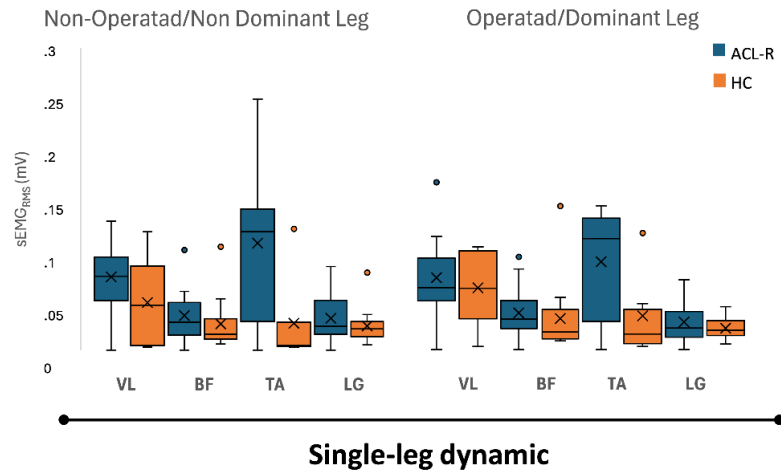


Figure 3.5. cont.

Double-leg dynamic closed eyes



Single-leg dynamic

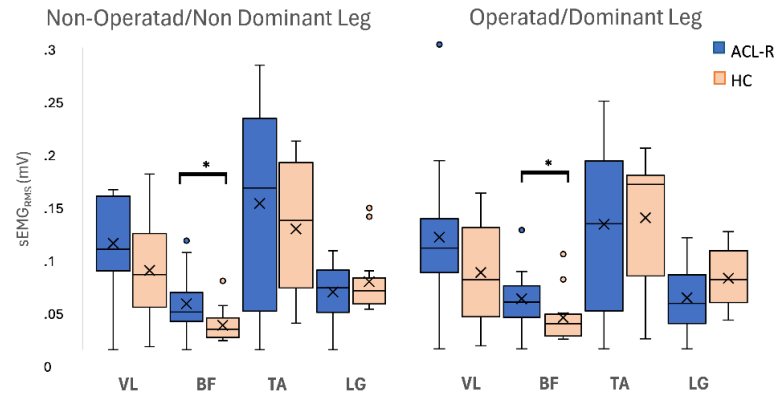


Figure 3.5. Surface electromyography root-mean-square (sEMG_{RMS}) activity of leg muscles synchronized with dynamic body sway assessment. Differences between anterior cruciate ligaments reconstruction group (ACL-R) and healthy control group (HC). VL= vastus lateralis, BF= biceps femoris, TA= tibialis anterior, LG= lateral gastrocnemius. * = $P < 0.05$. (From: La Greca et al., 2025a).

Table 3.6. sEMG_{RMS} activity values of ACL-R group and healthy control group during static body sway assessment. (From: La Greca et al., 2025a)

Variables	Operated leg/dominant leg			Non Operated leg/Non dominant leg		
Double-leg static CE	ACL-R	HC	P	ACL-R	HC	P
VL (mV)	0.019 ± 0.018	0.015 ± .016	0.766	0.018 ± 0.010	0.011 ± 0.012	0.249
BF (mV)	0.023 ± 0.020	0.013 ± .015	0.680	0.026 ± 0.027	0.009 ± 0.008	0.240
TA (mV)	0.016 ± 0.016	0.009 ± .008	0.766	0.020 ± 0.022	0.005 ± 0.002	0.249
LG (mV)	0.014 ± 0.006	0.012 ± .008	0.699	0.021 ± 0.016	0.017 ± 0.016	0.342
Double-leg static OE	ACL-R	HC	P	ACL-R	HC	P
VL (mV)	0.018 ± 0.019	0.016 ± .016	1.000	0.017 ± 0.012	0.008 ± 0.008	0.246
BF (mV)	0.022 ± 0.020	0.017 ± .019	1.000	0.017 ± 0.013	0.011 ± 0.008	0.470
TA (mV)	0.025 ± 0.028	0.010 ± .010	1.000	0.028 ± 0.028	0.006 ± 0.003	0.232
LG (mV)	0.014 ± 0.007	0.012 ± .008	1.000	0.021 ± 0.016	0.012 ± 0.009	0.246
Single-leg static OL/DL	ACL-R	HC	P			
VL (mV)	0.029 ± 0.026	0.030 ± 0.028	1.000			
BF (mV)	0.030 ± 0.020	0.012 ± 0.006	0.060			
TA (mV)	0.066 ± 0.062	0.070 ± 0.045	1.000			
LG (mV)	0.030 ± 0.015	0.043 ± 0.024	0.450			

Single-leg static N-OL/N-DL	ACL-R	HC	<i>P</i>
VL (mV)	0.024 ± 0.013	0.025 ± 0.019	1.000
BF (mV)	0.026 ± 0.032	0.014 ± 0.009	0.772
TA (mV)	0.064 ± 0.040	0.081 ± 0.045	1.000
LG (mV)	0.038 ± 0.027	0.048 ± 0.031	1.000

Abbreviation: anterior cruciate ligament reconstruction group (ACL-R) health control group (HC); open eyes (OE); closed eyes (CE); operated leg (OL); non-operated leg (N-OL); dominant-leg (DL); non-dominant leg (N-DL); vastus lateralis (VL); biceps femoris (BF); vastus medialis (VM); tibialis anterior (TA); lateralis gastrocnemius (LG); *= $P < .05$.

Table 3.7. sEMG_{RMG} activity values of ACL-R group and healthy control group during dynamic body sway assessment. (From: La Greca et al., 2025a)

Variables	Operated leg/dominant leg			Non operated leg/Non dominant leg		
Double-leg dynamic CE	ACL-R	HC	<i>P</i>	ACL-R	HC	<i>P</i>
VL (mV)	0.075 ± 0.033	0.059 ± 0.032	0.696	0.078 ± 0.026	0.047 ± 0.042	0.339
BF (mV)	0.039 ± 0.025	0.030 ± 0.035	0.696	0.037 ± 0.026	0.026 ± 0.026	0.528
TA (mV)	0.097 ± 0.044	0.032 ± 0.042	0.088	0.122 ± 0.072	0.027 ± 0.043	0.104
LG (mV)	0.029 ± 0.017	0.021 ± 0.010	0.696	0.034 ± 0.021	0.024 ± 0.017	0.528
Double-leg dynamic OE	ACL-R	HC	<i>P</i>	ACL-R	HC	<i>P</i>
VL (mV)	0.084 ± 0.039	0.055 ± 0.035	0.132	0.086 ± 0.041	0.044 ± 0.040	0.052
BF (mV)	0.042 ± 0.024	0.021 ± 0.022	0.040*	0.035 ± 0.025	0.017 ± 0.013	0.028*
TA (mV)	0.108 ± 0.058	0.026 ± 0.036	0.040*	0.129 ± 0.074	0.025 ± 0.043	0.028*
LG (mV)	0.028 ± 0.021	0.030 ± 0.008	0.132	0.030 ± 0.021	0.017 ± 0.009	0.054
Single-leg dynamic OL/DL	ACL-R	HC	<i>P</i>			
VL (mV)	0.116 ± 0.063		0.074 ± 0.048			0.315
BF (mV)	0.053 ± 0.025		0.030 ± 0.024			0.024*
TA (mV)	0.138 ± 0.072		0.127 ± 0.065			0.816
LG (mV)	0.053 ± 0.030		0.068 ± 0.026			0.315
Single-leg dynamic N-OL/N-DL	ACL-R	HC	<i>P</i>			
VL (mV)	0.114 ± 0.034		0.081 ± 0.048			0.198
BF (mV)	0.050 ± 0.028		0.024 ± 0.017			0.012*
TA (mV)	0.166 ± 0.088		0.120 ± 0.067			0.766
LG (mV)	0.062 ± 0.023		0.068 ± 0.031			0.768

Abbreviation: anterior cruciate ligament reconstruction group (ACL-R) health control group (HC); open eyes (OE); closed eyes (CE); operated leg (OL); non-operated leg (N-OL); dominant-leg (DL); non-dominant leg (N-DL); vastus lateralis (VL); biceps femoris (BF); vastus medialis (VM); tibialis anterior (TA); lateralis gastrocnemius (LG); *= $P < 0.05$.

Despite the different populations examined, a common pattern emerged across Studies 1 and 2. Static balance conditions showed limited discriminatory capacity, whereas dynamic conditions consistently revealed alterations in postural control and neuromuscular organization. The combined analysis of CoP displacement and synchronized sEMG proved effective in detecting these differences.

This assessment approach was further applied to evaluate the chronic effects of exercise-based interventions in two case studies, showing greater sensitivity in identifying postural control adaptations compared with static balance assessments.

Study 3

A focus on exercise prescription and assessment for a safe return to sport participation following a patellar tendon reconstruction in a soccer player (Marinelli et al., 2024)

Participant

This case report evaluated the effectiveness of a six-month kinesiological return-to-sport program following patellar tendon reconstruction in regional football players.

Results

Static balance assessments under OE conditions showed a reduction in sway path, with a 48.68% decrease in the ML component and a 57.46% decrease in the AP component. Under CE conditions, ML and AP sway paths decreased by 37.08% and 35.85%, respectively, indicating an improvement in static postural performance over time. However, during dynamic conditions, the AP sway path increased under both OE and CE conditions by 11.88% and 79.06%, respectively (Figure 3.6), indicating incomplete recovery of dynamic postural control.

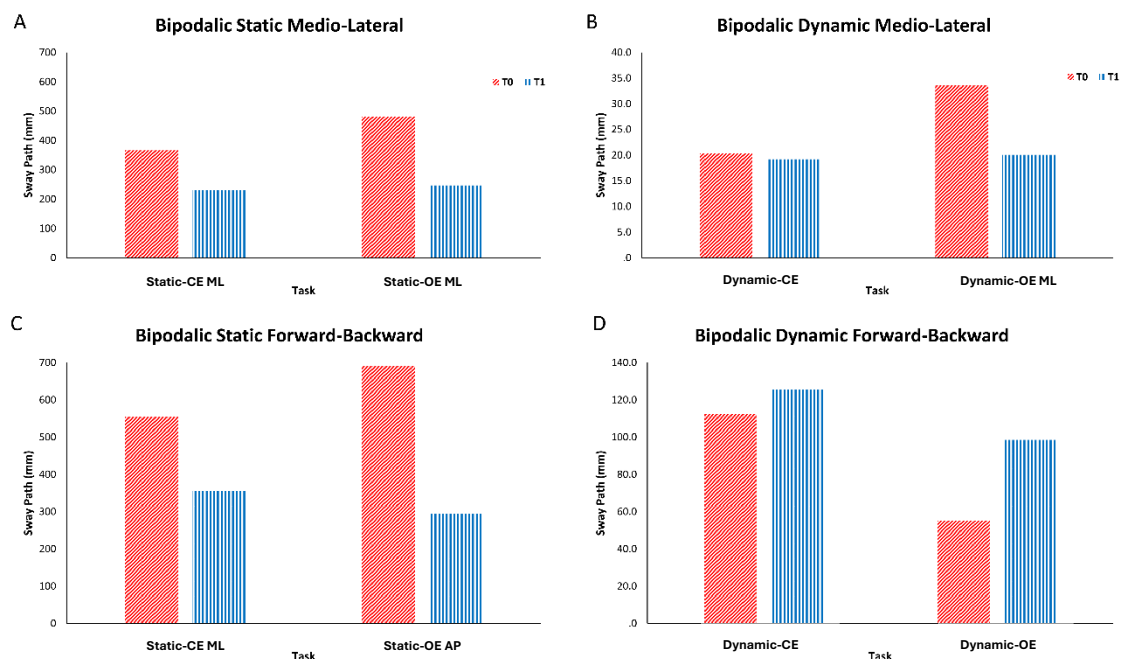


Figure 3.6. Body sway values in different conditions: A) static in medio-lateral (M-L) B) and dynamic M-L; C) static anterior-posterior (A-P) and D) dynamic A-P; with closed (CE) and open (OE) eyes. Differences between T0 and T1 were reported. (From: Marinelli et al., 2024)

Study 4

Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. (La Greca et al., 2025b)

Participant

This case study investigated the long-term (5-year) effects of individualized whole-body vibration (WBV) training on static and dynamic postural control in person with MS. The intervention was assessed during follow-ups (T0, T1, T2, T3 and T4).

Results

Body sway measurements showed significant improvements over time during dynamic conditions under both OE and CE ($p < 0.05$), whereas changes in static conditions were limited. Specifically, a significant improvement was observed only in static OE condition ($p < 0.05$). No significant changes were found in the static CE condition for total path, ML, or AP displacement ($p > 0.05$) (Table 3.8).

The largest reductions in body sway were observed during dynamic conditions, with an average decrease of approximately 60%.

Across studies, dynamic balance conditions consistently revealed postural and neuromuscular alterations that were absent or less evident during static assessments.

Table 3.8. Body sway is measured on the different test times in different conditions. The differences ($\Delta\%$) and the different follow-ups are also reported. (From: La Greca et al., 2025b)

Body sway	T0	T1	T2	T3	T4	P	T0-T1 $\Delta\%$	T0-T2 $\Delta\%$	T0-T3 $\Delta\%$	T0-T4 $\Delta\%$
S-CE Total path (mm)	795.9	755.6	859	526.9	458	0.094	-5	8	-34	-42
S-CE M-L path (mm)	433.1	506.8	539.8	354.7	225	0.179	17	25	-18	-48
S-CE A-P path (mm)	509.6	612	568.5	334.5	346	0.145	20	12	-34	-32
S-OE Total path (mm)	781.5	723.2	644.9	489.2	384	0.002*	-7	-17	-37	-51
S-OE M-L path (mm)	453.2	454.7	355.9	288.6	235	0.005*	0	-21	-36	-48
S-OE A-P path (mm)	517.3	598	466.4	338.9	263	0.038*	16	-10	-34	-49
D-CE Total path (mm)	1338.4	1191.2	834.7	819.1	527	0.004*	-11	-38	-39	-61
D-CE M-L path (mm)	57.7	40.9	31.2	38.2	21.2	0.043*	-29	-46	-34	-63
D-CE A-P path (mm)	122.6	52.7	59.3	58.9	52.8	0.185	-57	-52	-52	-57
D-OE Total path (mm)	1019.7	1002.4	810.6	785	477	0.017*	-2	-21	-23	-53
D-OE M-L path (mm)	44.1	30.8	34.2	33.9	25.2	0.107	-30	-22	-23	-43
D-OE A-P path (mm)	71.2	59.3	53.1	82.8	38	0.509	-17	-25	16	-47

Abbreviations: static (S); dynamic (D); closed eyes (CE); open eyes (OE); medio-lateral (M-L); anterior-posterior (A-P).

3.4 Balance under single- and double-leg conditions

In addition to static and dynamic balance conditions, postural control is frequently assessed using single-leg and double-leg stance tasks. These assessment modalities are particularly common in athletic populations, where balance performance is considered a key component of functional readiness and return-to-sport decision-making. Single-leg stance tasks are often proposed because they increase task difficulty by reducing the base of support and are thought to better challenge neuromuscular and sensorimotor control mechanisms.

However, single-leg balance tasks may not be feasible for all individuals, as they represent a complex motor task requiring adequate strength, coordination, and postural control. Moreover, single-leg conditions may be difficult to standardize across subjects and populations, potentially limiting their reliability and sensitivity in detecting subtle neuromuscular alterations.

Despite these limitations, single-leg stance assessments are widely adopted in sports medicine and rehabilitation contexts. In particular, current test batteries used to evaluate readiness to return to sport following ACL-R predominantly include single-leg motor tasks, typically comparing the operated limb with the non-operated limb (Roe et al., 2022). Consequently, most studies investigating postural control after ACL-R have focused on balance performance during single-leg standing (Hoffman et al., 1999; Lehmann et al., 2017; Mayer et al., 2015; Paterno et al., 2013; Zult et al., 2017).

Nevertheless, it remains unclear whether single-leg or double-leg conditions are more reliable for identifying persistent neuromuscular alterations in postural balance in individuals with ACL-R compared with healthy athletes (Howells et al., 2011). The heterogeneity of assessment protocols and outcome measures employed across studies may contribute to the variability in reported findings.

Recent evidence suggests that bilateral motor tasks may represent a reliable approach for detecting residual neuromuscular and postural control alterations following ACL-R (Cuthbert et al., 2021; Di Giminiani et al., 2023; Przysucha et al., 2024). In this context, Bodkin et al. (2018) reported that balance assessment during single-leg stance may not be sufficiently sensitive to identify persistent impairments in individuals with ACL-R upon return to sport. Conversely, balance in double-leg conditions has been investigated in a limited number of studies (Henriksson et al., 2001; Howells et al., 2011; Sahoo and Sahu, 2021), despite growing evidence supporting its sensitivity.

Double-leg balance assessments may provide greater sensitivity in identifying persistent

changes in neuromuscular and postural control than single-leg assessments (Bodkin et al., 2018). Bilateral motor tasks require greater neural network connectivity than unilateral tasks (Noble et al., 2014), consistent with the bilateral organization of lower-limb motor control (Luft et al., 2002). Following ACL-R, individuals often exhibit reduced motor output and increased bilateral corticospinal excitability, suggesting a reorganization of central motor pathways (Lepley et al., 2019). These adaptations indicate that the non-injured limb may not represent a valid internal control for functional comparisons (Grooms et al., 2017), as neuromuscular alterations appear to involve both limbs (Konishi et al., 2007; Zult et al., 2017).

Based on this rationale, the present study investigated postural balance under single-leg and double-leg conditions in athletes with ACL-R and matched healthy controls.

Study

An evaluation of postural balance in single-leg and double-leg conditions in patients with reconstruction of the anterior cruciate ligament and healthy controls. (La Greca et al., 2025a)

Participant

These results refer to the same group of athletes included in the sample of study 2 (La Greca et al., 2025a).

Methods

Setup for postural balance assessment

The experimental setup for postural balance assessment was identical to that described in the previous section. In addition, body sway was evaluated during single-leg static and single-leg dynamic tasks performed under open-eyes conditions. To ensure consistent positioning across trials, the force platform was marked with standardized toe and heel reference points. During double-leg conditions, the forefeet were slightly externally rotated (approximately 30°), and the intermalleolar distance was set at approximately 1.5 cm. During single-leg conditions, participants placed the supporting foot at the center of the platform, while the non-supporting leg was flexed at 90° at the femur–tibia axis, keeping the thigh aligned with the supporting leg (Figure 3.7). No significant within-group differences were found between the dominant (DL) and non-dominant leg (N-DL) in healthy controls, nor between the operated (OL) and non-operated leg (N-OL) in ACL-R participants. As the operated limb coincided with the dominant leg in most ACL-R participants (9 DL, 4 N-DL), comparisons were performed between the DL of healthy controls and the OL of the ACL-R group, and between the N-DL and N-OL.

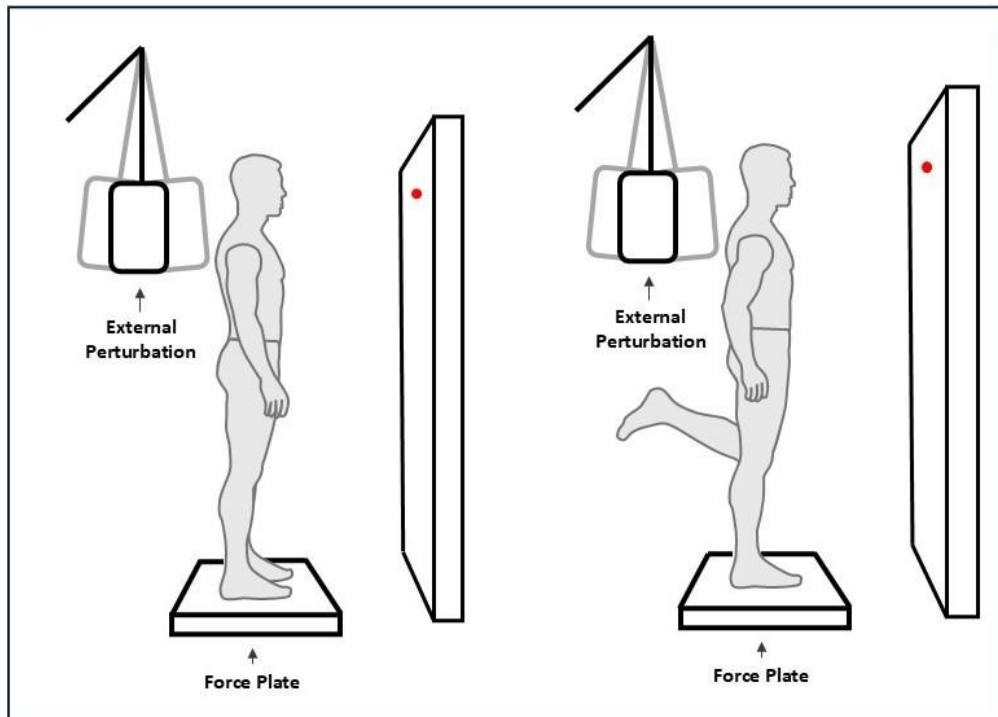


Figure 3.7. Experimental setup for assessment of body sway in dynamic upright standing in single- and double-leg condition (From: La Greca et al., 2026).

Statistical analysis

The XLSTAT 2013.2.07 software (Addinsoft; New York, NY, USA) were used for statistical analysis. Shapiro-Wilks's W test was used to assess the normality distribution of the variables. Non-parametric tests were implemented due to the absence of a standard Gaussian distribution. To determine the differences between the groups for the anthropometric measured at baseline the Mann-Whitney U test was used.

The dependent variables were compared with the Friedman test for each group during repeated measures over time. A Wilcoxon test for within-group comparisons and by a Mann–Whitney U test for between-group comparisons to determine differences. The Bonferroni correction was used to adjust the p value in relation to the number of contrasts carried out. Statistical significance was set at $\alpha = 0.05$, and the effect size was estimated by dividing the Z-statistic by the square root of the sample size ($r = z/\sqrt{N}$). The ES values were small at $r < 0.5$ (Cohen, 2013).

In each study no significant anthropometric variables differences were found (age, height, body mass, BMI and physical activity level), confirming group that the groups were well-matched.

Results

During static and dynamic conditions, single-leg balance tasks showed no differences between athletes with ACL-R and healthy controls ($p > 0.05$). CoP displacement parameters did not consistently differ between the operated and non-operated limbs ($p > 0.05$, Table 3.9).

Table 3.9. ACL-R group sway path values (M-L and A-P) of static and dynamic body sway; differences between operated and non-operated leg. (From: La Greca et al., 2025a)

Static Postural Balance					
Variables	OL	N-OL	P	ES	95% CI (mm)
Single-leg static M-L (mm)	850.16±165.1	867.8±133.2	1.000	0.110	-110.6–145.7
Single-leg static A-P (mm)	1145.5±409.2	1194.7±314.0	1.000	0.130	-262.5–360.0
Dynamic Postural Balance					
			P	ES	95% CI (mm)
Single-leg dynamic M-L (mm)	130.67±36.5	124.21±27.9	1.000	0.190	-34.2–21.3
Single-leg dynamic A-P (mm)	316.63±59.9	308.35±87.9	1.000	0.100	-72.6–56.0

Abbreviation: operated leg (OL); non-operated leg (N-OL); anterior-posterior (A-P); medio-lateral (M-L); Edge's g effect size (ES); confidence interval (95% CI); Adjusted P value with Holm-Bonferroni correction.

In contrast, during double-leg conditions, the ACL-R group showed significantly higher CoP displacement compared with health controls. Specifically, greater sway was observed in the anterior–posterior direction during static conditions ($p = 0.032$; ES = 1.20) and in the medio–lateral direction during dynamic conditions ($p = 0.012$; ES = 1.39) (Table 3.10).

Table 3.10. ACL-R group and health control group, sway path values (M-L and A-P) of static and dynamic body sway. (From: La Greca et al., 2025a)

Variables	ACL-R	HC	P
Static Postural Balance			
Double-leg OE M-L (mm)	308.8 ± 104.4	242.6 ± 56.1	0.204
Double-leg CE M-L (mm)	338.7 ± 114.7	271.7 ± 81.2	0.204
Double-leg OE A-P (mm)	550.3 ± 186.3	350.6 ± 145.9	0.032*
Double-leg CE A-P (mm)	658.5 ± 226.3	422.4 ± 176.2	0.032*
Single-leg operated-leg/dominant-leg M-L (mm)	850.1 ± 165.1	940.4 ± 214.4	0.660
Single-leg non operated-leg/ non dominant-leg M-L (mm)	867.8 ± 133.2	930.2 ± 245.4	1.000
Single-leg operated-leg/dominant-leg A-P (mm)	1145.5 ± 409.2	1082.7 ± 318.9	1.000
Single-leg non operated-leg/ non dominant-leg A-P (mm)	1194.7 ± 314.0	1005.6 ± 335.4	0.500
Dynamic Postural Balance			
Double-leg OE M-L (mm)	161.5 ± 48.6	86.6 ± 58.9	0.012*
Double-leg CE M-L (mm)	162.8 ± 54.7	107.9 ± 70.0	0.117
Double-leg OE A-P (mm)	362.5 ± 74.9	271.1 ± 198.2	0.740
Double-leg CE A-P (mm)	385.6 ± 88.1	269.0 ± 178.4	0.170
Single-leg operated-leg/dominant-leg M-L (mm)	130.7 ± 36.5	108.8 ± 42.4	1.000
Single-leg non operated-leg/ non dominant-leg M-L (mm)	124.2 ± 27.9	117.6 ± 55.5	1.000
Single-leg operated-leg/dominant-leg A-P (mm)	316.6 ± 59.9	302.8 ± 244.3	1.000
Single-leg non operated-leg/ non dominant-leg A-P (mm)	308.4 ± 87.9	261.4 ± 139.1	1.000

Abbreviation: anterior cruciate ligament reconstruction group (ACL-R) health control group (HC); open eyes (OE); closed eyes (CE); medio-lateral (M-L); anterior-posterior (A-P); * = $P < 0.05$.

Regarding neuromuscular activity, during the double-leg dynamic condition with OE, the ACL-R group showed significantly greater activation of the biceps femoris muscle in both the operated limb ($p = 0.040$; $ES = 1.05$) and the non-operated limb ($p = 0.028$; $ES = 0.83$) compared with healthy controls. Similarly, during the single-leg dynamic condition, the ACL-R group exhibited greater biceps femoris activation in both the operated limb ($p = 0.024$; $ES = 1.15$) and the non-operated limb ($p = 0.012$; $ES = 1.03$) compared with healthy controls.

In the double-leg dynamic condition, tibialis anterior sEMG activity was significantly higher in the ACL-R group than in healthy controls under OE conditions ($p = 0.040$; $ES = 1.62$ and $p = 0.028$; $ES = 1.82$ for OL/DL and N-OL/N-DL, respectively). No significant differences between-groups were observed for tibialis anterior activity during single-leg dynamic assessment ($p > 0.05$) (Table 3.7).

Overall, alterations in postural control and neuromuscular activation were less pronounced or absent during single-leg conditions. Dynamic double-leg tasks demonstrated greater sensitivity in identifying residual postural control and neuromuscular alterations in athletes with ACL-R compared with single-leg tasks.

3.5 Synthesis and applied clinical and functional implications

The present findings consistently indicate that postural balance alterations are more effectively identified when the neuromuscular system is challenged through dynamic and functionally demanding tasks, rather than through static assessments. This pattern emerged across different populations, including individuals with autism (ASD), athletes following ACL-R, and longitudinal case studies, highlighting the robustness and translational relevance of the adopted assessment paradigm (Di Giminiani et al., 2024; La Greca et al., 2025a).

Static balance assessments primarily reflect the ability to maintain an upright posture under predictable and minimally challenging conditions (Nashner and McCollum, 1985; Winter, 1995). Although useful for describing basic postural stability, static tasks often show limited sensitivity in distinguishing between clinical and control groups, as reported in both neurological and musculoskeletal populations (Di Giminiani et al., 2024; La Greca et al., 2025a). Nevertheless, due to their easy reproducibility and application, static assessments are widely used in research and, above all, in clinical practice (Bodkin et al., 2018; Howells et al., 2011; Nashner, 1977). As a consequence, reliance on static tasks alone may lead to an underestimation of functional postural deficits, particularly those that emerge when continuous sensory integration, adaptive motor responses, and neuromuscular coordination are required (Winter, 1995).

In individuals with ASD, alterations in postural control became evident under dynamic conditions during repeated imbalance. These findings are consistent with previous evidence (Cham et al., 2021) indicating that balance disturbances in ASD are more pronounced when task complexity and sensory integration demands increase. From a neurorehabilitation perspective, this has relevant clinical implications. Static balance tests may fail to capture the difficulties experienced during daily activities characterized by dynamic and unpredictable environmental demands. Dynamic balance assessments, especially when combined with synchronized CoP and sEMG analyses, may therefore provide a more sensitive and functionally relevant assessment of postural control in ASD, supporting both individualized intervention planning and outcome monitoring.

Similarly, in athletes with ACL-R, dynamic balance tasks revealed persistent postural and neuromuscular alterations despite completed rehabilitation and clearance for return to sport. Single-leg balance tasks, commonly included in clinical assessment batteries (Hoffman et al., 1999; Lehmann et al., 2017; Mayer et al., 2015; Paterno et al., 2013; Roe et al., 2022; Zult et al., 2017), showed limited ability to discriminate between ACL-R

athletes and healthy controls. This finding aligns with previous studies questioning the sensitivity of unilateral balance assessments for detecting residual impairments following ACL injury (Bodkin et al., 2018). In addition, the intrinsic complexity of single-leg tasks may reduce their discriminatory capacity, as they represent a challenging condition even for healthy individuals.

In contrast, dynamic double-leg balance conditions consistently identified greater postural sway and altered neuromuscular activation patterns in ACL-R athletes. Bilateral tasks impose higher demands on neural integration and more closely reflect the bipedal nature of sport-specific movements. The involvement of both the operated and non-operated limbs supports the presence of bilateral central nervous system adaptations following ACL injury, reinforcing the idea that the non-injured limb, in single-leg assessment may not represent a valid internal control for functional comparisons (Zult et al., 2017).

From a clinical perspective, these findings highlight the value of incorporating dynamic double-leg balance assessments into rehabilitation protocols and return-to-sport decision-making.

Trunk perturbations applied during dynamic double-leg conditions appeared to alter medio-lateral postural control in individuals with ACL-R, leading to increased tibialis anterior activation, likely reflecting a compensatory strategy (Nashner, 1982, 1977). This response deviates from the typical proximal–distal activation pattern, in which ankle musculature is minimally involved and postural adjustments are primarily mediated by higher-level neural control mechanisms (Nashner, 1977; Winter, 1995). By contrast, no differences between-groups were observed under single-leg conditions, possibly because the reduced base of support imposes similar stabilization strategies across individuals. These findings further support the greater sensitivity of bilateral dynamic assessments compared with unilateral tasks in detecting persistent motor control impairments following ACL-R.

Evidence from longitudinal case studies further strengthens the clinical relevance of dynamic balance assessment. Improvements observed under static conditions did not necessarily correspond to recovery of dynamic postural control (Marinelli et al., 2024). Dynamic tasks were sensitive to both persistent impairments and progressive adaptations following targeted exercise interventions (La Greca et al., 2025b). This underscores the importance of task complexity in capturing meaningful functional changes over time.

Overall, the results suggest that dynamic balance assessments, particularly when performed under double-leg conditions and combined with neuromuscular analysis,

provide greater sensitivity and ecological validity than static or unilateral testing alone. Integrating these approaches into neurorehabilitation and sports rehabilitation practice may enhance the identification of residual impairments, inform individualized intervention strategies, and improve the evaluation of functional recovery.

CHAPTER 4 – Kinematic and Gait Assessment

4.1 Locomotion as a complex motor task

Human locomotion represents a complex motor task that relies on the coordinated interplay of multiple joints, muscles, and neural control mechanisms. Walking is composed of a cyclic sequence of limb movements, known as the gait cycle, which is conventionally divided into stance and swing phases (Enoka, 2015; Winter, 1995). During the stance phase, the lower limb provides support, absorbs shock, and contributes to forward propulsion (Enoka, 2015). The swing phase allows limb advancement and repositioning for the subsequent step. Within this cycle, the ankle, knee, and hip joints exhibit distinct kinematic and temporal patterns, each contributing to stability, propulsion, and energy efficiency (Cavanagh, 1987).

The control of gait requires precise timing and amplitude of muscle activation. Muscles operate in coordinated synergies to generate joint torques, stabilize the body, and ensure smooth progression (Latash and Singh, 2024). These synergies are dynamically modulated according to task demands, environmental conditions, and individual constraints, highlighting the flexibility and adaptability of the motor system (Latash and Singh, 2024). For example, small changes in walking speed, surface compliance, or external perturbations induce adjustments in joint angles and muscle activation patterns without compromising overall gait function.

Locomotion also depends on continuous integration of sensory information. Somatosensory inputs from muscles, tendons, and joints provide feedback on limb position and load distribution, vestibular signals inform head orientation and balance, and visual cues contribute to obstacle negotiation and spatial orientation. The central nervous system integrates these inputs in real time, generating anticipatory and reactive motor commands that allow precise control of gait and adaptation to perturbations (Latash and Singh, 2024; Winter, 1995).

From a neuromechanical standpoint, gait exemplifies the concept of motor redundancy, in which multiple combinations of joint and muscle actions can achieve the same functional outcome. This redundancy enables flexibility in movement strategies, facilitates compensation for injury or fatigue, and allows the motor system to maintain

performance under varying conditions (Enoka, 2015). Furthermore, locomotion involves both spinal-level pattern generators, which produce rhythmic stepping, and higher-level supra spinal centers that adjust gait for voluntary modifications, obstacle avoidance, and environmental changes.

Overall, human walking is a highly adaptive motor behaviour in which biomechanics, neural control, and sensory integration interact to produce efficient and stable movement. Understanding these fundamental aspects provides the conceptual basis necessary for gait assessment and interpretation of joint kinematics and muscle activity.

4.2 Methods for locomotion assessment

Locomotion is a complex motor task that can be evaluated using a variety of methods, reflecting its multi-segmental and neuromuscular nature (Cavanagh, 1987; Enoka, 2015; Latash and Singh, 2024; Medved, 2021). Depending on the research or clinical objective, assessments may focus on global gait performance, joint-level kinematics, muscle activation, or ground reaction forces. The diversity of methods allows the selection of tools with different levels of complexity and detail.

Simple and widely used methods include spatiotemporal parameters, such as walking speed, step length, cadence, stance duration, and symmetry (Horwood and Chockalingam, 2023; Medved, 2021). These measures, often applied in clinical or practical contexts, provide a quick overview of gait performance and can detect gross asymmetries or functional deficits. Their main advantages are simplicity, low cost, and ease of implementation. However, they do not directly evaluate joint control or muscle activation and may miss subtle compensations or neuromuscular alterations. In research and advanced clinical assessments, locomotion is frequently evaluated using kinematic analysis (Cavanagh, 1987; Enoka, 2015) through video-based or sensor-based systems (Medved, 2021). These approaches provide detailed information on segmental movements, joint coordination, and movement strategies, allowing a finer characterization of gait patterns and the detection of subtle biomechanical deviations that simple spatiotemporal measures may overlook.

The kinematics of locomotion can be assessed using different approaches. Simple systems allow measurement in a single plane, providing basic joint angle information during gait. More advanced 3D motion capture systems, often marker-based, offer a complete reconstruction of joint movements in all spatial planes. These systems can be integrated

with force plates, inertial sensors, or surface electromyography (EMG) to obtain a comprehensive view of gait mechanics, muscle activity, and neuromuscular control (Latash and Singh, 2024; Medved, 2021).

In summary, locomotion assessment can be conducted using tools and approaches of variable complexity. Simple evaluations provide rapid screening of gait performance, whereas integrated methodologies (3D kinematics combined with EMG and/or force platforms) offer more detailed and functionally relevant information. This advanced approach is particularly useful in clinical and research settings and forms the foundation for segmental and global joint analysis.

4.2.1 Joint kinematic analysis

Joint kinematic analysis represents a core component of locomotion assessment, as it describes how body segments and joints move over time during gait. By quantifying joint angular displacements, kinematic analysis provides direct information on movement organization and biomechanical strategies adopted to accomplish locomotion (Enoka, 2015; Medved, 2021). From a motor control perspective, joint kinematics reflects the interaction between neural commands and mechanical constraints, offering insight into how movement is shaped to meet functional demands (Enoka, 2015).

Kinematic data are typically expressed as angle–time profiles, obtained by tracking joint rotations throughout the gait cycle. These profiles are often normalized to the percentage of the gait cycle, allowing comparison across individuals, conditions, or experimental sessions (Medved, 2021). The analysis of angle–time curves enable the identification of joint-specific patterns during key phases of gait, such as initial contact, mid-stance, push-off, and swing, as well as the evaluation of range of motion, timing of peak angles, and movement smoothness (Cavanagh, 1987).

The study of joint kinematics is particularly informative for examining dynamic joint behavior, as it highlights how joints contribute to propulsion, shock absorption, and limb advancement. Deviations from typical kinematic patterns may indicate altered motor strategies, compensatory mechanisms, or inefficient movement organization, even in the absence of overt spatiotemporal abnormalities (Enoka, 2015; Medved, 2021). For this reason, joint kinematic analysis is widely used in both clinical and research settings to investigate gait alterations associated with neurological, musculoskeletal, or sport-related conditions.

While angle–time representations provide detailed information on the behavior of individual joints, they primarily describe joint motion as a function of time. To further explore how joints interact with each other during locomotion, alternative representations have been proposed in the biomechanics literature. Among these, angle–angle diagrams, although less commonly used, allow the examination of coordination between adjacent joints by describing the relationship between their angular displacements, offering insight into inter-joint movement organization beyond time-based representations (Cavanagh, 1987; Enoka, 2015).

4.3 Angle–angle diagrams

The assessment of locomotion is commonly based on two main approaches. The first includes spatiotemporal parameters, which are widely used across different clinical contexts. These variables provide general information on gait performance and symmetry. The second approach relies on joint angle–time profiles of single joints, typically adopted in research settings or advanced clinical evaluations (Horwood and Chockalingam, 2023; Medved, 2021). Despite their broad use, these metrics present relevant limitations.

They are strongly dependent on walking speed, which affects both spatiotemporal variables and joint kinematic patterns. More complex analyses may also be difficult to synthesize and interpret in clinical contexts (Filli et al., 2018). Moreover, when gait alterations involve multiple joints simultaneously, traditional metrics often provide fragmented information rather than an integrated description of the movement pattern.

The literature highlights that walking speed influences not only basic gait parameters but also the relationships between joint angular changes across the stride cycle (Enoka, 2015; Winter, 1995). For this reason, speed control, for example through treadmill walking, is methodologically relevant, although it is rarely implemented in a systematic manner. In populations with gait impairments (e.g., neurological disease), locomotor alterations do not usually concern a single variable. Instead, pathological gait patterns emerge from the combination of multiple joint adaptations. Overall, traditional methods do not always allow an integrated and clinically intuitive interpretation of gait patterns.

To overcome these limitations, an alternative approach based on angle–angle diagrams has been proposed (Cavanagh, 1987; Hershler and Milner, 1980a). This method represents the angular changes of two adjacent joints plotted against each other. Time is not displayed

on the axes but is implicitly embedded in the progression of the loop. The resulting closed loop describes the entire gait cycle.

With this approach, the focus shifts from analysing how much and when a single joint moves to understanding how two joints cooperate during locomotion (Hershler and Milner, 1980a).

Angle–angle diagrams allow the extraction of both quantitative and qualitative information. The area of the loop represents the conjoint range of motion (ROM) of two adjacent joints, such as the hip–knee or knee–ankle complex. The perimeter of the loop reflects the degree of inter-joint coordination. From an interpretative perspective, a reduction in loop area indicates a global limitation of movement, whereas an increase in perimeter suggests poor coordination or rigid, “robot-like” movement strategies (Hershler and Milner, 1980b).

Beyond numerical values, the shape of the loop provides immediate qualitative information. Altered movement patterns can often be identified visually, facilitating interpretation (Cavanagh, 1987). This dual nature represents a key strength of the method, combining visual synthesis with quantitative measurement. Compared with conventional gait metrics, angle–angle diagrams integrate angular and temporal information within a single representation (Hershler and Milner, 1980a).

They allow straightforward comparison between different gait patterns and highlight compensatory adaptations involving multi-joint coordination, which may not be evident in angle–time profiles (Hershler and Milner, 1980a). Angle–angle diagrams can also reflect events occurring in the contralateral limb and capture global locomotor strategies. Despite these advantages, the systematic application of angle–angle diagrams in populations with neurological or motor impairments affecting gait remains limited. This represents a clear gap in the literature, considering the potential of this approach to provide integrated and clinically meaningful insights into locomotor coordination.

The aim of this section is to introduce the angle–angle diagram as an alternative and complementary tool for gait analysis. This method allows the integrated description of joint ROM and inter-joint coordination. The studies presented in this chapter apply this approach in individuals with autism spectrum disorder, extend their quantitative use in persons with Multiple Sclerosis (MS), and integrate kinematic data with EMG to strengthen the interpretation of neuromuscular control during locomotion.

Methods

Setup for gait analysis

The kinematic gait analysis was performed by using the SMART Motion Capture System (BTS, Bioengineering, Milan, Italy), equipped with optoelectronic cameras able to reconstruct the three-dimensional (3D) trajectories provided by specific reflective markers placed on the anatomic joint points of the subjects. Four cameras were used to delimit the shooting volume around the treadmill, on which each participant walked for 30 s at 1, 2 and 3 km/h while sEMG activity in the leg muscles was recorded. A 22-segment model was created for each subject (figure 4.1 B), and the reflective markers were placed at the seventh cervical vertebra, the first sacral vertebra, the acromion, the lateral epicondyle of the elbows, and the distal epiphysis of the radius of both upper limbs; in both lower limbs, the markers were placed at the femoral greater trochanter, the lateral condyle of the femur, the lateral malleolus, the calcaneus, and the distal epiphysis of the fifth metatarsal. The 9-element data were considered in the analysis of the lower limbs using the SMART Analyser (BTS, Bioengineering), in which three angles over time were plotted in the sagittal plane motion, namely the hip angle (absolute), the knee angle (relative), and the ankle angle (relative). The hip angle is formed by the thigh axis and the vertical of the hip passing through the pelvis. The knee angle is formed by the extension of the thigh axis and shank axis, whereas the ankle angle is formed by the shank axis and foot axis, passing over the lateral malleolus and the fifth metatarsal segment. Two gait cycles for each leg were sampled; the average was calculated to plot two angle–angle and angle–time relations. The selected cycles were as close to the middle of the gait test duration as possible so that the participants could adapt themselves to the preselected speed on the treadmill.

The angle–angle diagrams were determined according to Cavanagh (1987). Five points were chosen (Figure 4.1C): toe off (TO), maximum knee flexion (MKF), maximum hip flexion (MHF), foot support (FS), and maximum flexion of the knee during the contact phase (MKF-C). The frame with the minimum value of the knee angle was selected for MKF, the frame with the minimum value of the hip angle was selected for MHF, and the frame with the minimum value of the knee angle during the contact phase was selected for MKF-C. For TO and FS, the considered frames corresponded to those in which the value of the X vector of velocity changed from negative to positive and vice versa (Zeni et al., 2008). The areas and perimeters of the angle–angle diagrams were calculated by using AutoCAD 2023 (Autodesk, San Rafael, CA, USA). The shape of the loops formed

by the angle–angle diagrams was calculated by dividing the perimeter multiplied by the root-mean-square of the area (Hershler and Milner, 1980a).

$$Shape = \frac{Perimeter}{\sqrt{Area}}$$

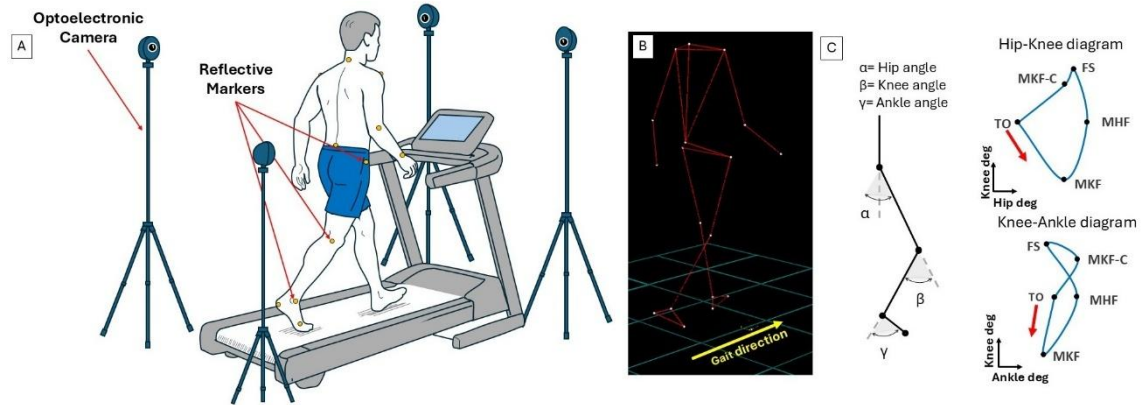


Figure 4.1. Representative setup of gait analysis. A) Kinematic and synchronized sEMG activity during walking on treadmill. B) Body reconstruction obtained by the Motion Capture System. C) Stick figure of the lower limb with hip angle (α), knee angle (β), and ankle angle (γ), and angle-angle diagrams of hip-knee angle and knee-ankle angle of one autism spectrum disorder (ASD) participant (TO = toe off, MKF = maximum knee flexion, MHF = maximum hip flexion, FS = foot strike, MKF-C = maximum knee flexion during the contact phase).

Statistical analysis

In each of the studies the Statistics Toolbox implemented in the MATLAB software interface (version 204 R2024b; MathWorks, Natick, MA, USA), and the XLSTAT 2013.2.07 software (Addinsoft; New York, NY, USA) were used for statistical analysis. Shapiro-Wilks's W test was used to assess the normality distribution of the variables. Non-parametric tests were implemented due to the absence of a standard Gaussian distribution. To determine the differences between the groups for the anthropometric measured at baseline the Mann-Whitney U test was used.

The angle–time profiles were compared with the Kolmogorov–Smirnov test. The other dependent variables were compared with the Wilcoxon test for within-group comparisons and by a Mann–Whitney U test for between-group comparisons to determine differences. The Bonferroni correction was used to adjust the p value in relation to the number of contrasts carried out. Statistical significance was set at $\alpha = 0.05$, and the effect size was

determined using Hedges' g , which was considered to be small at $g < 0.5$, moderate at $0.5 < g < 0.8$, and large at $g > 0.8$ or was estimated by dividing the Z -statistic by the square root of the sample size ($r = z/\sqrt{N}$). The ES values were small at $r < 0.5$ (Cohen, 2013a). In each study no significant anthropometric variables differences were found (age, height, body mass, BMI and physical activity level), confirming group that the groups were well-matched.

In the case report to evaluate the long-term effects of the intervention for each variable the following formula was used:

$$\Delta\% = \left[\frac{\text{follow up} - T0}{T0} \right] * 100$$

In addition, a linear regression model was constructed across the follow-ups, this model examined the trend of the dependent variables over the time. The significance level (α) was set to 0.05.

Study 1

Angle-Angle Diagrams in the Assessment of Locomotion in Persons with Multiple Sclerosis: A Preliminary Study. (Di Giminiani et al., 2022)

Participants

Persons with multiple sclerosis (PwMS) with a low level of disability that allowed them to walk without any external support were recruited from the Centre for the Diagnosis and Treatment of Demyelinating Diseases in the Hospital of L'Aquila-Sulmona-Avezzano (8 females/2 males: age 52.7 ± 10.3 years; stature 165.0 ± 8.8 cm; body mass 67.2 ± 11.2 kg; body mass index 24.8 ± 4.5 kg·m⁻²). HCs who had similar characteristics to those of the PwMS were recruited (8 females/2 males: age 52.3 ± 10.9 years; stature 165.5 ± 7.3 cm; body mass 60.7 ± 9.5 kg; body mass index 22.1 ± 2.7 kg·m⁻²). The inclusion criterion for PwMS was a multiple sclerosis diagnosis with an EDSS score of 4/5, whereas the exclusion criteria were other diseases of the neuromuscular system. For the healthy controls (HC), the inclusion criteria were an age of between 50 and 70 years and motor skills that did not prejudice walking, whereas the exclusion criteria were growth and development anomalies, pharmacological therapy and present or history of musculoskeletal and nervous diseases.

The walking speed was determined through the timed 25-foot walk (T25FW) (Motl et al., 2017). On the treadmill, the experimental group walked at a 1-km/h speed for a duration of 30 s (the mean value obtained during the T25FW test), while the HCs performed for the same duration at 1 and 3 km/h. The experimental procedure was conducted according to the ethical standards of the Declaration of Helsinki.

Results

The area of the knee-angle/ankle-angle loops between PwMS vs. HC-1 at the same speed showed significant differences ($p < 0.05$, ES = 0.80, figure 4.2A), whereas the hip-angle/knee-angle loops did not show significant differences ($p > 0.05$; ES = 0.12). The perimeter differences at a speed of 1 km/h approaches significance between the knee-angle/ankle-angle loops ($p = 0.086$, ES = 0.53), but not between the hip-angle/knee-angle loops ($p = 0.678$; ES = 0.10) (figure 4.2B). The area and perimeter comparisons within the HCs when they walked at different speeds (HC-1 vs. HC-3) were significant either between the hip-angle/knee-angle loops ($p < 0.001$, ES = 1.34/1.42) or between the knee-

angle/ankle-angle loops ($p < 0.001$, $ES = 0.85/1.27$) (figure 4.2A, B).

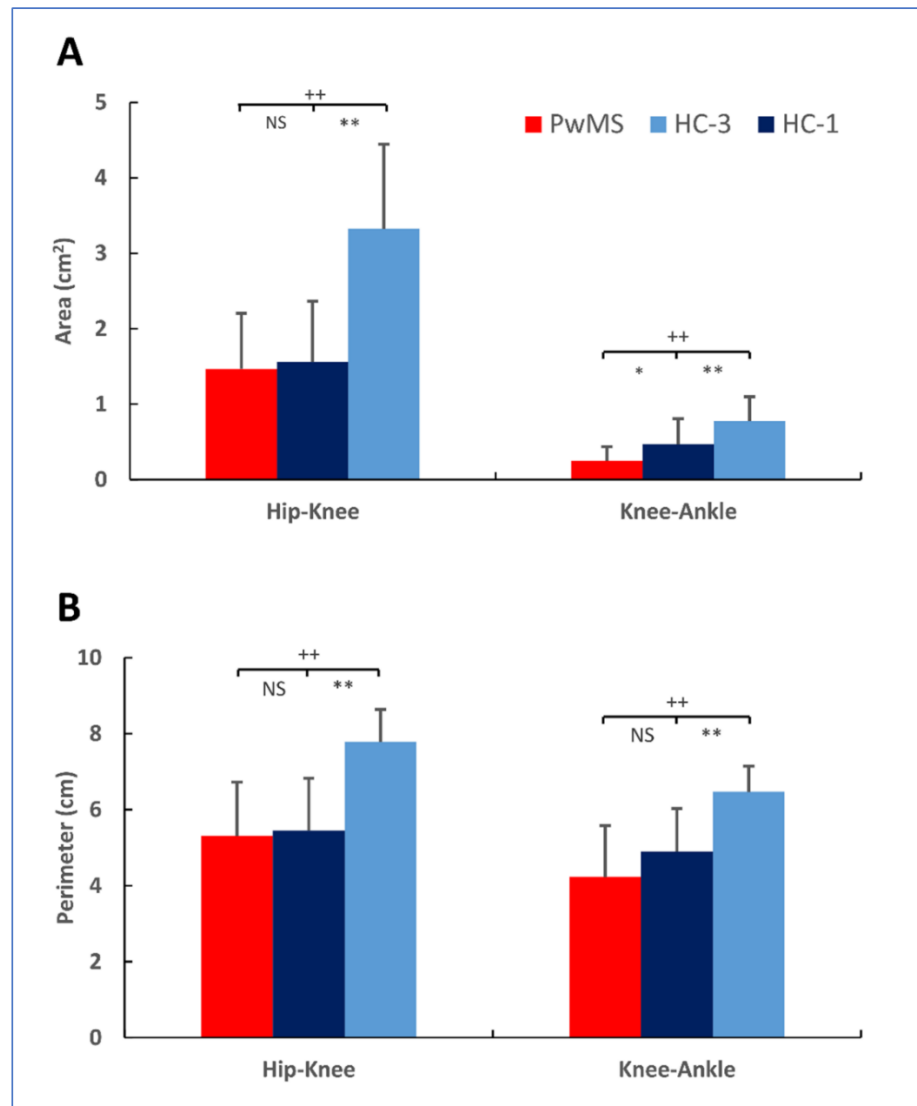


Figure 4.2. Values (SD) of A) areas and B) perimeters in the hip-knee and knee-ankle diagrams are reported in both groups (PwMS, HC-1 = 1 km/h and HC-3 = 3 km/h). * Difference between PwMS and HC-1 ($p < 0.05$); ++ Difference between HC-3 and PwMS ($p < 0.001$); ** Difference between HC-3 and HC-1 ($p < 0.001$); NS = no significant difference between HC-1 and PwMS ($p > 0.05$). (From: Di Giminiani et al., 2022)

These findings suggest that the angle-angle diagram, when determined between the knee and the ankle joints, is a sensitive approach to detect gait alterations in ROM and coordination in PwMS with respect to controls while walking on a treadmill at the same speed. In addition, this method show high reliability of angular variables (mean ICC 0.82).

Study 2

Locomotion and Postural Control in Young Adults with Autism Spectrum Disorders: A Novel Kinesiological Assessment. (Di Giminiani et al., 2024)

Participants

The study included twenty young adult males: 10 with typical development (TD; age: 23.5 ± 2.5 years; height: 1.79 ± 0.09 m; body mass: 70.1 ± 10.9 kg; body mass index: 22.0 ± 2.6 kg·m⁻²; leg preference: right) and 10 with autism spectrum disorder (ASD; age: 22.4 ± 3.4 years; height: 1.75 ± 0.05 m; body mass: 78.4 ± 10.6 kg; body mass index: 25.2 ± 3.9 kg·m⁻²; leg preference: 9 right, 1 left).

The sample size estimation for the primary outcomes (angle–angle diagram and sEMG synchronised during walking) was computed a priori using of statistical software for power analysis (G*Power 3.1.9.4, Heinrich Heine-Dusseldorf University, Düsseldorf, Germany). The computation was performed in relation to the study design (*t*-test family for parametric and nonparametric distribution), setting the effect size (ES), and using the protocol for a power analysis (test attributes, large ES [1.8], $\alpha = 0.05$, power $[1 - \beta] = 0.96$, total sample size $n = 20$ participants). The diagnosis of ASD was as follows: 8 with Level 1 and 2 with Level 2 (mean QI = 81.6, standard deviation = 18.8). The inclusion criteria were the absence of any history of injury, fracture, or morphological changes in the lower extremities. Individuals with ASD were recruited from the Reference Regional Centre for Autism (CRRA) in L'Aquila (Italy). They had been diagnosed according to the criteria of the Diagnostic and Statistical Manual of Mental Disorders (5th ed., DSM-5; American Psychiatric Association [APA], 2013) (American Psychiatric Association, 2013) and the Autism Diagnostic Observation Schedule-2 (ADOS-2) (Lord et al., 2012). Each participant had ADOS-2 scores that were above the cut-off (mean communication and social interaction = 12.2 ± 2.8 ; mean stereotyped behaviours and restricted interests = 0.8 ± 0.8). The experimental procedure was conducted according to the ethical standards of the Declaration of Helsinki. The informed consent was obtained from all the participants and/or their legal guardians.

The participants became familiar with the experimental procedure before the measurements (a different day). The measurements lasted about 2 h for each participant, and the protocol was carried out between 10:00 and 13:00 h. The Joint kinematics analysis of the gait was performed both with angle-time profiles and angle-angle diagrams.

Results

The angle–time profiles determined at the hip, knee, and ankle joints did not show significant differences between the right and left leg within each group (ASD or TD; $p > 0.05$) (Figure 4.3). Moreover, there were no significant angle–time profile differences between the ASD and TD groups ($p > 0.05$) (Figure 4.3).

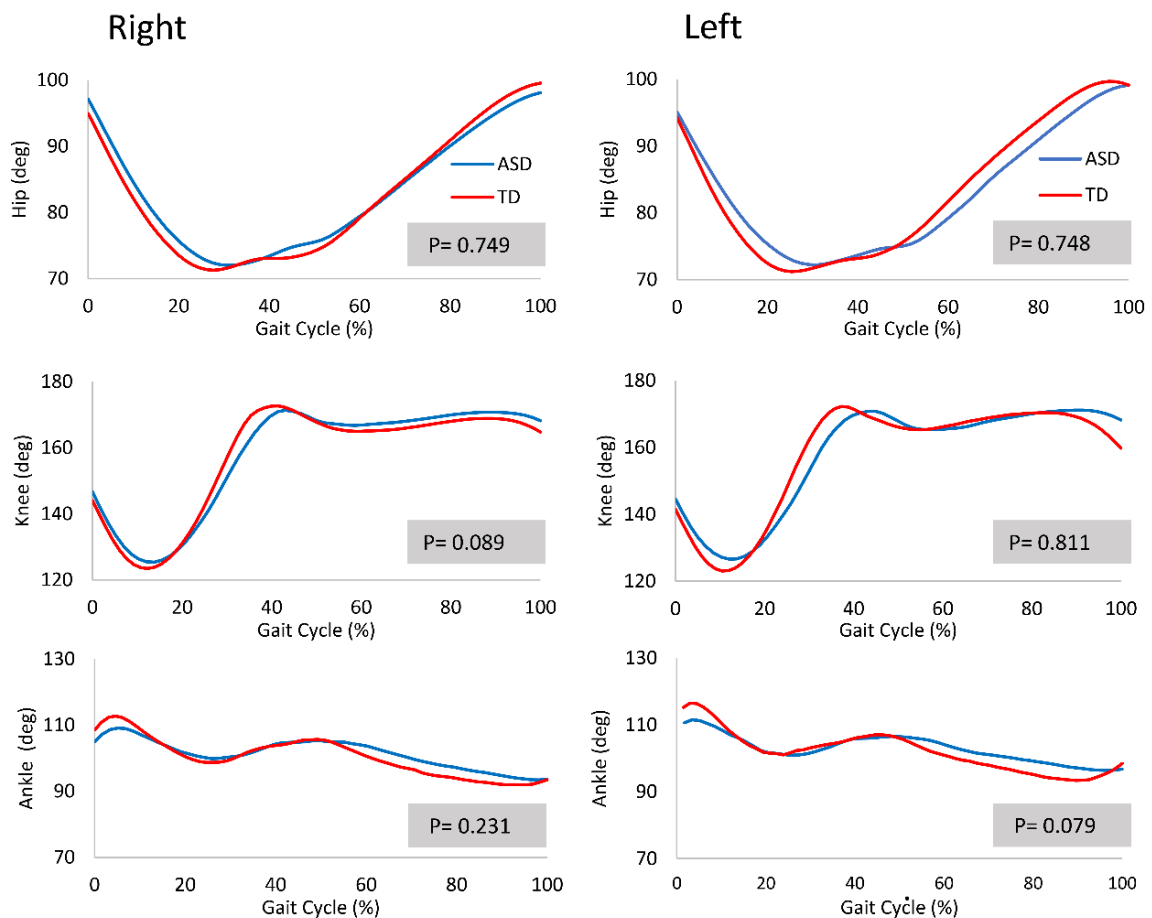


Figure 4.3. Angle–time profiles (hip, knee, and ankle) during walking on a treadmill (speed = 3 km/h) in the experimental (ASD group) and typical developmental (TD group) subjects. (From: Di Giminiani et al., 2024)

The area, perimeter, and shape of the hip–knee and knee–ankle diagrams did not show significant differences between the right and left legs within each group (ASD or TD; $p > 0.05$). On the contrary, the area described by the knee–ankle diagrams for the right leg showed a significant difference between the ASD and TD groups ($p < 0.05$; ES = 2.32). However, the shape and perimeter defined by the knee–ankle diagram were not significantly different between the groups ($p > 0.05$). For the left leg, there were no

significant differences between the ASD and TD groups ($p > 0.05$; Figure 4.4). The hip–knee diagrams showed no significant differences in the area, perimeter, and shape between the ASD and TD groups ($p > 0.05$; Figure 4.4).

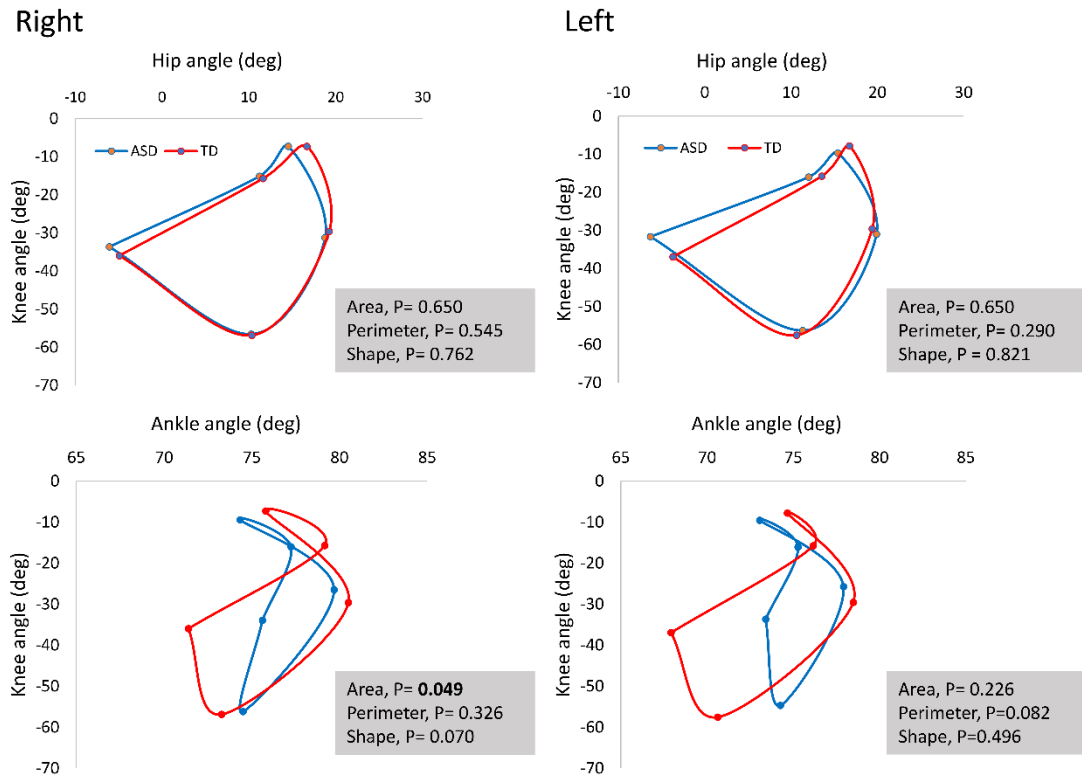


Figure 4.4. Angle–angle diagrams (hip–knee and knee–ankle) during walking on a treadmill (speed = 3 km/h) in the experimental (ASD group) and typical developmental (TD group) subjects. (From: Di Giminiani et al., 2024)

The angle–angle diagrams detected group differences not captured by traditional angle–time analysis, with ankle angle–angle parameters showing excellent sensitivity and specificity in the receiver operating characteristic (ROC) curve (area under curve= 94.4%) for discriminating between ASD and TD (figure 4.5).

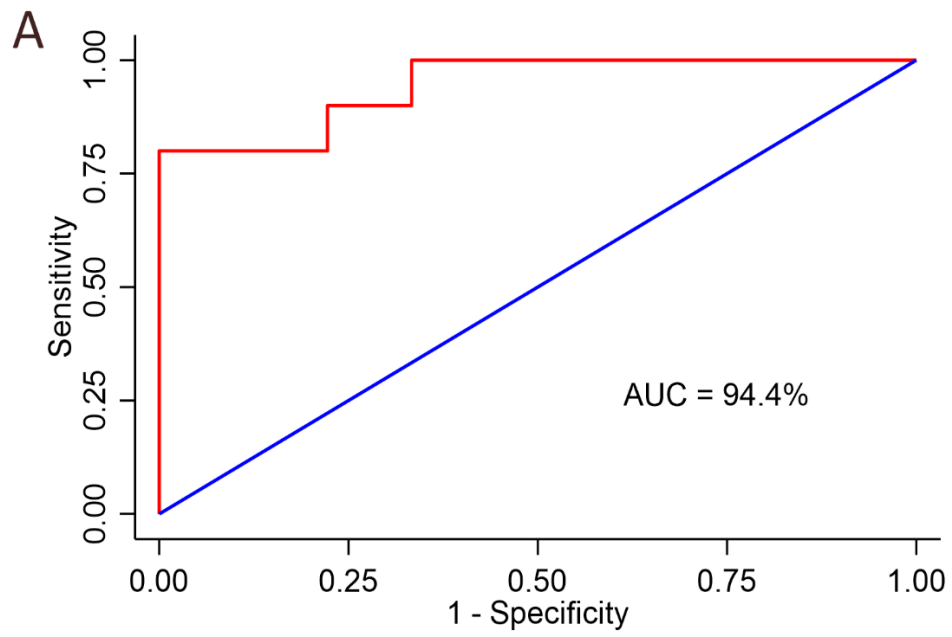


Figure 4.5. ROC curve of knee–ankle diagram during walking. (From: Di Giminiani et al., 2024)

Study 3

Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. (La Greca et al., 2025b)

Participant

This case report investigated the long-term (5-year) effects of individualized whole-body vibration (WBV) training on locomotion in a person with multiple sclerosis (PwMS).

The intervention was assessed during 5 follow-ups (T0, T1, T2, T3 and T4).

Results

The hip–knee diagrams showed an increase in the overlap of loops between weakest leg and strongest leg (from T0 to T4) at all speeds (figure 4.6). The hip–knee and knee–ankle diagrams exhibited significant regression trend over time only in the strongest leg, perimeter, and shape at a speed of 1 km/h ($p < 0.05$). There were no significant changes in any of the other variables ($p > 0.05$). The perimeter exhibited a higher Δ in the weakest limb at T4 (hip–knee = 17% and knee–ankle = 36%). At T2, the participant became able to walk on the treadmill at 2 km/h and 3 km/h, unfeasible speeds for the first two test times (T0 and T1).

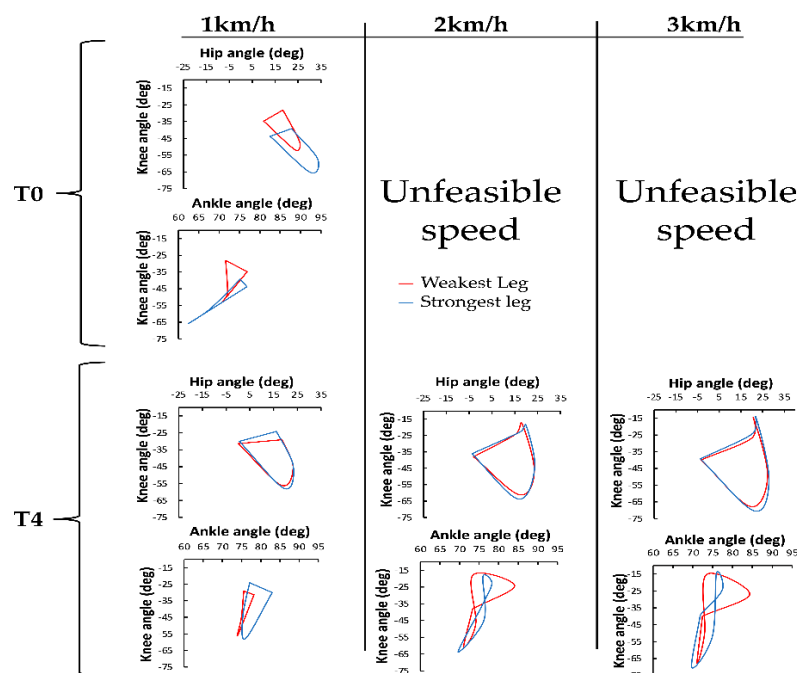


Figure 4.6. The angle-angle diagrams of strongest and weakest legs during gait on treadmill. Difference between the baseline (T0) and the last follow-up (T4). (From: La Greca et al., 2025b)

4.4 Integration of kinematics and electromyography

Locomotion is a complex motor task resulting from the interaction between biomechanical constraints and neuromuscular control mechanisms (Enoka, 2015; Latash and Singh, 2024). Kinematic analysis describes the observable outcome of movement, such as joint trajectories and segment coordination (Cavanagh, 1987), but it does not directly reflect the neural strategies responsible for movement generation. For this reason, kinematic data alone may be insufficient to fully interpret motor behavior, particularly when different neuromuscular strategies lead to similar movement patterns (Enoka, 2015; Latash and Singh, 2024).

Surface electromyography (sEMG) provides complementary information by capturing the timing and magnitude of muscle activation during gait. EMG data allows the investigation of how muscles are recruited and coordinated to produce locomotor patterns, offering insight into underlying control strategies, motor adaptations, and compensatory mechanisms (Latash and Singh, 2024). According to Enoka (2015), integrating EMG with movement analysis is essential to understand how the nervous system organizes muscle activity to meet task demands under varying mechanical and environmental conditions.

The functional interpretation of EMG signals is strongly dependent on their temporal relationship with movement events. During walking, muscle activation patterns are phase-specific and closely linked to distinct functional demands of the gait cycle, such as weight acceptance, propulsion, and limb advancement (Enoka, 2015; Latash and Singh, 2024). Synchronizing EMG with kinematic data enables the association of muscle activity with specific joint motions and gait phases, thereby improving the interpretability of both signals (Medved, 2021; Winter, 1995). Without this synchronization, EMG activity may be difficult to contextualize and may lead to ambiguous conclusions regarding muscle function.

The integration of kinematics and EMG is particularly valuable for identifying neuromuscular alterations that may not be evident from kinematic variables alone. Changes in muscle timing, amplitude, or coordination can occur in the absence of clear differences in joint angles or spatiotemporal parameters. In such cases, EMG analysis enhances the sensitivity of gait assessment by revealing altered activation strategies, increased co-contraction, or rigid motor patterns that reflect compensatory control mechanisms (Latash and Singh, 2024).

In the studies presented in this chapter, kinematic data were systematically synchronized with sEMG recordings during stance phase of the gait. This integrated approach was

adopted to move beyond a purely descriptive analysis of joint motion and to provide a functional interpretation of locomotor patterns. By combining joint kinematics with muscle activation profiles, it was possible to better characterize neuromuscular organization during gait and to identify differences between groups or conditions that were not detectable through kinematic analysis alone.

The Integration of kinematics and electromyography represents a robust methodological framework for the assessment of locomotion. It allows a more comprehensive evaluation of gait by linking movement outcomes to their neuromuscular determinants, thereby improving the functional relevance of gait analysis in both research and advanced clinical contexts.

Methods

Setup for synchronised sEMG activity during gait

The sEMG activity was detected in the following muscles of both the legs: vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and lateralis gastrocnemius (LG). sEMG was recorded using bipolar electrodes (Ambu Neuroline 720 00-S/25, electrode diameter 45×22 , interelectrode distance 2.5 cm, Ambu A/S, Baltorpbakken, Denmark) in 4 muscles: VL, BF, TA, and LG (figure 4.7). The electrodes were placed on both legs, according to the international surface EMG for non-invasive assessment of muscles (SENIAM) recommendations (Hermens et al., 2000). To minimize the impedance ($<5 \text{ k}\Omega$) before the electrodes' application, the skin was shaved, gently abraded with sandpaper (P320) without producing redness, and cleaned with alcohol. The exact location on the muscle belly was marked to ensure correct sensor positioning across the inter-day EMG recordings. In addition, the electrodes and the modules were fixed with elastic strips (Flexa Elast, Pic Solution, Pikkdare S.p.A.) to prevent motion artifacts. The wireless data synchronization unit (MuscleLab 6000, Ergotest-innovation, Porsgrunn, Norway) characteristics were as follows: a built-in radio frequency (RF) module ML6RFM02 with 2.4 GHz and 1 mW and a typical wireless range in an open space of 20 m. The modules of the MuscleLab characteristics were as follows: RF characteristics of 2.4 GHz, 1 mW; sample rate of 1.2 kHz; high-pass filter of 20 Hz; bandwidth of 20–500 Hz; input signal range of $\pm 5 \text{ mVp-p}$; noise of $14 \text{ }\mu\text{Vp-p}$ ($2.2 \text{ }\mu\text{V}_{\text{RMS}}$); and resolution of $0.33 \text{ }\mu\text{V/bit}$. Utilizing a preamplifier that was situated near the pickup point and directly connected to the electrodes, the raw sEMG signal was amplified and filtered at 1.2 kHz. The frequencies of 20 Hz and 500 Hz were used to band-pass filter the signals using a fourth-

order Butterworth filter. The preamplifier's common mode rejection ratio was 100 dB. In accordance with the standards for reporting sEMG data (International Society of Electrophysiology and Kinesiology, <https://isek.org/resources/>, accessed on 24 July 2025), the EMG signal was converted to a root-mean-square (RMS) signal using a hardware circuit network (frequency response 20–500 kHz, integrating moving average filter with 100 ms width, total error $\pm 0.5\%$). The peak sEMG_{RMS} value of the respective muscle during the stance phase of walking was used to normalize the sEMG_{RMS} signals of the muscles during walking (Cronin et al., 2015).

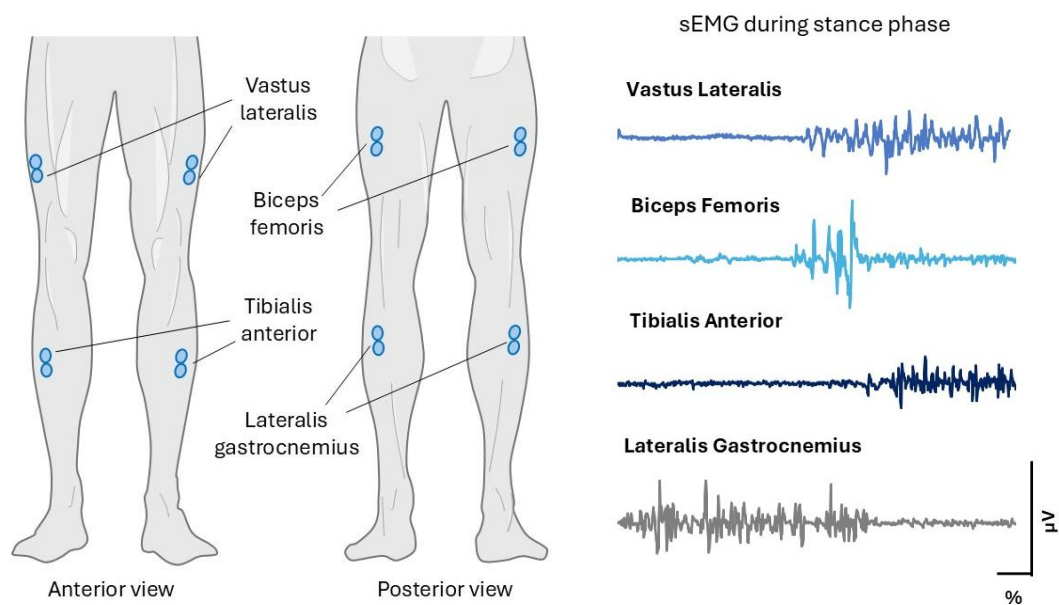


Figure 4.7. On the left, a representative example of electrode placement. On the right a representative sEMG activity of the dominant leg of one ASD participant during the stance phase.

Statistical analysis

In each of the studies the XLSTAT 2013.2.07 software (Addinsoft; New York, NY, USA) and/or Statistics Toolbox implemented in the MATLAB software interface (version 2014b; MathWorks, Natick, MA, USA), were used for statistical analysis.

Shapiro-Wilks's W test was used to assess the normality distribution of the variables. Non-parametric tests were implemented due to the absence of a standard Gaussian distribution. To determine the differences between the groups for the anthropometric measured at baseline, the Mann-Whitney U test was used.

The dependent variables were compared with the Wilcoxon test for within-group

comparisons and by a Mann–Whitney U test for between-group comparisons to determine differences. The Bonferroni correction was used to adjust the p value in relation to the number of contrasts carried out. Statistical significance was set at $\alpha = 0.05$, and the effect size was determined using Hedges' g , which was considered to be small at $g < 0.5$, moderate at $0.5 < g < 0.8$, and large at $g > 0.8$ or was estimated by dividing the Z-statistic by the square root of the sample size ($r = z/\sqrt{N}$). The ES values were small at $r < 0.5$ (Cohen, 2013).

In each study no significant anthropometric variables differences were found (age, height, body mass, BMI and physical activity level), confirming group that the groups were well-matched.

In the case report, to evaluate the long-term effects of the intervention for each variable, the following formula was used:

$$\Delta\% = \left[\frac{\text{follow up} - T0}{T0} \right] * 100$$

In addition, a linear regression model was constructed across the follow-ups, this model examined the trend of the dependent variables over the time. The significance level (α) was set to 0.05.

Study 1

Angle-Angle Diagrams in the Assessment of Locomotion in Persons with Multiple Sclerosis: A Preliminary Study. (Di Giminiani et al., 2022)

Participant

These results refer to the same group of PwMS included in the sample of study 1 (Di Giminiani et al., 2022).

Results

The sEMG_{RMS} activities in the leg muscles did not show significant differences among the ten contact phases in either group ($p > 0.05$; Figure 4.8); however, in PwMS, the confidence intervals were wider than those of the controls. Activation of the vastus lateralis was significantly higher in PwMS than in HCs, even when they walked at higher speeds (PwMS vs. HC-1, $p < 0.001$, ES = 1.90; PwMS vs. HC-3, $p < 0.001$, ES = 1.84). Similarly, biceps femoris muscle activation was higher in PwMS than in HC-1 ($p < 0.05$, ES = 1.30) but not when PwMS were compared to HC-3 ($p < 0.001$; ES = 1.84). In contrast, PwMS showed less activation in the tibialis anterior when compared to HC-1 ($p < 0.001$; ES = 1.76) and to HC-3 ($p < 0.001$; ES = 1.84). A similar pattern showed the lateral gastrocnemius muscles in PwMS versus HC-1 ($p < 0.001$, ES = 1.70) and versus HC-3 ($p < 0.001$; ES = 1.85). By increasing the speed from 1 km/h to 3 km/h, HCs increased their activation in the vastus lateralis (HC-1 vs. HC-3, $p < 0.05$, ES = 1.91), biceps femoris (HC-1 vs. HC-3, $p < 0.001$, ES = 1.87) and lateral gastrocnemius muscles (HC-3 vs. HC-1, $p < 0.05$, ES = 1.58). In contrast, the tibialis anterior did not increase its activation at higher speeds (HC-1 vs. HC-3, $p > 0.05$; ES = 0.47).

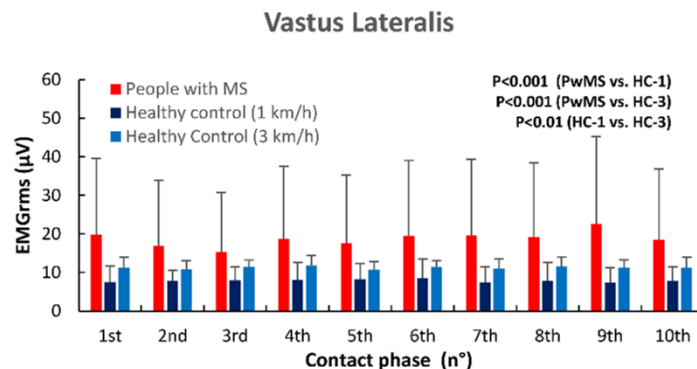


Figure 4.8. Cont.

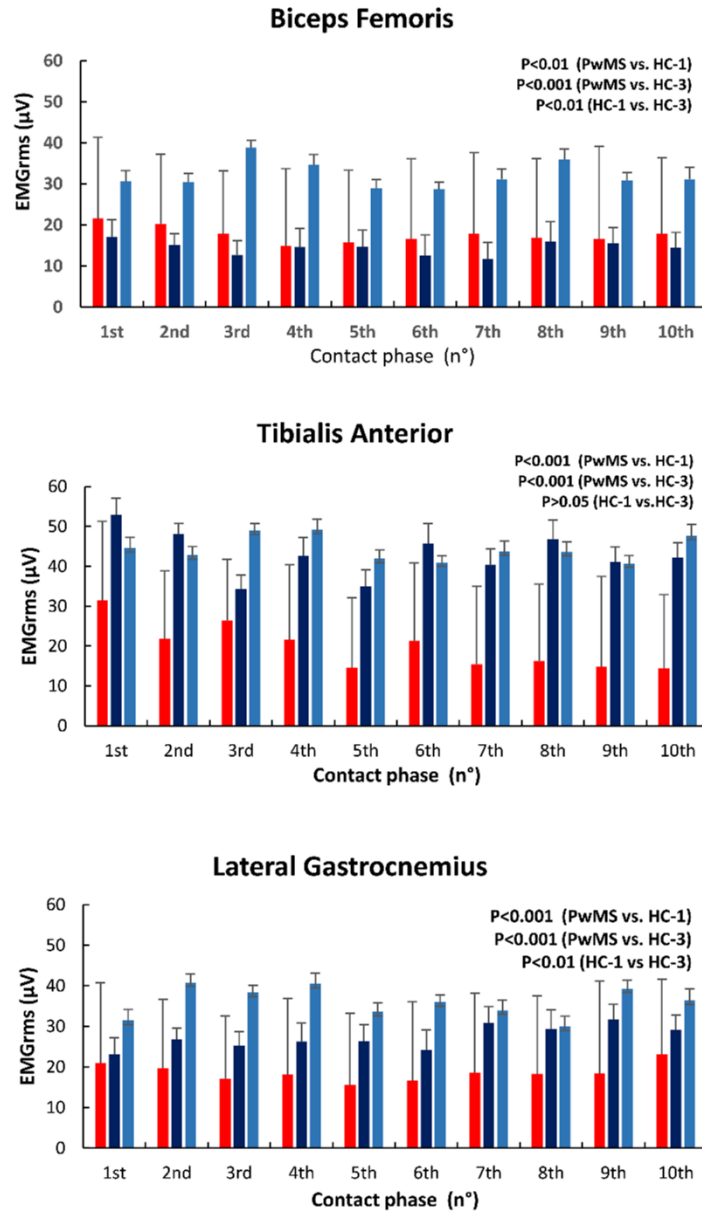


Figure 4.8. Average sEMG_{RMS} (SD) values of leg muscles (vastus lateralis, biceps femoris, tibialis anterior and lateral gastrocnemius) recorded in 8 participants for the weakest lower limb in PwMS (1 km/h) and for the dominant limb in HCs (1 km/h and 3 km/h). (From: Di Giminiani et al., 2022)

PwMS exhibited altered and more variable muscle activation patterns compared with healthy controls, characterized by higher proximal muscle activity and reduced distal muscle activation, with a limited ability to modulate neuromuscular output in response to increasing walking speed.

Study 2

Locomotion and Postural Control in Young Adults with Autism Spectrum Disorders: A Novel Kinesiological Assessment. (Di Giminiani et al., 2024)

Participants

These results refer to the same group of ASD and TD people, included in the sample of study 2 (Di Giminiani et al., 2024).

Results

During the contact phase of gait, the sEMG activity of the lateral gastrocnemius (LG) was significantly higher in the TD group compared with the ASD group ($p < 0.05$; $ES = 1.39$), whereas the other muscles showed similar activation in both groups ($p > 0.05$) (Figure 4.9).

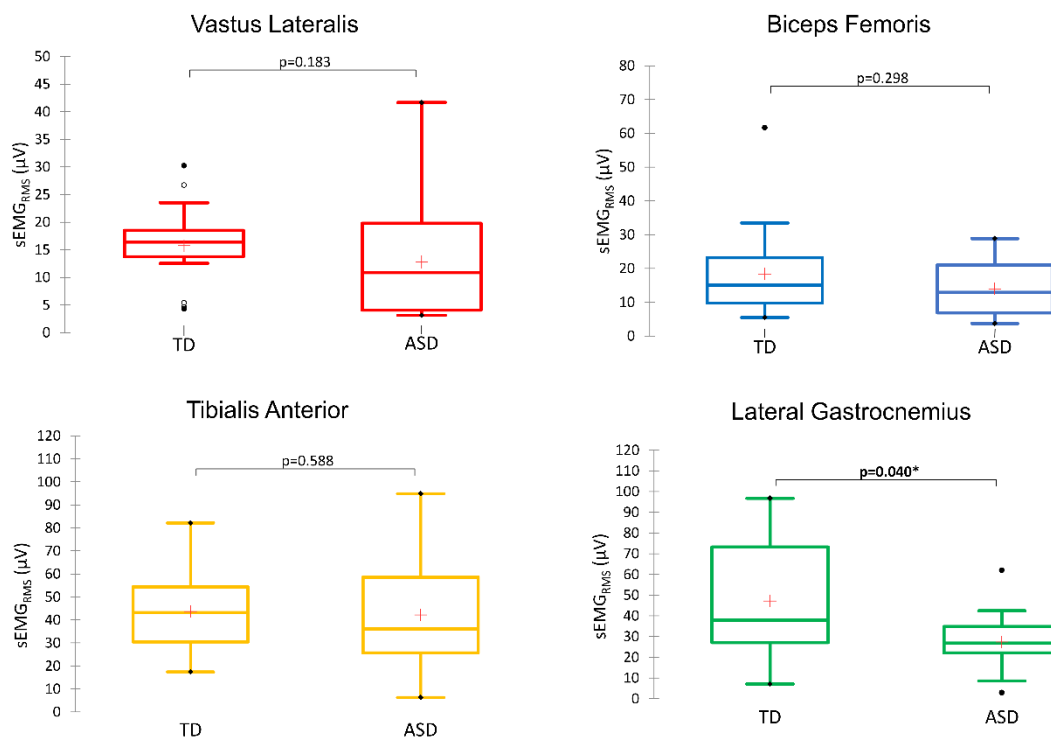


Figure 4.9. sEMG activity in the stance phase of two consecutive steps in the experimental (ASD group) and typical developmental (TD group) subjects. * $p < 0.05$. (From: Di Giminiani et al., 2024)

Moreover, the sEMG parameter LG showing excellent sensitivity and specificity in the receiver operating characteristic (ROC) curve (area under curve= 81.1%) (figure 4.10) for

discriminating between ASD and TD during stance phase of gait.

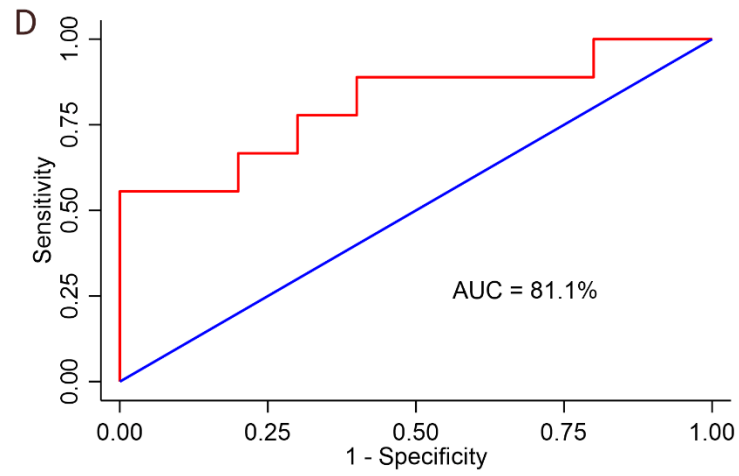


Figure 4.10. ROC curve of Lateralis gastrocnemius during walking. (from Di Giminiani et al., 2024)

Study 3

Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. (La Greca et al., 2025b)

Participant

These results refer to the same participant included in the case report, study 3 (La Greca et al., 2025b).

Results

The normalized sEMG activity did not exhibit any significant linear changes over time at any of the velocities ($p > 0.05$). But the weakest limb muscles at the final assessment (T4) exhibited an increase in $\Delta\%$, compared to the beginning (T0), at all velocities for BF ($\Delta\%$ average = 22%), TA ($\Delta\%$ average = 40%), and LG ($\Delta\%$ average = 46%).

The results indicate an improvement over time in the weakest leg, through increased EMG activity in the thigh and shank muscles.

4.5 Synthesis and applied clinical and functional implications

The findings highlight the importance of gait assessment approaches that explicitly address coordination and neuromuscular organization. Walking is not merely the result of isolated joint movements but emerges from the functional integration of multiple joints and muscles acting across time (Enoka, 2015; Winter, 1995). Therefore, clinically relevant alterations may remain undetected when assessment is limited to spatiotemporal variables or to single-joint angle–time profiles.

Across the three studies, angle–angle diagrams consistently revealed differences in locomotor organization, that in some case they were not captured by traditional kinematic descriptors (i.e. angle-time).

In persons with multiple sclerosis, the first study demonstrated that angle–angle diagrams, particularly at the knee–ankle level, were sensitive to reductions in conjoint range of motion and coordination when walking speed was controlled. While hip–knee representations did not reveal significant group differences, the knee–ankle diagrams showed a reduced loop area and a tendency toward altered perimeter, indicating constrained distal joint behaviour. These findings are consistent with previous qualitative observations of altered intralimb coordination in MS (Filli et al., 2018) but extend the literature by providing a quantitative characterization of these changes. Importantly, the associated EMG data revealed increased activation of proximal muscles (vastus lateralis and biceps femoris) and reduced activation of distal muscles (tibialis anterior and gastrocnemius). These altered muscle activation could be explained by a different recruitment pattern of motor units in proximal vs. distal muscles (Di Sapio et al., 2014) and represents a compensatory strategy that could increase when the balance is impaired in PwMS (Boudarham et al., 2016).

Such neuromuscular patterns align with models of compensatory control in neurological populations, where increased proximal activation is used to stabilize gait in the presence of distal deficits or balance impairments (Enoka, 2015; Winter et al., 1996).

In individuals with autism spectrum disorder, the angle-angle diagram has proven to be highly sensitive and specific in highlighting gait abnormalities. Despite the absence of significant differences in joint angle–time profiles between ASD and typically developed participants, knee–ankle angle–angle diagrams revealed a reduced loop area in the ASD group, indicating a lower conjoint angular excursion. This result suggests that traditional kinematic analyses may overlook subtle but functionally meaningful alterations in coordination.

The high sensitivity and specificity observed in the ROC analysis reinforce the discriminatory potential of this approach. Although the results are very promising, our kinematic gait analysis is not comparable with other published studies (Armitano et al., 2020; Bennett et al., 2021; Calhoun et al., 2011; Chester and Calhoun, 2012; Pauk et al., 2017; Vernazza-Martin et al., 2005; Weiss et al., 2013) due to marked differences in experimental designs, participants, and descriptive gait measures. Particularly in most studies in the literature, the age of participants ranged from 6 to 14 years, gait was often analysed using spatial and temporal measures (Armitano et al., 2020; Weiss et al., 2013), and no study analysed gait using angle-angle diagrams synchronized with sEMG in people with ASD.

From a neuromuscular perspective, a reduced activation of the lateral gastrocnemius during stance was observed in the ASD group, providing a physiological explanation for the altered ankle-related coordination detected by the angle-angle diagrams.

These findings are coherent with previous reports of altered ankle strategies and reduced distal control in ASD (Hallett et al., 1993).

Together, kinematic and EMG results suggest that locomotor alterations in ASD may reflect differences in motor organization and muscle recruitment rather than overt joint-level restrictions.

The longitudinal case study provides further insight into the applied value of integrating angle-angle diagrams and EMG. Over five years of individualized intervention, improvements in walking capacity were accompanied by progressive changes in joint coordination and muscle activation. Angle-angle diagrams revealed increased visual overlap and symmetry index between limbs and speed-dependent adaptations in loop area, perimeter, and shape, (La Greca et al., 2025b) indicating improved coordination rather than uniform increases in joint range of motion. These kinematic adaptations were mirrored by changes in EMG activity, particularly increased activation of distal muscles in the previously weakest limb. Such findings seem to suggest excellent sensitivity for the angle-angle diagram, which can detect the functional improvements induced by the intervention.

Across all three studies, a consistent pattern emerges: angle-angle diagrams are effective in identifying alterations in inter-joint coordination that are not detectable through conventional kinematic metrics, while synchronized EMG provides the necessary physiological context to interpret these alterations. This integrated framework allows differentiation between constrained movement due to limited joint excursion and altered

coordination driven by neuromuscular strategies, such as increased co-contraction or proximal compensation (Latash and Singh, 2024).

From an applied clinical perspective, these findings suggest that gait assessment should move beyond purely descriptive metrics toward methods capable of capturing coordination and control. Angle–angle diagrams offer a concise and interpretable representation of multijoint behaviour, making them suitable for monitoring disease progression, evaluating intervention efficacy, and identifying individual-specific locomotor strategies. When combined with EMG, they provide insight into the neuromuscular mechanisms underlying observed gait patterns, improving the functional relevance of assessment.

Beyond clinical populations, the implications of these findings extend to the field of sport and exercise science. Efficient locomotion in athletic performance relies on optimal inter-joint coordination and adaptable muscle activation patterns (Enoka, 2015). Alterations in angle–angle relationships may reflect suboptimal movement strategies, increased mechanical stress, or compensatory patterns associated with fatigue or injury risk (Hershler and Milner, 1980b, 1980a). Thus, the integrated use of angle–angle diagrams and EMG may support performance monitoring, technique optimization, and injury prevention strategies, complementing traditional biomechanical analyses.

In summary, the results of this chapter support the use of integrated kinematic and electromyographic analyses to advance the functional interpretation of gait. Angle–angle diagrams, particularly when synchronized with EMG, provide a robust framework to assess joint coordination, neuromuscular strategies, and adaptive capacity during locomotion. This approach holds relevance not only for advanced clinical assessment but also for applied sport and exercise contexts, where understanding movement quality and coordination is essential for optimizing performance and preventing dysfunction.

CHAPTER 5 – Force–Time Profile Assessment

5.1 The force–time profile as a neuromechanical indicator

The force–time profile represents the temporal evolution of force production during the execution of a motor task. From a biomechanical standpoint, it describes the external manifestation of internal neuromuscular processes acting within the constraints imposed by task mechanics and body geometry (Winter, 2009; Zatsiorsky, 2002). Unlike single-point force measures, the force–time profile captures how force is generated, modulated, and released over time, thereby providing a dynamic representation of motor output (Enoka, 2015).

Force–time analysis can be applied to both static tasks and dynamic movements, which differ in the mechanical conditions under which force is produced. In static tasks, such as isometric contractions, joint configuration remains constant, and the measured force reflects the ability of the neuromuscular system to generate force without segmental motion (Enoka, 2015). Under these conditions, force output is primarily determined by neural drive and muscle–tendon force-generating capacity (Bosco, 1997; Latash and Singh, 2024). In contrast, during dynamic tasks, force production occurs concurrently with joint motion and segment acceleration, and the recorded force reflects the combined effects of muscle action, inertial properties, and external loads (Bosco, 2002; Winter, 2009; Zatsiorsky, 2002). Despite these differences, the force–time profile provides a unifying framework to investigate force regulation across motor task.

Experimentally, force–time profiles are obtained using force sensors or force platforms, which allow high-resolution recording of externally applied forces or ground reaction forces. In biomechanics context, the force platforms are widely considered the reference in force assessment. These sensors directly quantify the net interaction between the body and the environment and offer precise temporal resolution (Medved, 2021; Winter, 2009). This temporal accuracy is critical for identifying key mechanical events within a task, such as force onset, peak force, and force decline.

Several functionally parameters can be extracted from the force–time signal (Payton et al., 2018). Peak force reflects the maximal mechanical output achieved under specific task constraints and is commonly used as an indicator of maximal strength capacity (Bosco,

1997; Winter, 2009; Zatsiorsky, 2002). Impulse is defined as the time integral of force and represents the cumulative mechanical effect of force application. For this reason, it is particularly informative in tasks involving changes in momentum, such as push-off or landing phases. (Bosco, 2002; Medved, 2021; Zatsiorsky, 2002). Rate of force development (RFD) describes how rapidly force is produced and is strongly influenced by neural factors, including motor unit recruitment strategies and discharge rates, especially during the early phase of contraction (Bosco and Viru, 1995; Enoka, 2015; Payton et al., 2018; Winter, 2009).

Beyond isolated parameters, the overall shape of the force–time curve provides insight into motor strategy and neuromuscular organization. Variations in curve slope, timing of peak force, or force decay may reflect alternative solutions adopted by the motor system to satisfy task demands. From a neuromechanical perspective, these features emerge from the interaction between neural control, muscle–tendon mechanical properties, and task-specific constraints, rather than from muscle strength alone (Bosco, 1997; Enoka, 2015; Zatsiorsky, 2002).

Importantly, modifications in force–time characteristics may occur in the absence of changes in peak force. Such alterations can reflect adaptations induced by training, fatigue, injury, or neurological conditions, indicating changes in force regulation rather than maximal capacity (Bosco, 2002; Medved, 2021). This sensitivity makes force–time profiling particularly suitable for detecting subtle changes in neuromuscular function that may not be evident through traditional strength measures (only peak force).

In summary, the force–time profile constitutes a robust neuromechanical indicator that links mechanical output to the underlying organization of the neuromuscular system. By integrating information on force magnitude and temporal structure, it provides a coherent framework for the analysis of force production in both static and dynamic tasks.

5.2 Force–time profiles in static tasks

Static tasks represent a controlled experimental condition in which joint motion is absent or minimal, and the external force is generated against a resistance (Enoka, 2015). From a biomechanical perspective, this configuration reduces the influence of the inertial effects, allowing the force signal to reflect the output of the neuromuscular system (Winter, 2009; Zatsiorsky, 2002). For this reason, static force assessments are widely used as a first level assessment tool to investigate force production capacity and neuromuscular

control.

In clinical and rehabilitative contexts, static tasks offer an additional and relevant advantage. The absence of joint displacement limits the generation of potentially harmful joint torques and shear forces (Enoka, 2015), making isometric testing a safer option for individuals recovering from injury or surgery (i.e. anterior cruciate ligament reconstruction) (Fleming et al., 2005; Knezevic et al., 2012). Under these conditions, force production can be monitored without exposing the musculoskeletal system to excessive mechanical stress, while still providing meaningful information on neuromuscular function (Payton et al., 2018).

Within static conditions, the force–time signal becomes more sensitive to neural drive and motor unit recruitment strategies, as less confounded by movement-related factors. This characteristic makes force–time profiles particularly suitable for isolating the capacity to generate force, and to reduce the interpretative complexity that typically characterizes dynamic tasks (Winter, 2009; Zatsiorsky, 2002).

From a functional perspective, static assessments provide a valuable framework for examining muscle actions during isometric tasks under different conditions. In this context, we present two studies in which distinct static tasks were used to obtain specific functional information (Enoka, 2015).

The first study focused on maximal isometric contractions performed within an open kinetic chain paradigm (i.e., leg extension). In this task, participants were asked to exert force against fixed resistance. Such protocols are widely adopted to characterize neuromuscular function, explore asymmetries between limbs, and examine differences between populations or experimental conditions, including pre- and post-intervention comparisons.

The second study examined isometric tasks involving simpler and more controlled tasks, such as handgrip strength production. Despite their relative mechanical simplicity, these tasks are considered functionally significant, as they are closely related to activities of daily living and as an index of neurocognitive function (American College of Sports Medicine et al., 2026; Bohannon, 2019; McGrath et al., 2022). Rather than emphasizing maximum production, these assessments primarily capture aspects of force modulation. Importantly, even in the absence of movement, force production is not constant over time. The force–time profile obtained during static tasks typically exhibits a rising phase, a plateau or quasi-steady phase, and temporal fluctuations around the target force level. These features reflect underlying neuromuscular processes, including the rate of motor

unit recruitment, firing frequency modulation, and feedback-driven force adjustments (Enoka, 2015; Farina et al., 2014).

Several parameters can be extracted from static force–time profiles at a conceptual level. Peak or maximal force represents the upper limit of force-generating capacity. The rate of force development describes the ability to increase force rapidly and is considered highly sensitive to neural factors, particularly during the early phase of contraction (Aagaard et al., 2010, 2002).

From a neuromechanical perspective, static tasks minimize the contribution of stretch–shortening mechanisms, and inertial components. As a result, the measured force reflects predominantly the neural drive to the muscle. This characteristic makes static force–time profiles especially suitable for integration with surface electromyography, allowing a more direct interpretation of muscle activation patterns (Enoka, 2015; Farina et al., 2014). Despite static tasks do not replicate the complexity of dynamic or sport-specific movements, they offer high diagnostic sensitivity and reproducibility. In both clinical monitoring and athletic evaluation, static force–time analysis represents a foundational level of assessment (Payton et al., 2018; Winter, 2009).

5.2.1 Maximal isometric contractions

Maximal voluntary isometric contractions (MVICs) are widely used to assess neuromuscular function in both healthy and clinical populations. By constraining joint motion and eliminating external work, isometric tasks allow force production to be quantified under controlled mechanical conditions, limiting the influence of inertial and kinematic factors (Knezevic et al., 2012). As a result, MVICs are commonly adopted to evaluate maximal force capacity, inter-limb differences, and neuromuscular deficits across training, rehabilitation, and return-to-activity contexts.

Importantly, MVICs provide information beyond peak force. During an isometric contraction, force develops progressively over time because of motor unit recruitment and firing rate modulation (Wilson and Murphy, 1996). However, many functional actions occur within short time frames that preclude the attainment of maximal force. The rate of force development (RFD), derived from the early phase of the force–time curve, captures the capacity to generate force rapidly and is highly sensitive to neural factors, making it a relevant complement to maximal force assessment (Aagaard et al., 2002; Wojtys and Huston, 1994).

When synchronized with surface sEMG, MVIC testing allows a more comprehensive

characterization of neuromuscular function. The combined analysis of force–time and EMG signals provides insight into neural drive, agonist–antagonist coordination, and co-activation strategies, which cannot be inferred from force data alone (Solomonow et al., 1987). This integrated approach is particularly valuable in contexts where neuromuscular control plays a central role in functional performance.

In athletes undergoing anterior cruciate ligament reconstruction, these considerations become especially relevant. Following ACL injury and reconstruction, functional assessment plays a crucial role in guiding rehabilitation and supporting a safe return to sport, given the substantial proportion of athletes who do not regain pre-injury performance levels and the persistently high risk of re-injury (Arderm et al., 2011; Wiggins et al., 2016). Strength evaluation of the knee extensors and flexors is therefore a central component of post-ACLR assessment and is commonly performed under isokinetic or isometric conditions, often in combination with functional test batteries (Knezevic et al., 2012).

Isometric testing offers specific advantages in this population. When performed without joint displacement, it allows force production to be quantified while limiting anterior shear forces and excessive joint loading, making it suitable even in advanced stages of rehabilitation (Fleming et al., 2005; Knezevic et al., 2012). Moreover, lower-limb isometric strength normalized to body mass has been shown to be more predictive of self-reported knee function than traditional limb symmetry indices (Pamukoff et al., 2017).

Traditionally, strength assessments after ACLR are performed unilaterally, comparing the operated limb with the non-operated limb. This approach implicitly assumes that the contralateral limb represents a valid reference. However, emerging evidence indicates that ACL injury and reconstruction are associated with bilateral neural alterations, including changes in corticospinal excitability and motor control that affect both limbs (Lepley et al., 2019). These findings support the concept that ACLR should be considered not only a peripheral musculoskeletal injury but also a condition involving central neurophysiological reorganization (Kapreli et al., 2009).

Neuroimaging and neurophysiological studies further suggest that bilateral lower-limb tasks elicit greater activation of cortical, cerebellar, and subcortical structures compared with unilateral tasks, even when unilateral tasks require higher force output (Grooms et al., 2017; Noble et al., 2014). As bilateral motor control appears to be a fundamental feature of lower-extremity function, the exclusive use of unilateral assessments may overlook clinically relevant neuromuscular asymmetries and compensatory strategies

(Grooms et al., 2017; Mirkov et al., 2017).

Based on these considerations, the assessment of maximal isometric knee extension performed under both unilateral and bilateral conditions, combined with synchronized surface electromyography, may provide a more comprehensive characterization of neuromuscular function following ACLR. This approach allows the investigation of force production, rapid force development, and muscle co-activation within a framework that better reflects the bilateral organization of lower-limb motor control.

Study

Neuromuscular Characteristics of Unilateral and Bilateral Maximal Voluntary Isometric Contractions following ACL Reconstruction. (Di Giminiani et al., 2023)

Methods

Participants

Twenty male athletes competing at regional–national level participated in this cross-sectional study with repeated measures (Table 5.1). The ACLR group consisted of athletes who had undergone unilateral anterior cruciate ligament reconstruction and completed a standardized rehabilitation program, achieving full return to sport 6–9 months post-surgery. Rehabilitation was conducted in a national health system–affiliated center following established clinical guidelines (Beynnon et al., 2005). Control participants reported no history of ligament or meniscal injuries.

Table 5.1. Participants characteristics. (From: Di Giminiani et al., 2023)

Variables	Group	
	ACL-R Group (n = 10)	Health Control Group (n = 10)
Age (years)	23.4±2.1	24.0±3.4
Stature (cm)	182.±9.0	180.3±10.7
Body Mass (Kg)	78.6±9.9	74.9±13.
BMI (Kg/m ²)	27.7±1,9	22.8±2.7
Sport level	Competitive	Competitive
Leg Dominance	Right (n=9) / Left (n=1)	Right (n=9) / Left (n=1)
Operated Leg	Right (n=6) / Left (n=4)	N/A
Graft type	SGT (n=13)	N/A
Event Distribution	No-Contact Mechanism	N/A
Post-Operative Period	6 Months-2 Years	N/A

Abbreviations: body mass index (BMI), semitendinosus–gracilis tendon (SGT).

Exclusion criteria for both groups included musculoskeletal or neurological disorders, neuromuscular pathologies, intervertebral disc herniation, cardiac arrhythmias, epilepsy, or other relevant comorbidities.

Participants attended two laboratory sessions. The first session served for familiarization with unilateral and bilateral MVICs. The second session, performed at least 48 h later (Courel-Ibáñez et al., 2020), included the experimental protocol with synchronized acquisition of force–time and surface sEMG signals during MVICs.

Setup for MVC assessment

During the testing session, participants completed a standardized 20-min warm-up consisting of a general phase (treadmill running at $7 \text{ km} \cdot \text{h}^{-1}$ and static stretching) followed by a specific phase including eccentric–concentric and isometric squat exercises.

MVICs were performed on a leg-extension device instrumented with a strain gauge (Muscle-Lab, Bosco System, Ergotest Innovation, Norway) under unilateral (operated-limb, OL; and non-operated limb, N-OL) and bilateral conditions. The knee joint was fixed at 120° , monitored via electrogoniometry (Muscle-Lab, Bosco-System, Ergotest Innovation, Stathelle, Norway) and participants were seated with standardized alignment and stabilization to minimize extraneous movements.

For each condition, participants performed up to three MVIC trials separated by 1-min rest intervals, and the highest force value was retained for analysis. Participants were instructed to produce force as fast and as hard as possible, with standardized verbal encouragement. Contractions were sustained until a force plateau was reached, and force–time signals were monitored in real time (Muscle-Lab, Bosco-System, Ergotest Innovation, Stathelle, Norway).

The sEMG was recorded synchronously from the vastus lateralis, rectus femoris, vastus medialis, and biceps femoris bilaterally. Skin preparation followed standard procedures, and electrode (T3402 M, nickel-plated brass, electrode diameter = 1 cm, interelectrode distance = 2 cm; Thought Technology Ltd., Montreal, QC, Canada) placement adhered to SENIAM guidelines (Hermens et al., 2000). The EMG electrodes and cables were secured with elastic bands (Pic Elasticflex, 3M Italia, Pioltello, Italy) to prevent motion artefacts. sEMG signals were processed as root mean square (RMS) values and time-aligned with force–time data using the Muscle-Lab system. The sEMG detection technique entailed full-wave true RMS conversion of the signal from the preamplifier with a sampling frequency of 100 Hz (sampling of the converted signal). The size of the averaging window was 100 ms (averaging is analogous) with a resolution of a 16 bit A/D converter. The EMG preamplifier characteristics were as follows: voltage supply, $\pm 5 \text{ VDC}$; input impedance, $2 \text{ G}\Omega$; common mode rejection rate, 100 dB; gain at 100 Hz, 500; 3 dB low-cut frequency, 8 Hz; and 3 dB high-cut frequency, 1.2 kHz.

MVICs were identified from the force–time signal using Muscle-Lab software. A 400-ms window centered on the peak force ($\pm 200 \text{ ms}$) was used to compute the sEMG RMS of the knee extensor (vastus lateralis, VL; vastus medialis, VM; rectus femoris, RF) and flexor (biceps femoris, BF) muscles during unilateral and bilateral executions (figure 5.1).

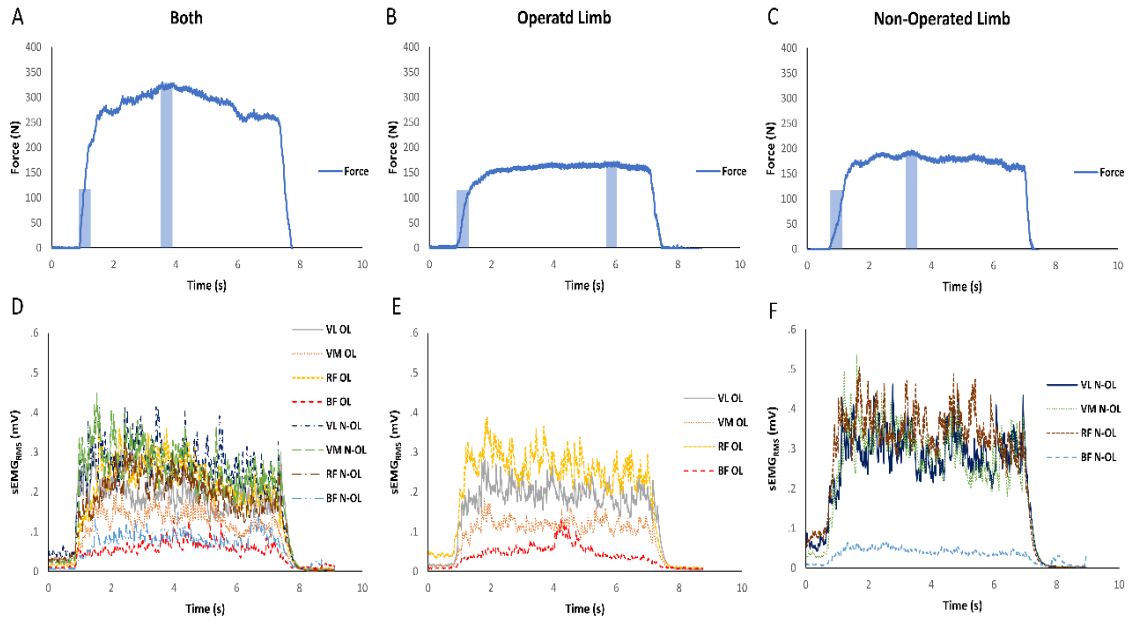


Figure 5.1. Representative example of force-time and surface electromyography root-mean-square (sEMGRMS)-time signals recorded during unilateral (operated leg, OL and non-operated leg, N-OL) and bilateral maximal voluntary isometric contractions (MVICs) during the leg extension. The sEMGRMS activity was recorded in the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), and biceps femoris (BF) muscles. The larger shaded areas represent the window (0.4 s) used to compute the synchronised muscles sEMGRMS around the peak of MVICs. The rate of force development (RFD) and the synchronised sEMGRMS were computed 0.20 s after the force exertion (the smallest shaded areas). (From: Di Giminiani et al., 2023)

The RFD was calculated from the initial portion of the force–time curve during MVICs, using a 200-ms window from force onset (Aagaard et al., 2002). Within this interval, sEMGRMS of the involved muscles were computed (Figure 5.1). Quadriceps muscle force (F_m) was estimated using static torque equilibrium at the knee joint (Finni et al., 2000):

$$\begin{aligned} \sum \tau_K &= 0 \quad \Rightarrow \quad \tau_m + \tau_e = 0 \\ (b_m \times F_m) + (b_e \times F_e) &= 0 \quad \Rightarrow \quad F_m = -\frac{b_e \times F_e}{b_m} [N] \end{aligned}$$

At the knee joint (point K), two torques act: the muscle torque (τ_m) generated by the quadriceps muscle force (F_m), and the external torque (τ_e), given by the external lever arm (b_e) multiplied the measured external force (F_e). External lever arms (b_e) were measured using video-based markers on the lateral femoral condyle and lateral malleolus

and analysed with Kinovea software. The quadriceps moment arm (b_m) was calculated according to Hosseinzadeh et al. (Hosseinzadeh et al., 2020) (figure 5.2).

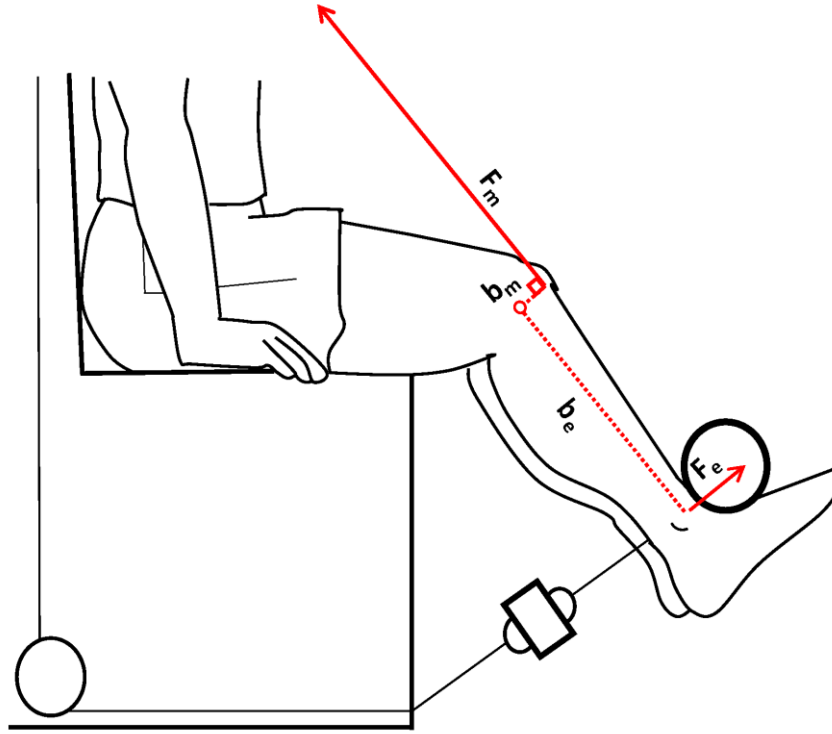


Figure 5.2. Schematic representation of quadriceps muscle force calculation during isometric leg extension. The external force (F_e) is measured by a strain gauge. The muscle force (F_m) is calculated. The external lever arm (b_e) and the muscle moment arm (b_m) are also indicated in the picture. (From: Di Giminiani et al., 2023)

The co-activation index (CI) between knee extensors and flexors was calculated as follows (Di Giminiani et al., 2020):

$$CI = \frac{\left[\sum_{i=1}^n \frac{\text{lower EMG}_i}{\text{higher EMG}_i} (\text{lower EMG}_i + \text{higher EMG}_i) \right]}{n}$$

where i is the sample number and n is the total number of samples. Lower EMG_i and higher EMG_i correspond to the minimum and maximum EMG values within the interval, respectively. This approach quantifies co-activation without distinguishing agonists from antagonists' roles during leg extension.

Statistics

The data were not normally distributed, as revealed by the Shapiro–Wilk’s W test and a non-parametric statistical procedure was used. Differences in several variables between the operated and non-operated legs were tested using the Wilcoxon signed-rank test/two-tailed test. The analyses were performed using XLSTAT (Statistical and Data Analysis Solution, Addinsoft, New York, NY, USA 2022). Statistical significance was set at $p \leq 0.05$, and the meaningfulness of significant outcomes was estimated by calculating the ES of Cohen (Cohen, 2013).

Results

The muscle force exerted during MVIC between the two legs was not significant within the ACL (OL vs. N-OL) and CG (Right vs. Left) ($p > 0.05$). RFD did not show any significant differences between unilateral and bilateral exercises within each group ($p > 0.05$).

The VL sEMG_{RMS} activity during MVIC was significantly higher in the OL than that in the N-OL when execution was performed bilaterally ($p < 0.05$; ES = 1.35),

During bilateral execution, the BF sEMG_{RMS} activity related to RFD also showed a higher activation in the OL than that in the N-OL ($p < 0.05$; ES = 0.49) (Figure 5.3B). On the contrary, the comparison between the right and the left leg was not significant in the CG ($p > 0.05$).

The CI indexes of the VL-BF, RF-BF, and VM-BF muscles in the OL were higher than that of the N-OL when the exertion was performed bilaterally ($p < 0.05$, ES= 0.33 – 0.58) and not unilaterally ($p > 0.05$) (Figure 5.4).

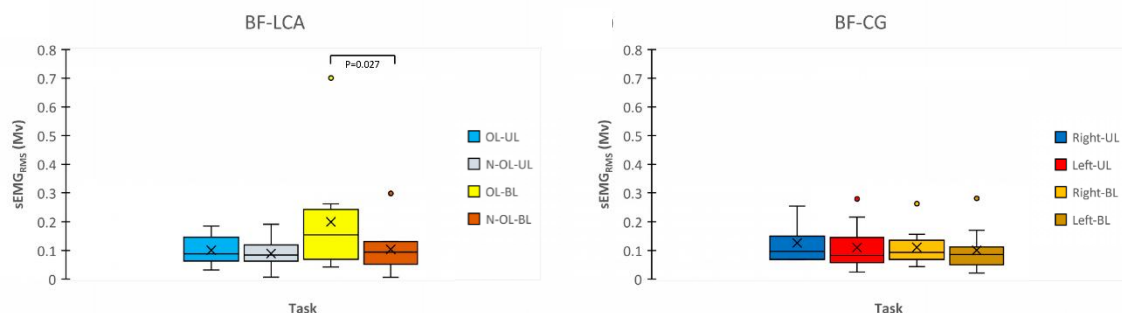


Figure 5.3. Surface electromyography root-mean-square (sEMG_{RMS}) of biceps femoris (BF) relative to the rate of force development (RFD) is reported with box plots for ACL and CG. OL, Operated Leg; N-OL, Non-Operated; UL, Unilateral; BL, Bilateral. (From: Di Giminiani et al., 2023)

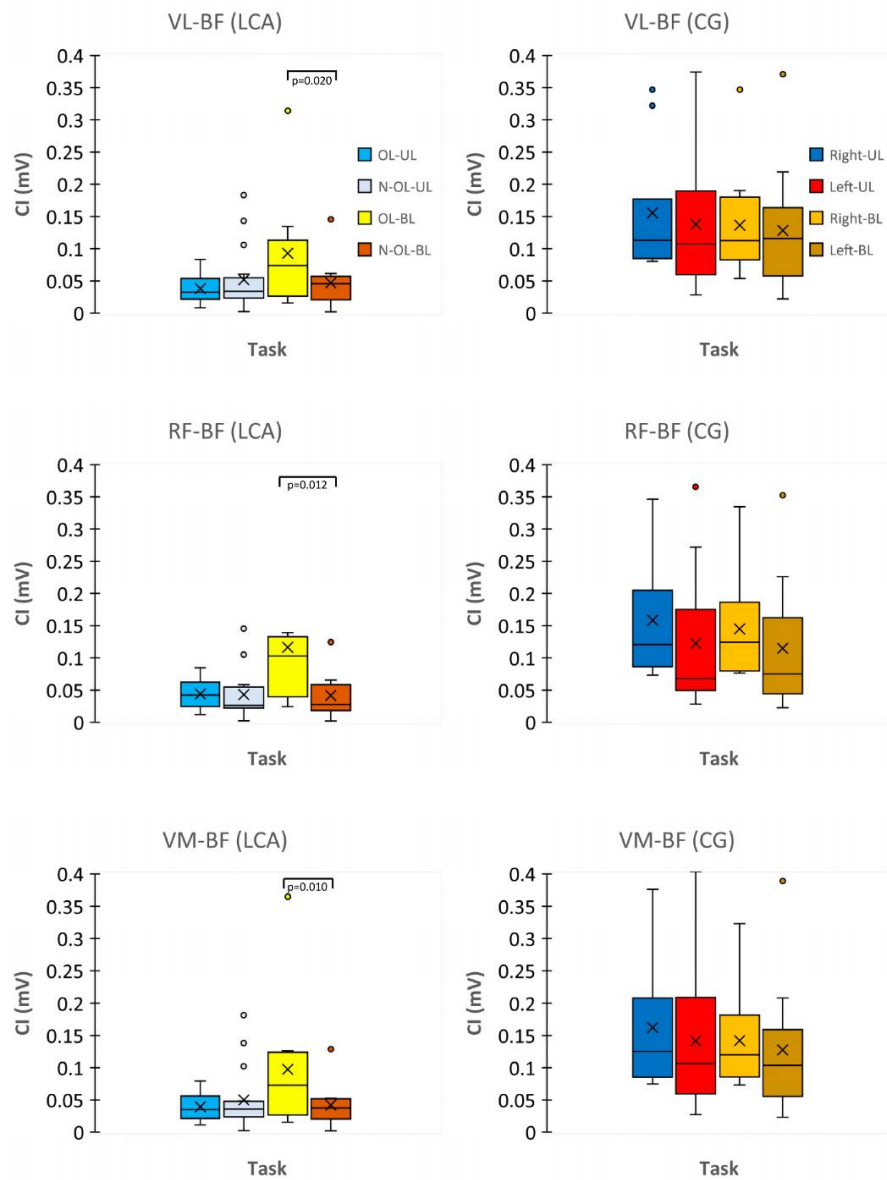


Figure 5.4. Coactivation Index (CI; $sEMG_{RMS}$) of leg muscles (vastus lateralis-biceps femoris, VL-BF; rectus femoris-biceps femoris, RF-BF; vastus medialis-biceps femoris, VM-BF) relative to the rate of force development (RFD) is reported with box plots in ACL and CG. OL, Operated Leg; N-OL, Non-Operated Leg; UL, Unilateral; BL, Bilateral. (From: Di Giminiani et al., 2023)

Although maximal force and RFD did not differ between limbs, bilateral execution revealed limb-specific differences in muscle activation and co-activation patterns, highlighting neuromuscular asymmetries that were not detectable during unilateral tasks.

5.2.2 Hand-grip strength: force–time profiles

Handgrip strength (HGS) is widely used as a reliable indicator of neuromuscular function in both healthy and clinical populations. Owing to its strong association with global muscle strength, functional capacity, and health outcomes, HGS has been proposed as a comprehensive marker of physical impairment and even as a “vital sign” of health status (McGrath et al., 2022; Vaishya et al., 2024). Standardized assessment procedures, such as those recommended by the American College of Sports Medicine (American College of Sports Medicine et al., 2026), typically involve maximal voluntary squeezing efforts lasting 3–5 s performed with a handheld dynamometer, with peak force recorded as the primary outcome.

Beyond peripheral strength, HGS reflects broader aspects of neuromuscular integrity and central motor function. Grip strength has been associated with overall brain health and central nervous system involvement, supporting its use as an index of global neuromuscular status rather than an isolated upper-limb measure (Bohannon, 2019). In clinical populations, reductions in HGS are linked to decreased independence in activities of daily living, such as eating, drinking, and dressing, highlighting its functional relevance (Iyengar et al., 2009). In individuals with multiple sclerosis, low grip strength has been proposed as an indicator of widespread neuromuscular impairment, capturing alterations in motor control that extend beyond localized muscular deficits (MacKenzie et al., 2024). Despite its broad clinical relevance, traditional HGS assessment is generally limited to the measurement of peak force, providing a single scalar outcome. From a biomechanical perspective, this approach neglects the temporal organization of force production and fails to capture how force is generated over time. However, functional performance often depends not only on maximal force capacity but also on the ability to rapidly develop force, a characteristic strongly influenced by neural drive and motor unit recruitment dynamics.

The analysis of the force–time profile during isometric handgrip contractions allows the extraction of additional parameters, such as the RFD, offering a more refined characterization of neuromuscular function. RFD reflects the early phase of force generation and is particularly sensitive to neural impairments, making it a meaningful complement to peak force assessment (Aagaard et al., 2002). When synchronized with sEMG, force–time analysis further enables the investigation of muscle activation strategies underlying grip force production, providing insights that cannot be inferred from maximal force alone.

Within this framework, the following study adopted an instrumented handgrip device equipped with a force sensor to quantify not only maximal grip force but also force–time–derived parameters and sEMG activity. Although the intervention was not directly applied to the upper limbs, handgrip assessment was included as a measure of global neuromuscular function, given its established association with overall physical impairment and central motor integrity (Bohannon, 2019; McGrath et al., 2022). The inclusion of an instrumented handgrip assessment therefore represents a methodological advancement over traditional peak-force-only evaluations, allowing a more sensitive and comprehensive monitoring of neuromuscular adaptations over time.

Study

Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. (La Greca et al., 2025b)

Method

Participant

This case report investigated the long-term (5-year) effects of individualized WBV training on locomotion in a PwMS of a 68-year-old man with MS (body mass 70 kg, height 1.8 m, and BMI: 21.6 kg/m²). The intervention was assessed during 4 follow-ups (T1, T2, T3 and T4).

Setup for Handgrip Strength Test and Synchronized sEMG Activity

Handgrip strength was measured using a customized dynamometer with a strain gauge (Muscle-Lab, Bosco-System 4000e, Ergotest Innovation, Norway). Participants were seated with a straight back, feet flat, shoulder abducted, elbow at 90°, forearm on the table, and wrist neutral. The dynamometer was adjusted to the hand and positioned consistently across trials. Participants performed three maximal isometric contractions of 3–4 s each, applying force smoothly; the highest trial was retained for analysis.

Muscle activity of the anterior deltoid (DA), biceps brachii (BB), and palmaris longus (PL) were recorded using sEMG. Electrodes were placed according to standard guidelines, with skin prepared to maintain impedance <5 k Ω and cables secured to prevent artifacts (Hermens et al., 2000). Triode electrodes (T3402 M, nickel-plated brass, 1 cm diameter, 2 cm interelectrode distance; Thought Technology Ltd.) Signals were recorded as full-wave RMS with a 100 ms averaging window and 16-bit resolution; preamplifier: ± 5 VDC, 2 G Ω input impedance, 100 dB common mode rejection, gain 500 at 100 Hz, 8 Hz low-cut, 1.2 kHz high-cut (Muscle-Lab 4000e, Bosco-System).

Force–time data were collected synchronously with sEMG. sEMG_{RMS} was calculated over a 400 ms window around the peak (± 200 ms), and the rate of force development (RFD) was computed over the first 200 ms. Inter-hand differences were expressed as the ratio of stronger to weaker hand, with asymmetries classified as symmetric (<15%, <1.15), mild (15–25%, 1.15–1.25), or moderate (>25%, >1.25) (McGrath et al., 2022).

Statistics

The Statistics Toolbox implemented in the MATLAB software interface (version 2024b; MathWorks, Natick, MA, USA), were used for statistical analysis.

To evaluate the long-term effects of the intervention for each variable the following formula was used:

$$\Delta\% = \left[\frac{\text{follow up} - T0}{T0} \right] * 100$$

In addition, a linear regression model was constructed across the follow-ups, this model examined the trend of the dependent variables over the time. The significance level (α) was set to 0.05.

Results

There were no significant linear changes in the mechanical variables (peak force and RFD) over time ($p > 0.05$). However, the RFD of the weakest limb appeared to exhibit a trend in the direction of significance ($p = 0.079$). The peak force symmetry ratio between the weakest and strongest hand demonstrated a substantial increase over time ($p = 0.024$), achieving the symmetry values at T4 (1.09). The BB muscle of the weakest limb exhibits a substantial increase in sEMG activity over time ($p = 0.046$). Other muscles did not change significantly over time ($p > 0.05$) (table 5.2).

Table 5.2. Handgrip strength test values and synchronized sEMG_{RMS} activity are reported on different test occasions. The $\Delta\%$ differences and the symmetry ratio between W-S peak of force are also shown. (From: La Greca et al., 2025b)

Variables	T1	T2	T3	T4	P	T1-T2 $\Delta\%$	T1-T3 $\Delta\%$	T1-T4 $\Delta\%$
W hand F peak (N)	143.5	134.8	168.8	202.05	0.159	-6	18	41
S hand F peak(N)	179.3	165.9	195.3	220.32	0.103	-7	9	23
peak F W/S (S-ratio)	1.25	1.23	1.16	1.09	0.024*	/	/	/
W hand RFD (N/s)	278.0	424.5	417.4	696.82	0.079	53	50	151
S hand RFD (N/s)	503.7	427.8	628.7	1013.5	0.142	-15	25	101
W DA EMG _{RMS} (mV)	0.151	0.163	0.195	0.183	0.163	8	29	21
W BB EMG _{RMS} (mV)	0.084	0.110	0.203	0.214	0.046*	31	142	155
W PL EMG _{RMS} (mV)	0.064	0.057	0.067	0.069	0.385	-11	5	8
S DA EMG _{RMS} (mV)	0.070	0.116	0.094	0.080	0.948	66	34	14
S BB EMG _{RMS} (mV)	0.142	0.052	0.130	0.186	0.514	-63	-8	31
S PL EMG _{RMS} (mV)	0.142	0.236	0.245	0.145	0.959	66	73	2

Abbreviations: Symmetry ratio (S-ratio); weakest (W); strongest (S); force (F); rate of force development (RFD); deltoid anterior (DA); biceps brachii (BB); palmaris longus (PL).

Although peak strength and RFD did not show significant linear changes over time, the results indicate a progressive improvement in strength symmetry between the limbs, accompanied by a neuromuscular activation improvement in the weaker limb over the time.

5.3 Force–time profiles in dynamic tasks

While static tasks allow force production to be examined under mechanically constrained conditions, dynamic tasks represent a more complex and functionally representative situation in which force is generated while body segments are in motion (Enoka, 2015). In these conditions, joint angles vary continuously, segments undergo acceleration and deceleration, and the externally measured force reflects the combined effects of neuromuscular activation, inertial properties, gravitational forces, and interaction with the environment (Winter, 2009; Zatsiorsky, 2002). Consequently, force–time profiles obtained during dynamic tasks convey information not only on force magnitude, but also on the temporal organization of force relative to movement execution.

From a biomechanical perspective, force–time profiles in dynamic tasks differ fundamentally from those observed under static conditions. The recorded signal does not represent muscle force per se, but the net external force resulting from the interaction between muscle action and task mechanics within a moving system (Enoka, 2015). As a result, the interpretation of dynamic force–time data requires explicit consideration of movement phases, joint kinematics, and task-specific constraints, as force production cannot be dissociated from the motion it generates or controls (Zatsiorsky, 2002).

A defining characteristic of dynamic force–time profiles is their phase-dependent structure. Within a single motor task, distinct mechanical phases can be identified, such as braking and propulsion, eccentric and concentric actions, or landing and take-off phases, particularly in plyometric movements (Bosco, 2002). Each phase is associated with specific force–time features, including the rate of force increase, peak force timing and magnitude, and force attenuation patterns. These features reflect how the neuromuscular system regulates force to manage external loads, control body motion, and maintain mechanical stability throughout the task (Bosco, 2002; Winter, 2009).

Several biomechanically meaningful variables can be extracted from force–time signals during dynamic tasks (Payton et al., 2018). Peak force describes the maximal external load encountered or produced, whereas impulse represents the cumulative effect of force application and is central to movements involving changes in momentum. Temporal variables, such as time to peak force, contact time, and flight time, provide insight into movement efficiency and coordination. In tasks dominated by the stretch–shortening cycle, indices derived from the relationship between force and time, such as the reactive strength index (RSI), are commonly used to characterize the ability to rapidly transition from eccentric to concentric force production (Bosco, 2002; Jarvis et al., 2022; Zatsiorsky,

2002).

Beyond isolated metrics, the overall shape of the force–time curve reflects the motor strategy adopted to meet the mechanical demands of the task. Variations in curve morphology—such as differences in force rise slope, braking duration, or force distribution across phases—indicate alternative neuromechanical solutions that emerge from the interaction between neural control, muscle–tendon mechanical properties, and external constraints, rather than from maximal force capacity alone (Bosco, 2002; Enoka, 2015).

Importantly, dynamic force–time analysis is particularly sensitive to alterations in neuromuscular control that may not be evident in static assessments. Changes in timing, coordination, or force modulation can occur without corresponding changes in peak force, revealing adaptations related to training, fatigue, injury, or neurological conditions (Enoka, 2015; Winter, 2009). This sensitivity makes dynamic force–time profiling especially suitable for identifying qualitative modifications in movement execution, rather than solely quantifying maximal output.

From an applied standpoint, the selection and design of dynamic assessment tasks represent a critical methodological aspect. To effectively characterize biomechanical and neuromuscular behavior, dynamic tasks should be structured to elicit movement patterns that reflect those encountered in sport practice or daily activity. Task-specific assessments enhance the functional relevance of force–time analysis and improve its ability to capture meaningful biomechanical characteristics of the athlete or individual being evaluated (Payton et al., 2018). Within this context, dynamic force–time profiling provides a conceptual and methodological basis for the analysis of complex motor tasks such as jumping and landing, which are addressed in the following sections.

5.3.1 Drop vertical jump and landing strategies

Plyometric jumping tasks have long been used to evaluate and train explosive and reactive strength capacities in sport contexts (Montoro-Bombú et al., 2023). Among these, the DVJ represents a representative model for investigating neuromuscular behavior under high mechanical demand, due to the rapid transition from landing to propulsion and the prominent involvement of the stretch–shortening cycle (SSC) (Bosco et al., 1981, 1982a; Bosco and Komi, 1979). Consequently, DVJ tasks are widely adopted both in performance-oriented assessments and in applied settings related to injury prevention and movement screening (Cacolice et al., 2020; Cowling et al., 2003).

From a biomechanical perspective, the DVJ is characterized by a marked interaction between the body and the environment, resulting in rapid variations in vertical ground reaction forces (vGRF) over short time intervals (Bosco et al., 1982b). During ground contact, force modulation reflects the ability of the neuromuscular system to attenuate impact loads during the braking phase and to rapidly generate force during the subsequent propulsion phase. These responses are strongly influenced by drop height, which determines the magnitude of the stretch load imposed on the musculotendinous system and shapes the resulting neuromechanical behaviour (Bosco and Komi, 1979; Di Giminiani and Petricola, 2016; Komi and Bosco, 1978).

The force–time profile obtained during DVJ tasks provides a direct representation of these mechanisms. Distinct movement phases, such as impact, active braking, and propulsion, can be identified within the vGRF–time signal, each characterized by specific force magnitudes, rates of force development, and temporal features (Bosco and Komi, 1979; Komi and Bosco, 1978). Parameters derived from the force–time curve, including peak forces, impulse, contact time, and indices related to reactive strength, therefore offer insight into both performance outcomes and landing load management strategies (Montoro-Bombú et al., 2023).

Importantly, force regulation during DVJ tasks reflects not only mechanical constraints but also the integration of proprioceptive afferent input with central neural control processes (Nicol et al., 2006). At optimal stretch loads, neural potentiation may enhance force and power production during propulsion (Bosco et al., 1982b; Komi and Bosco, 1978; Nicol et al., 2006), whereas excessive loading can elicit inhibitory responses mediated by Golgi tendon organs, resulting in altered force–time characteristics aimed at preserving musculoskeletal integrity (Gollhofer et al., 1987). These adaptations are clearly expressed in the shape and magnitude of the force–time profile.

Within this context, vGRF–time analysis represents a sensitive approach for detecting changes in jump performance and landing strategies that may not be evident from outcome measures alone. Variations in force peaks, temporal force distribution, or phase-specific loading patterns can reveal subtle modifications in neuromuscular control, particularly relevant in athletes who frequently perform plyometric actions during sport practice (Henrique et al., 2023; Young, 2006).

In the following study, the DVJ was employed as a task-specific assessment to examine how force–time variables respond under different mechanical conditions. By analyzing selected parameters of the vGRF–time profile across multiple drop heights and acute

verbal cues, the study highlights the capacity of force–time analysis to capture meaningful variations in jump execution and landing mechanics in female volleyball athletes, for whom reactive strength and efficient load management are central performance requirements.

Study

The acute effect of verbal instructions on performance and landing when dropping from different heights: the ground reaction force-time profile of drop vertical jumps in female volleyball athletes. (La Greca et al., 2024)

Method

Participant

Sixteen regional-level female volleyball players voluntarily participated in this study (age: 21.3 ± 2.6 years; stature: 1.66 ± 0.10 m; body mass: 62.0 ± 8.1 kg; BMI: 22.2 ± 1.8). Participants performed several drop vertical jumps to examine the effects of different mechanical conditions on selected parameters of the vGRF–time profile (flight time, first and second vGRF peaks, and reactive strength index). The measurements were conducted on two separate occasions, with a minimum of one day between each visit. During the first lab visit, the participants familiarized themselves with the experimental procedure. During the second laboratory visit, prior to the measurements, the participants underwent a 15-min warm-up.

Setup for jump and landing assessment

The vGRF-time profile was recorded during each DVJ and the following parameters were considered for analysis: FT (the time spent in air between the final push-off phase and before the initial ground contact at landing), the 1st peak (the peak of vGRF produced during the braking-propulsion phase), the 2nd peak (the peak of vGRF produced during the landing phase), and the RSI (the ratio between flight time and contact time) (Jarvis et al., 2022) (Figure 5.5).

Statistical analysis

A single-group repeated-measures design was adopted to analyse kinetic and kinematic variables derived from the vGRF–time profile. Statistical analyses were performed using XLSTAT (version 2013.2.07; Addinsoft, New York). Normality of the dependent variables was confirmed by the Shapiro–Wilk test. A two-way repeated-measures mixed-model ANOVA with a compound symmetry covariance structure was used, with Bonferroni correction applied to adjust p-values for multiple comparisons. Effect sizes for the ANOVA were quantified using partial eta squared (η^2), interpreted as small (0.01), moderate (0.06), or large (0.14), while effect sizes for pairwise contrasts were calculated

using Hedges' g , classified as small (<0.5), moderate ($0.5\text{--}0.8$), or large (>0.8). Statistical significance was set at $\alpha = 0.05$.

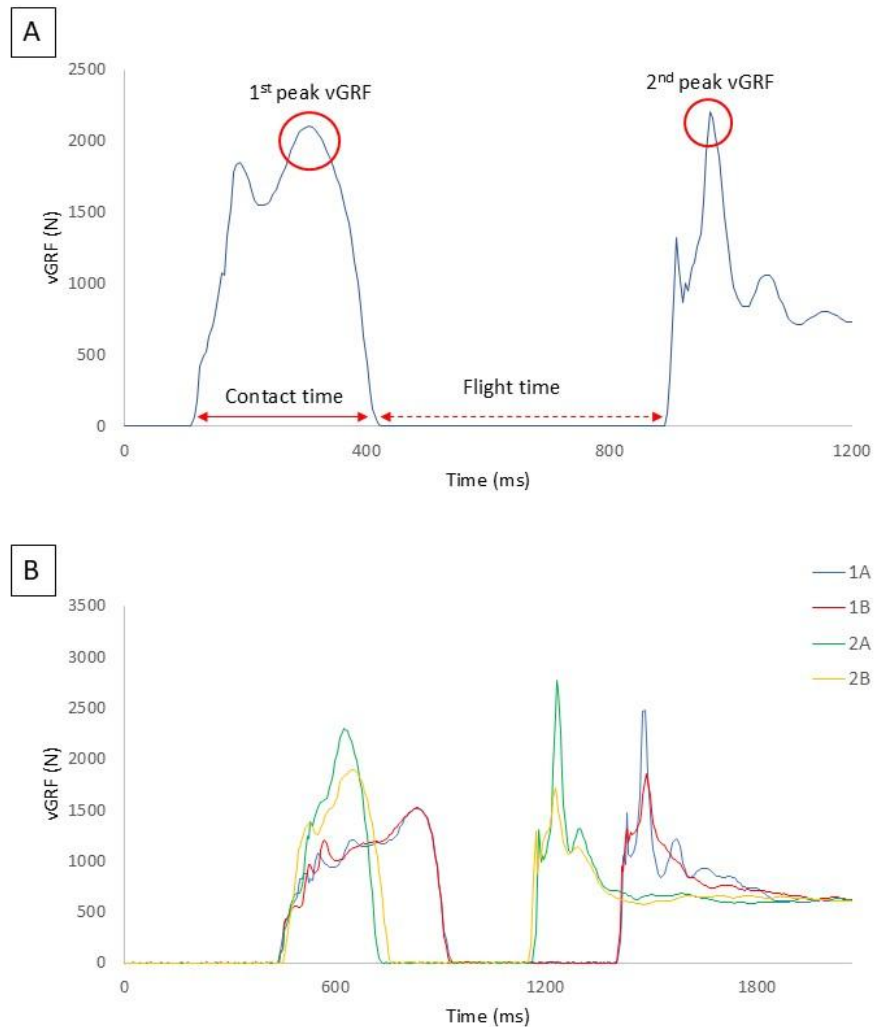


Figure 5.5. A) vGRF-time profile and the underlined parameters (1st peak vGRF; 2nd peak vGRF; contact time; flight time). B) representatives force-time profiles performed with the different verbal instructions (1A: “jump as high as possible”; 2A: “jump as quickly as possible”; 1B: “jump as high as possible and during the landing attempt to dampen the impact at ground contact”; 2B: “jump as high as quickly as possible and during the landing attempt to dampen the impact at ground contact”). vGRF, vertical ground reaction force. (From: La Greca et al., 2024)

Results

The analysis of the vGRF–time profile during drops vertical jumps revealed that flight time was influenced by drop height and verbal instructions ($F_{(2,180)} = 3.419$; $P = 0.035$; $\eta_p^2 = 0.04$ – $F_{(3,180)} = 3.182$; $P = 0.024$; $\eta_p^2 = 0.05$ respectively).

The braking phase (1st peak of vGRF) increased with higher drop heights, reflecting

greater impact loads [$F_{(2,180)} = 20.036$; $P = 0.0001$; $\eta_p^2 = 0.18$].

Conversely, the landing phase (2nd peak of vGRF) was primarily affected by verbal instructions ($F_{(3,180)} = 58.659$; $P < 0.0001$; $\eta_p^2 = 0.49$), rather than by drop height ($F_{(2,180)} = 0.318$; $P = 0.728$; $\eta_p^2 = 0.003$).

The reactive strength index (RSI) was influenced by drop height [$F_{(2,180)} = 3.636$; $P = 0.028$; $\eta_p^2 = 0.04$] and verbal instructions [$F_{(3,180)} = 69.800$; $P = 0.0001$; $\eta_p^2 = 0.54$] representing the efficiency of the transition from braking to propulsion, (Figure 5.6).

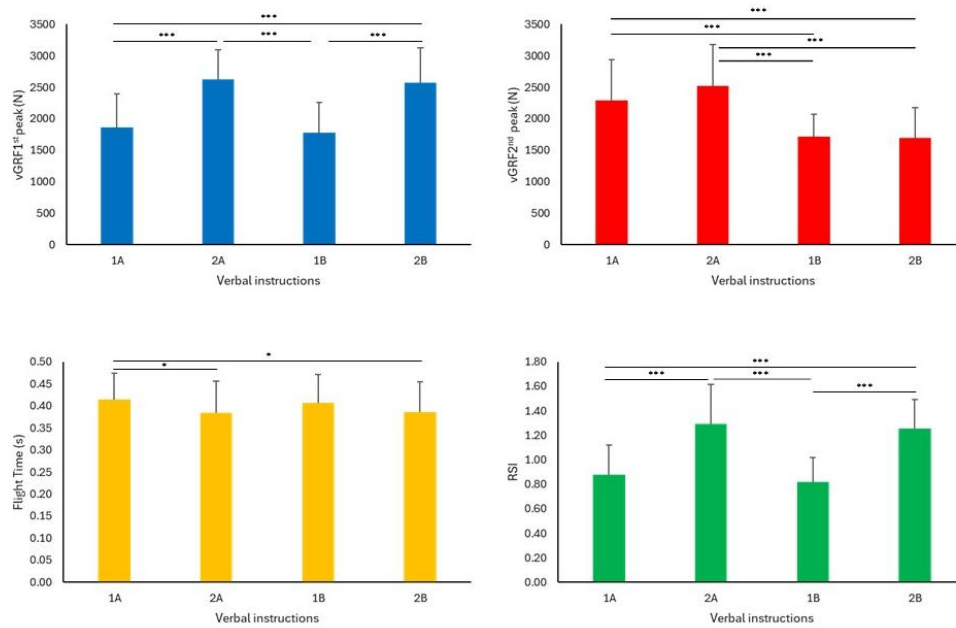


Figure 5.6. Mean values and standard deviation of the different parameters are reported. *Significant difference: $P \leq 0.05$; ***significant difference: $P < 0.01$. vGRF, vertical ground reaction force; RSI, reactive strength index. (From: La Greca et al., 2024)

These results indicate that the analysis of the vGRF force–time profile during drop vertical jumps allows the detection of phase-specific modulations in force regulation, highlighting subtle adjustments in impact attenuation and reactive force production that reflect underlying neuromuscular control strategies beyond conventional performance outcomes.

5.3.2 Performance and asymmetries

The DVJ is traditionally assessed through the analysis of the vGRF–time profile obtained from a single force platform, providing a global representation of force production during landing and propulsion phases (Bosco et al., 1982b; Komi, 2000). As discussed in the previous subsection, force–time analysis during DVJ tasks is particularly effective in identifying phase-specific features of force regulation, impact attenuation, and reactive strength during plyometric actions.

However, when performance is characterised by asymmetrical motor demands, single-platform assessments may mask relevant between-limb differences in force–time behaviour. This limitation is especially pertinent in sports involving repeated unilateral or role-differentiated actions, where each limb is exposed to distinct mechanical and neuromuscular demands over time (Douda et al., 2002; Virgile and Bishop, 2021). In these contexts, asymmetries may emerge not only in force magnitude, but also in the temporal organisation of force application across the different phases of the movement.

The DVJ represents an ecologically valid task for investigating these phenomena, as it reproduces key biomechanical characteristics of sport-specific explosive actions, including rapid stretch–shortening cycles, high impact forces, and short ground contact times (Bobbert and Knoek Van Soest, 2001; Di Cagno et al., 2009; Komi, 2000). When examined through the force–time profile, the DVJ allows the simultaneous evaluation of performance-related outcomes and landing load management strategies, which are relevant both for training monitoring and injury prevention purposes (Davies et al., 2015). The use of dual force platforms enables the simultaneous acquisition of synchronized force–time profiles for each limb, overcoming the limitations of single-platform assessments. This methodological approach allows the quantification of asymmetries related to sport specialisation not only in peak forces or impulses, but also in the temporal distribution of force across movement phases. Consequently, asymmetry is no longer interpreted solely as a difference in force magnitude between limbs, but as a potential expression of distinct neuromuscular strategies adopted to meet task-specific mechanical demands.

Importantly, inter-limb asymmetries should be interpreted within a functional perspective. In highly trained athletes, asymmetrical force–time patterns may reflect adaptive specialisation rather than impairment, particularly when sport-specific tasks require differentiated roles between limbs. For this reason, the evaluation of asymmetry through multiple computational approaches, as proposed in the literature, provides a more

comprehensive representation of lower-limb function and helps to avoid oversimplified interpretations based on a single asymmetry index (Bishop et al., 2019; Parkinson et al., 2021).

In this context, the analysis of synchronized limb-specific force–time profiles during DVJ tasks represents a valuable tool for exploring how asymmetries manifest during high-demand plyometric actions. The following study applies this approach to young rhythmic gymnasts, with the aim of characterising performance-related force–time features and inter-limb asymmetries using dual force platform assessment.

Study

Drop vertical jump performance and asymmetries in young rhythmic gymnasts.

(Passacantando et al., 2025)

Method

Participant

This study involved 15 rhythmic gymnastics athletes of silver and gold levels, competing in series C Gold Group Championships. All participants were involved in an intensive training schedule, practicing six days per week, four hours per session. Their anthropometric characteristics are summarized in Table 5.3.

Table 5.3. Anthropometric Characteristics. (From: Passacantando et al., 2025)

Variable	Median (IQR)
Age (Years)	12 (3)
Stature (cm)	155 (16)
Body Mass (Kg)	43 (14)
BMI (Kg/m ²)	18.4 (4)
Years of practice (Years)	8 (3)
CSA (cm ²)	PREFERRED 41.1 (35.11)
	NON PREFERRED 41.3 (45.89)

Abbreviation: BMI (body mass index), CSA (cross-sectional area), IQR (interquartile range)

Setup for jump and landing assessment

Participants familiarised with the experimental procedures during the two weeks preceding testing. On the test day, a standardized 40-min warm-up was performed, including ankle mobility, plyometric activities, spinal and lower-limb mobility, and flexibility exercises.

The DVJs were performed from a 30-cm box in randomized order, stepping off with either the preferred or non-preferred limb. The preferred limb was defined as the limb used to initiate technical jumps during rhythmic gymnastics routines. Participants were instructed to keep their hands on their hips, land bilaterally, and immediately execute a quickly vertical jump.

VGRF–time data were collected using dual force platforms (Musclelab, 6000 system, Ergotest Technology, Norway; sampling frequency 1000 Hz), enabling synchronized acquisition of limb-specific force signals. Data processing was performed using

Musclelab software (v10.4).

The following variables were extracted from the vGRF–time profile: flight time (FT); total contact time (CT), including braking and propulsion phases; reactive strength index (RSI = FT/CT) (Kotsifaki et al., 2022); phase-specific vGRF peaks associated with initial impact (1st vGRF peak), braking (2nd vGRF peak), propulsion (3rd vGRF peak), and landing (4th vGRF peak); and peak power output during the propulsive phase. Phase identification was based on changes in vGRF exceeding a threshold of approximately 0–20 N (Figure 5.7).

Inter-limb asymmetry was quantified using two approaches. A normalized bilateral strength asymmetry index (BSA-1) was calculated relative to the preferred limb (Parkinson et al., 2021). In addition, absolute inter-limb differences (BSA-2) were computed for each force–time variable to account for phase-specific limb roles and avoid assumptions regarding limb dominance (Bates et al., 2013).

$$BSA - 1 = \left(\frac{Pref - NPref}{Pref} \right) * 100$$

$$BSA - 2 = Pref - NPref$$

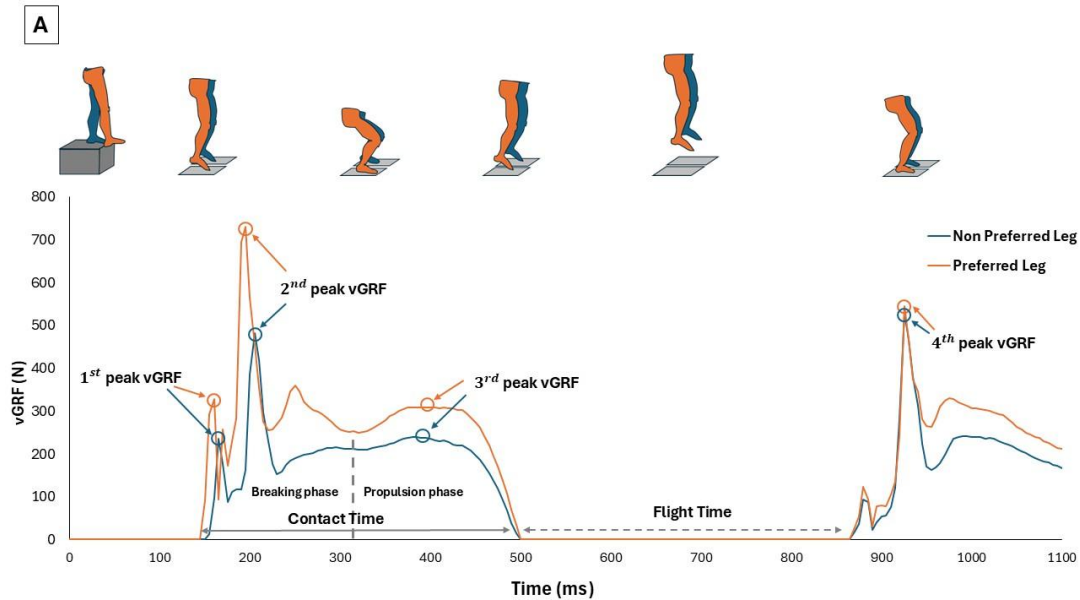


Figure 5.7. Cont.

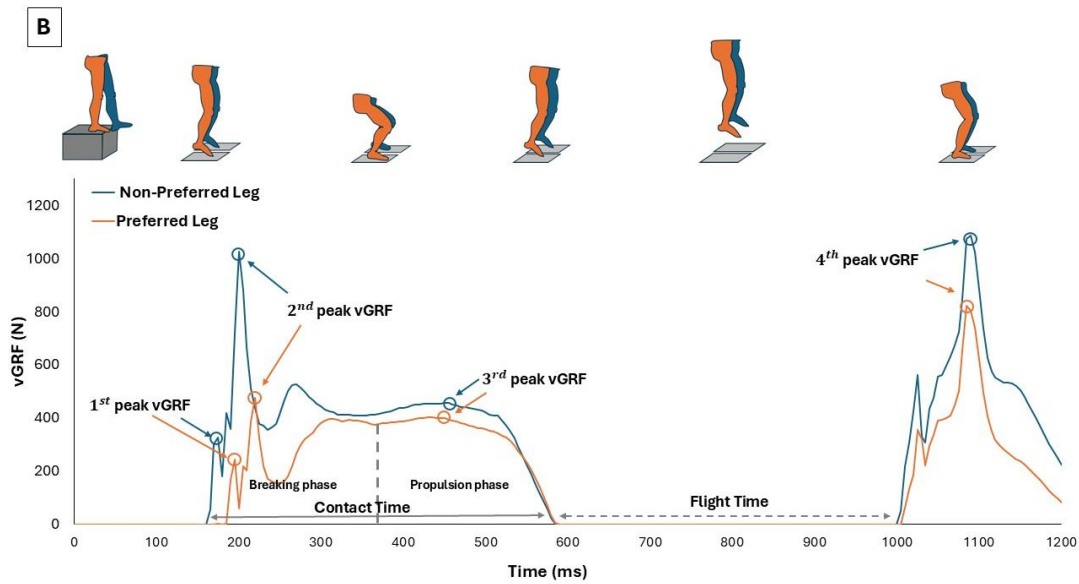


Figure 5.7. vGRF-time profile and the underlined parameters (1st peak vGRF; 2nd peak vGRF; 3rd peak vGRF; 4th peak vGRF; contact time; flight time). (From: Passacantando et al., 2025)

Statistical analysis

The data were not normally distributed, as revealed by the Shapiro–Wilk’s W test; therefore, a non-parametric statistical procedure was used. Differences in several variables between the preferred and non-preferred limbs during DVJs, were tested using the Wilcoxon signed-rank test/two-tailed test, and the p-values were computed using an exact method. The analyses were performed using XLSTAT (Statistical and Data Analysis Solution, Addinsoft, New York, NY, USA 2022). Statistical significance was set at $p \leq 0.05$, and the meaningfulness of significant outcomes was estimated by dividing the Z-statistic by the square root of the sample size ($r = z / \sqrt{N}$) (Fritz et al., 2012). The ES values were considered to be small at $r < 0.3$, moderate at $0.3 < r < 0.5$, and large at $r > 0.5$ (Cohen, 2013b).

Results

When RG athletes stepped off the box with the preferred limb, a significant inter-limb difference was observed at the 1st vGRF peak ($p = 0.026$; ES = 0.57) (Figure 5.8). The median (IQR) asymmetry values were 20.1% (43.4) for BSA-1 and 101.7 N (188.1) for BSA-2.

A significant difference was also detected at the 3rd vGRF peak ($p = 0.038$; ES = 0.54), with BSA-1 equal to 16.2% (26.1) and BSA-2 equal to 63.7 N (167.7). No significant inter-limb differences emerged at the 2nd ($p = 0.079$) or 4th vGRF peak ($p = 0.639$) (Figure

5.8)

When athletes stepped off the box with the non-preferred limb, a significant inter-limb difference was found at the 2nd vGRF peak ($p = 0.008$; $ES = 0.68$), with BSA-1 equal to -13.9% (74.1) and BSA-2 equal to -81.8 N (322.5).

Additionally, during step-off with the preferred limb, a significant positive correlation was observed between BSA-2 calculated at the 3rd vGRF peak and power output ($r = 0.54$, $p = 0.038$) (Figure 5.9).

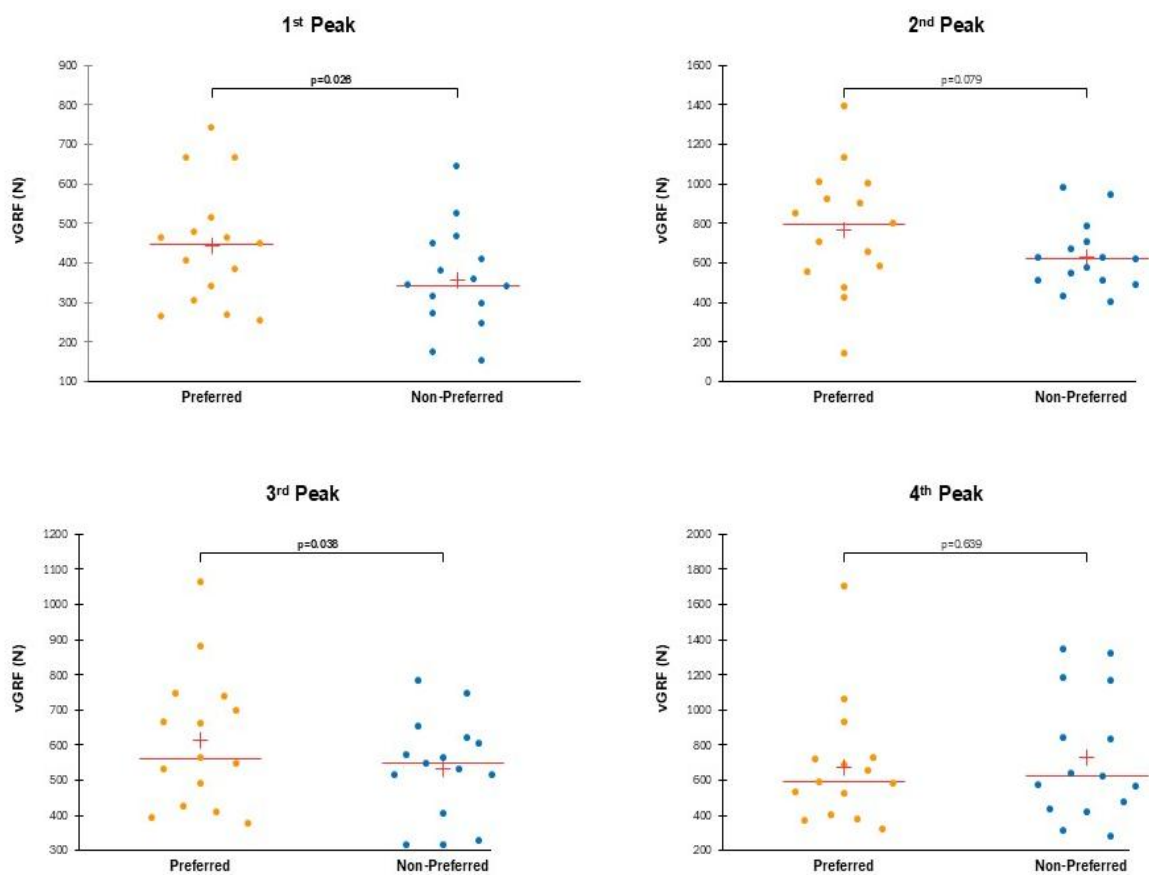


Figure 5.8. Scattergram plot of the different vGRF peaks, when dropping with preferred leg, are reported. vertical ground reaction force (vGRF). (From: Passacantando et al., 2025).

These findings confirm that dual force-time analysis during drop vertical jumps enables the identification of phase-specific and task-dependent inter-limb asymmetries during drop vertical jumps, providing insight into inter-limb force production strategies beyond conventional performance measures.

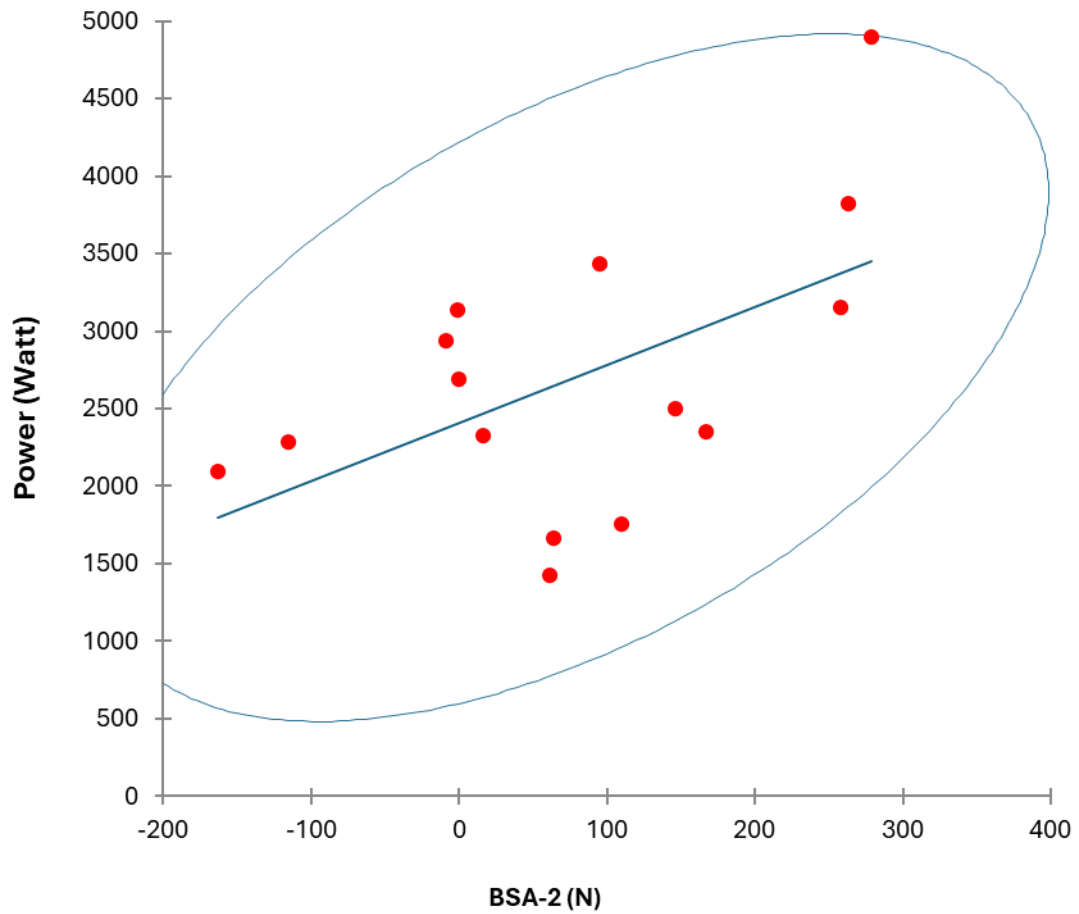


Figure 5.9. Correlation between power (during preferred step-off from the box) and the bilateral strength asymmetries (BSA-2). The BSA-2 was calculated at the 3rd peak of vGRF-time profile. vGRF: vertical ground reaction force. (From: Passacantando et al., 2025)

5.4 Synthesis and applied clinical and sport contexts implications

Force–time profile–based assessment appears to allow the analysis of neuromuscular performance beyond the limitations of traditional tests that rely exclusively on maximal values or composite performance indices. The temporal analysis of force production enables the examination of how force is developed, modulated, and coordinated across the different phases of a motor task, providing more sensitive information on the neuromuscular control strategies adopted by the individual (Enoka, 2015; Payton et al., 2018).

In clinical settings, this approach is particularly relevant in individuals with anterior cruciate ligament reconstruction, in which unilateral tests may fail to detect residual deficits despite the restoration of apparently symmetrical force and rate of force development values. In such cases, the execution of bilateral tasks has shown a greater ability to reveal alterations in agonist–antagonist coordination and load distribution between limbs, suggesting the adoption of protective neuromuscular strategies that do not emerge during unilateral assessments (Di Giminiani et al., 2023). This indicates that task bilateral represents a more sensitive evaluative constraint for identifying latent functional deficits in clinically rehabilitated individuals.

A similar principle emerges in the assessment of handgrip strength. Although the traditional handgrip test is widely used as a global indicator of neuromuscular function (American College of Sports Medicine et al., 2026), it provides a point estimate of maximal force and does not allow the investigation of the mechanisms underlying force production. The integration of handgrip assessment with force–time profile analysis makes it possible to distinguish between maximal force, rate of force development, and neuromuscular activation strategies, thereby identifying adaptations that would remain undetected when using conventional hydraulic devices. In this context, increases in rate of force development and improvements in inter-limb symmetry, even in the absence of significant changes in peak force, suggest possible central and cortical adaptations in motor control (La Greca et al., 2025b).

In the sport context, the application of force–time analysis to the Drop Vertical Jump (DVJ) allows the decomposition of performance into its functional phases, highlighting differences between eccentric, concentric, and final landing phases. In rhythmic gymnastics athletes, for example, overall performance metrics (flight time, contact time, reactive strength index, and power) were preserved, while specific asymmetries emerged in vertical ground reaction force peaks. These findings are consistent with different

functional roles of the limbs during the execution of technical jumps (Passacantando et al., 2025). Although such asymmetries do not impair short-term performance, they may represent a potential risk factor if not monitored over time (Helme et al., 2021).

Moreover, the present findings demonstrate that the force–time profile is also sensitive in detecting motor control strategies adopted during landing. During Drop Vertical Jump assessments with targeted verbal instructions, force–time analysis revealed a reduction in vertical force peaks without compromising the required jump performance (La Greca et al., 2024). This observation reinforces the concept that assessment should not be limited to quantifying how much force is produced but should also consider how force is managed over time.

Overall, the performance assessment tests based on force–time profile analysis proposed in this thesis represent an innovative evaluative approach. These tests are designed according to the characteristics of the individual and the specific context, whether clinical or sport. They allow the selection of the most appropriate motor constraint, including bilateral or unilateral execution and the presence or absence of verbal instructions, and enable the identification of specific neuromuscular adaptations that traditional tests are unable to detect.

Therefore, force–time profile–based assessment can be considered a more specific, sensitive, and functional tool for the analysis of performance and neuromuscular control, providing a solid foundation for targeted evaluation strategies aligned with the objectives of monitoring, injury prevention, and performance development.

CHAPTER 6 – Discussion and Applied Implications

6.1 General discussion

The results of this thesis consistently show that the proposed assessment methods can describe, detect, and characterize changes, adaptations, or alterations more accurately and specifically than traditional or commonly used evaluations (Di Giminiani et al., 2024, 2023, 2022; La Greca et al., 2025a, 2025b). The precise nature of the detected features varies across populations. In ACL-R athletes, dynamic bilateral tests revealed persistent postural and neuromuscular alterations, even after clinical rehabilitation (Di Giminiani et al., 2023). In PwMS, these methods identified subtle and specific adaptations in joint coordination and muscle activation that provides a neuromuscular explanation for the alteration detected by the angle-angle diagram (Di Giminiani et al., 2022; La Greca et al., 2025b). In individuals with ASD, the tests provided a more descriptive and detailed representation of already established gait and postural alterations, highlighting coordination and neuromuscular patterns in a functionally meaningful way (Di Giminiani et al., 2024).

Dynamic bilateral balance assessments, particularly when synchronized with EMG, allowed a precise characterization of postural strategies, including compensatory muscle activation and proximal–distal coordination (Di Giminiani et al., 2024; Winter, 1995). This approach provided greater accuracy than static or unilateral tasks, revealing functional aspects of postural control that are not captured by traditional clinical tests (Bodkin et al., 2018; Winter, 1995).

Angle–angle diagrams synchronized with EMG proved highly effective in detecting fine changes in coordination and neuromuscular control. Knee–ankle diagrams in PwMS revealed constrained distal movement and compensatory proximal muscle activation (La Greca et al., 2025b, 2022), whereas in ASD participants, reduced loop area and altered muscle recruitment reflected differences in motor organization rather than joint limitations (Di Giminiani et al., 2024). ROC analyses confirmed the high sensitivity and specificity of this approach. Compared with literature, traditional gait descriptors often fail to capture these subtle coordination changes, especially when EMG information is not integrated (Armitano et al., 2020; Boudarham et al., 2016; Filli et al., 2018).

Force–time profile analysis further complements kinematic and EMG measures. In

clinical and sport settings, these tests can reveal phase-specific neuromuscular strategies and inter-limb asymmetries that remain hidden when only peak values are considered. In our study, force–time profiles demonstrated the potential to capture subtle adjustments in force modulation and control, which could indicate early adaptations or compensatory strategies during task execution (La Greca et al., 2024). Drop Vertical Jump assessments highlighted eccentric-concentric and landing phase differences without altering global performance metrics, aligning with previous reports on limb-specific functional roles in athletes (Helme et al., 2021; Passacantando et al., 2025). Bilateral tasks were again more sensitive than unilateral execution, consistent with the higher ICCs observed across balance and strength assessments (Di Giminiani et al., 2023; La Greca et al., 2025a). Overall, across populations and tasks, the integrated use of dynamic, bilateral, and physiologically informed assessments allows a more accurate, specific, and functional characterization of postural control, gait, and neuromuscular performance than standard or commonly applied tests. These methods not only detect persistent deficits in clinical populations, but also describe subtle changes, compensatory strategies, or functional adaptations in neurological and developmental conditions, providing a valuable tool for rehabilitation, clinical monitoring, and performance optimization. Importantly, our findings also highlight the possibility of adapting the tests to the characteristics of the sample, allowing acquisition of the maximum amount of functionally relevant information with respect to daily life activities and sport practice, thereby enhancing ecological validity and applicability in real-world contexts.

6.2 Continuum between special and healthy populations

Our findings suggest that neuromuscular function, coordination, and postural control may exist along a continuum rather than as categorical states. Alterations, adaptations, or subtle differences appear to vary progressively from clinical populations to healthy individuals. In ASD, our assessments provide a detailed description of already present functional differences, highlighting patterns and magnitudes of variation more clearly than conventional evaluations. In people with multiple sclerosis, small adjustments and compensatory strategies can be observed even when gross function appears preserved. In ACL-reconstructed athletes, persistent alterations coexist with adaptive responses during dynamic or bilateral tasks. Even in healthy athletic populations, variations in coordination, force control, or inter-limb strategies are detectable, reflecting natural

differences or sport-specific adaptations.

These observations indicate that the assessments we propose could enable a deeper understanding of neuromuscular behaviour in different populations. While these tools do not aim to classify individuals or serve as formal diagnostics, they offer a flexible approach to capture differences, adaptations, and functional patterns in relation to everyday activities and sport-specific movements.

Overall, the results support the idea that postural, locomotor, and force-control assessments can be adapted to the sample and context, potentially providing richer information about how individuals organize movement and respond to functional challenges. This perspective emphasizes possibilities and trends rather than definitive conclusions, underlying the exploratory and applied nature of our methods.

6.3 Implications for rehabilitation and sport training periodization

Considering the results discussed in this thesis, it is suggested that biomechanical assessments based on task-specific constraints and integrated neuromuscular analysis may support more informed rehabilitation and sport training programme. Rather than relying exclusively on isolated or maximal performance outcomes, the proposed approaches emphasize how movement is organized, controlled, and adapted under different functional demands.

In rehabilitation settings, the use of dynamic and context-dependent tasks may help clinicians better describe residual impairments, emerging adaptations, or compensatory strategies that are not evident during traditional assessments. This is particularly relevant when recovery is evaluated beyond symptom resolution or return to baseline strength values. By manipulating task constraints, such as bilateral versus unilateral execution or the presence of external perturbations, assessments can be aligned more closely with the functional demands of daily activities and sport-specific actions (Enoka, 2015; Winter, 1995). Within this framework, neuromuscular measures provide complementary information on how force and coordination are managed over time, rather than on how much force is produced.

In sport training contexts, the integration of force–time analysis and coordination-based metrics offers practical insights into movement quality and load management. The force–time profile, for example, allows coaches and practitioners to monitor how athletes absorb, generate, and redistribute forces across task phases, supporting the identification

of potentially inefficient or asymmetric strategies even when overall performance is preserved (Payton et al., 2018). This information may be useful for guiding exercise selection, technical refinement, and progression strategies aimed at optimizing performance while limiting excessive mechanical stress.

Across both clinical and sport domains, an important implication emerging from this work is the flexibility of the proposed assessments. Tests can be adapted to the characteristics of the population, the stage of rehabilitation or training, and the specific functional goals. This adaptability supports a more individualized approach, in which assessment is not a fixed protocol but a tool that evolves with the individual to capture meaningful changes in motor organization and coordination (Latash, 2008; Winter, 2009), related to real-life motor actions and sport practice.

The proposed methods should be considered exploratory rather than prescriptive, they highlight the potential value of assessment strategies that prioritize movement organization, neuromuscular control, and task specificity to better inform rehabilitation and training programming.

6.4 Methodological limitations of the thesis

Some methodological limitations should be acknowledged to ensure a correct interpretation of the findings and to define the limits of their applicability. These considerations are presented for scientific completeness and do not undermine the overall validity of the proposed approaches.

A first limitation concerns the complexity of the assessment setups. The protocols required advanced instrumentation, controlled environments, and relatively long testing procedures, which may limit their feasibility in routine clinical practice. Moreover, several of the gold-standard tools used, such as force platforms, motion analysis systems, and surface electromyography, are expensive and not always easy to find.

A further limitation relates to the integration of biomechanical variables. Although the combined analysis of kinetics, kinematics, and electromyography provides a more comprehensive understanding of neuromuscular function, not all our studies included comprehensive integration of multiple variables. Nevertheless, the strict methodological rigor of the protocols allowed the identification of meaningful and functionally relevant adaptations, even when analyses were focused on specific domains. Future assessment context should aim to systematically integrate kinetic, kinematic, and electromyographic

measures to further enhance interpretability.

Finally, another aspect to consider is the specificity of the population investigated. The studies were conducted on selected clinical and athletic groups, and therefore the results cannot be generalized to all populations. The observed adaptations and alterations should be interpreted in relation to the functional demands and characteristics of the tested individuals.

Despite these limitations, the consistency of findings across different tasks and populations supports the robustness of the proposed assessment approaches. The limitations primarily reflect practical and contextual constraints rather than conceptual weaknesses. Overall, the rigorous execution of the protocols allowed the extraction of relevant information on movement control and neuromuscular strategies, supporting the value of these methods as advanced and promising tools for biomechanical assessment in clinical and sport contexts.

6.5 Future perspectives in biomechanical assessment

Future developments in biomechanical assessment are and will continue to be significantly influenced by technological advances that enable more integrated, accessible, and efficient assessment tools. The progressive reduction in costs and increased availability of validated wearable sensors, portable force measurement systems, and wireless electromyography should facilitate the implementation of multidimensional assessments outside of specialized laboratory settings.

In this context, the integration of machine learning and artificial intelligence techniques represents a promising perspective. These approaches can support data processing, pattern recognition, and automated feature extraction, simplify acquisition procedures and enable real-time or near real-time analysis (Edriss et al., 2025). Such developments may reduce the methodological difficulty associated with complex biomechanical configurations, while preserving the quality and interpretability of the information collected.

From a methodological point of view, future assessment protocols should continue to be designed according to the specific characteristics of the population and the functional task under investigation. The synchronized acquisition of kinetic, kinematic, and electromyographic variables remains a key objective to improve the functional relevance of biomechanical evaluations. Technological advances may allow the development of faster, more user-friendly, and task-specific setups that better reflect real-life and sport-

related motor demands.

To further strengthen the scientific robustness of these approaches, future studies should include larger sample sizes and wider populations. Although several of the proposed tests have demonstrated good reliability and, in some cases, high sensitivity and specificity, their validity should be further explored through direct comparisons with traditional assessment methods. This step is essential to contextualize their added value within existing evaluation frameworks.

Longitudinal study designs also represent an important future direction. Monitoring biomechanical and neuromuscular adaptations over time, across different stages of rehabilitation, training, or disease progression, would provide deeper insight into the functional significance of the observed changes. Moreover, extending these approaches to other populations characterized by specific motor impairment may further clarify their applicability and limitations.

In general, future perspectives in biomechanical assessment point to the progressive integration of advanced technologies, individualized protocol design, and longitudinal validation. In this context, the approaches proposed in this thesis should be considered as evolving methodological tools, with the potential to support more specific, sensitive, and function-oriented assessment strategies in both clinical and sport contexts.

CONCLUSIONS

This thesis investigated biomechanical and neuromuscular assessment approaches aimed at improving the functional interpretation of human movement across clinical and sport contexts. Moving beyond traditional assessments based on isolated, maximum, or summary outcome measures, the proposed methodologies focused on the analysis of movement organization, force–time characteristics, and neuromuscular control strategies under task-specific conditions.

Across different populations and motor tasks, the results consistently showed that assessments incorporating dynamic conditions, bilateral execution, and the integration of electromyographic variables can describe movement behaviour with greater specificity and functional relevance. Rather than merely identifying the presence or absence of impairment, these approaches allowed a more detailed characterization of how motor strategies, coordination patterns, and force modulation emerge and adapt in response to task demands.

In clinical populations, the findings highlighted how subtle alterations or residual adaptations may persist despite apparently normal performance when evaluated with conventional tests. In sport settings, the same models have proven useful for identifying specific characteristics and asymmetries that are not necessarily limit performance in the short term but may be relevant for monitoring load management and movement quality. Importantly, these observations support the idea of a functional continuum between special and healthy populations, in which differences are often quantitative and strategy-related rather than categorical.

The research project presented in this thesis supports the use of biomechanical assessment as a flexible and context-dependent tool, which is valuable not only for its measurement accuracy, but also in its ability to adapt to the characteristics of the individual and the functional meaning of the task. Within this perspective, the proposed methods should be considered as complementary and evolving solutions, capable of enhancing evaluation strategies for rehabilitation, performance monitoring, and movement analysis.

References

- Aagaard, P., Simonsen, E.B., Andersen, J.L., Magnusson, P., Dyhre-Poulsen, P., 2002. Increased rate of force development and neural drive of human skeletal muscle following resistance training. *Journal of Applied Physiology* 93, 1318–1326. <https://doi.org/10.1152/japplphysiol.00283.2002>
- Aagaard, P., Suetta, C., Caserotti, P., Magnusson, S.P., Kjær, M., 2010. Role of the nervous system in sarcopenia and muscle atrophy with aging: strength training as a countermeasure. *Scandinavian Med Sci Sports* 20, 49–64. <https://doi.org/10.1111/j.1600-0838.2009.01084.x>
- American College of Sports Medicine, Ozemek, C., Bonikowske, A.R., Christle, J.W., Gallo, P.M. (Eds.), 2026. ACSM's guidelines for exercise testing and prescription, Twelfth edition. ed. Wolters Kluwer, Philadelphia, PA.
- American Psychiatric Association, 2013. Diagnostic and Statistical Manual of Mental Disorders, Fifth Edition. ed. American Psychiatric Association. <https://doi.org/10.1176/appi.books.9780890425596>
- Ardern, C.L., Webster, K.E., Taylor, N.F., Feller, J.A., 2011. Return to sport following anterior cruciate ligament reconstruction surgery: a systematic review and meta-analysis of the state of play. *Br J Sports Med* 45, 596–606. <https://doi.org/10.1136/bjsm.2010.076364>
- Armitano, C.N., Bennett, H.J., Haegele, J.A., Morrison, S., 2020. Assessment of the gait-related acceleration patterns in adults with autism spectrum disorder. *Gait & Posture* 75, 155–162. <https://doi.org/10.1016/j.gaitpost.2019.09.002>
- Bahr, R., Krosshaug, T., 2005. Understanding injury mechanisms: a key component of preventing injuries in sport. *Br J Sports Med* 39, 324–329. <https://doi.org/10.1136/bjsm.2005.018341>
- Bates, N.A., Ford, K.R., Myer, G.D., Hewett, T.E., 2013. Impact differences in ground reaction force and center of mass between the first and second landing phases of a drop vertical jump and their implications for injury risk assessment. *Journal of Biomechanics* 46, 1237–1241. <https://doi.org/10.1016/j.jbiomech.2013.02.024>
- Bennett, H.J., Ringleb, S.I., Bobzien, J., Haegele, J.A., 2021. Walking lower extremity biomechanics of adolescents with autism spectrum disorder. *Journal of Biomechanics* 119, 110332. <https://doi.org/10.1016/j.jbiomech.2021.110332>
- Bernstein, N., 1967. *Coordinate and Regulation of Movements*. Pergamon P.
- Beynon, B.D., Uh, B.S., Johnson, R.J., Abate, J.A., Nichols, C.E., Fleming, B.C., Poole, A.R., Roos, H., 2005. Rehabilitation after Anterior Cruciate Ligament Reconstruction: A Prospective, Randomized, Double-Blind Comparison of Programs Administered over 2 Different Time Intervals. *Am J Sports Med* 33, 347–359. <https://doi.org/10.1177/0363546504268406>
- Bishop, C., Turner, A., Maloney, S., Lake, J., Loturco, I., Bromley, T., Read, P., 2019. Drop Jump Asymmetry is Associated with Reduced Sprint and Change-of-Direction Speed Performance in Adult Female Soccer Players. *Sports* 7, 29. <https://doi.org/10.3390/sports7010029>
- Bobbert, M.F., Knoek Van Soest, A.J., 2001. Why Do People Jump the Way They Do?: Exercise and Sport Sciences

- Reviews 29, 95–102.
<https://doi.org/10.1097/00003677-200107000-00002>
- Bodkin, S.G., Slater, L.V., Norte, G.E., Goetschius, J., Hart, J.M., 2018. ACL reconstructed individuals do not demonstrate deficits in postural control as measured by single-leg balance. *Gait & Posture* 66, 296–299.
<https://doi.org/10.1016/j.gaitpost.2018.06.120>
- Bohannon, R.W., 2019. Grip Strength: An Indispensable Biomarker For Older Adults. *CIA Volume* 14, 1681–1691.
<https://doi.org/10.2147/CIA.S194543>
- Bosco, C., 2002. La valutazione della forza con il test di Bosco, 2. rist. ed. Società stampa sportiva, Roma.
- Bosco, C., 1999. Strenght Assessment with the Bosco's Test. Italian society of sport science.
- Bosco, C., 1997. La forza muscolare: aspetti fisiologici ed applicazioni pratiche. Società stampa sportiva, Roma.
- Bosco, C., Komi, P.V., 1979. Mechanical characteristics and fiber composition of human leg extensor muscles. *Europ. J. Appl. Physiol.* 41, 275–284.
<https://doi.org/10.1007/BF00429744>
- Bosco, C., Komi, P.V., Ito, A., 1981. Prestretch potentiation of human skeletal muscle during ballistic movement. *Acta Physiologica Scandinavica* 111, 135–140.
<https://doi.org/10.1111/j.1748-1716.1981.tb06716.x>
- Bosco, C., Tihanyi, J., Komi, P.V., Fekete, G., Apor, P., 1982a. Store and recoil of elastic energy in slow and fast types of human skeletal muscles. *Acta Physiologica Scandinavica* 116, 343–349.
<https://doi.org/10.1111/j.1748-1716.1982.tb07152.x>
- Bosco, C., Viitasalo, J.T., Komi, P.V., Luhtanen, P., 1982b. Combined effect of elastic energy and myoelectrical potentiation during stretch-shortening cycle exercise. *Acta Physiologica Scandinavica* 114, 557–565.
<https://doi.org/10.1111/j.1748-1716.1982.tb07024.x>
- Bosco, C., Viru, A., 1995. *Biologia dell'allenamento*. Società stampa sportiva, Roma.
- Boudarham, J., Hameau, S., Zory, R., Hardy, A., Bensmail, D., Roche, N., 2016. Coactivation of Lower Limb Muscles during Gait in Patients with Multiple Sclerosis. *PLoS ONE* 11, e0158267.
<https://doi.org/10.1371/journal.pone.0158267>
- Cacolice, P.A., Carcia, C.R., Scibek, J.S., Phelps, A.L., 2020. Ground Reaction Forces Are Predicted with Functional and Clinical Tests in Healthy Collegiate Students. *JCM* 9, 2907.
<https://doi.org/10.3390/jcm9092907>
- Calhoun, M., Longworth, M., Chester, V.L., 2011. Gait patterns in children with autism. *Clinical Biomechanics* 26, 200–206.
<https://doi.org/10.1016/j.clinbiomech.2010.09.013>
- Cavanagh, P.R., 1987. The Biomechanics of Lower Extremity Action in Distance Running. *Foot & Ankle* 7, 197–217.
<https://doi.org/10.1177/107110078700700402>
- Cham, R., Iverson, J.M., Bailes, A.H., Jennings, J.R., Eack, S.M., Redfern, M.S., 2021. Attention and sensory integration for postural control in young adults with autism spectrum disorders. *Exp Brain Res* 239, 1417–1426.
<https://doi.org/10.1007/s00221-021-06058-z>
- Chester, V.L., Calhoun, M., 2012. Gait Symmetry in Children with Autism. *Autism Research and Treatment* 2012, 1–5.
<https://doi.org/10.1155/2012/576478>
- Cohen, J., 2013a. *Statistical Power Analysis for the Behavioral Sciences*, 0 ed. Routledge.
<https://doi.org/10.4324/9780203771587>
- Cohen, J., 2013b. *Statistical Power Analysis for the Behavioral Sciences*, 0 ed.

- Routledge.
<https://doi.org/10.4324/9780203771587>
- Coni, A., Van Ancum, J.M., Bergquist, R., Mikolaizak, A.S., Mellone, S., Chiari, L., Maier, A.B., Pijnappels, M., 2019. Comparison of Standard Clinical and Instrumented Physical Performance Tests in Discriminating Functional Status of High-Functioning People Aged 61–70 Years Old. *Sensors* 19, 449. <https://doi.org/10.3390/s19030449>
- Courel-Ibáñez, J., Hernández-Belmonte, A., Cava-Martínez, A., Pallarés, J.G., 2020. Familiarization and Reliability of the Isometric Knee Extension Test for Rapid Force Production Assessment. *Applied Sciences* 10, 4499. <https://doi.org/10.3390/app10134499>
- Cowling, E.J., Steele, J.R., McNair, P.J., 2003. Effect of verbal instructions on muscle activity and risk of injury to the anterior cruciate ligament during landing. *Br J Sports Med* 37, 126–130. <https://doi.org/10.1136/bjsm.37.2.126>
- Cronin, N.J., Kumpulainen, S., Joutjärvi, T., Finni, T., Piitulainen, H., 2015. Spatial variability of muscle activity during human walking: The effects of different EMG normalization approaches. *Neuroscience* 300, 19–28. <https://doi.org/10.1016/j.neuroscience.2015.05.003>
- Cuthbert, M., Comfort, P., Ripley, N., McMahon, J.J., Evans, M., Bishop, C., 2021. Unilateral vs. bilateral hamstring strength assessments: comparing reliability and inter-limb asymmetries in female soccer players. *Journal of Sports Sciences* 39, 1481–1488. <https://doi.org/10.1080/02640414.2021.1880180>
- Davies, G., Riemann, B.L., Manske, R., 2015. CURRENT CONCEPTS OF PLYOMETRIC EXERCISE. *Int J Sports Phys Ther* 10, 760–786.
- De Luca, C.J., 1997. The Use of Surface Electromyography in Biomechanics. *Journal of Applied Biomechanics* 13, 135–163. <https://doi.org/10.1123/jab.13.2.135>
- Di Cagno, A., Baldari, C., Battaglia, C., Monteiro, M.D., Pappalardo, A., Piazza, M., Guidetti, L., 2009. Factors influencing performance of competitive and amateur rhythmic gymnastics—Gender differences. *Journal of Science and Medicine in Sport* 12, 411–416. <https://doi.org/10.1016/j.jsams.2008.01.006>
- Di Giminiani, R., Di Lorenzo, D., La Greca, S., Russo, L., Masedu, F., Totaro, R., Padua, E., 2022. Angle-Angle Diagrams in the Assessment of Locomotion in Persons with Multiple Sclerosis: A Preliminary Study. *Applied Sciences* 12, 7223. <https://doi.org/10.3390/app12147223>
- Di Giminiani, R., Giovannelli, A., Capuano, L., Izzicupo, P., Di Blasio, A., Masedu, F., 2020. Neuromuscular Strategies in Stretch–Shortening Exercises with Increasing Drop Heights: The Role of Muscle Coactivation in Leg Stiffness and Power Propulsion. *IJERPH* 17, 8647. <https://doi.org/10.3390/ijerph17228647>
- Di Giminiani, R., La Greca, S., Marinelli, S., Attanasio, M., Masedu, F., Mazza, M., Valenti, M., 2024. Locomotion and Postural Control in Young Adults with Autism Spectrum Disorders: A Novel Kinesiological Assessment. *JFMK* 9, 185. <https://doi.org/10.3390/jfmk9040185>
- Di Giminiani, R., Marinelli, S., La Greca, S., Di Blasio, A., Angelozzi, M., Cacchio, A., 2023. Neuromuscular Characteristics of Unilateral and Bilateral Maximal Voluntary Isometric Contractions following ACL Reconstruction. *Biology* 12, 1173. <https://doi.org/10.3390/biology12091173>
- Di Giminiani, R., Petricola, S., 2016. The Power Output-Drop Height Relationship to Determine the

- Optimal Dropping Intensity and to Monitor the Training Intervention. *Journal of Strength and Conditioning Research* 30, 117–125.
<https://doi.org/10.1519/JSC.0000000000001076>
- Di Sapio, A., Bertolotto, A., Melillo, F., Sperli, F., Malucchi, S., Troni, W., 2014. A new neurophysiological approach to assess central motor conduction damage to proximal and distal muscles of lower limbs. *Clinical Neurophysiology* 125, 133–141.
<https://doi.org/10.1016/j.clinph.2013.06.018>
- Douda, H., Lapidis, K., Tokmakidis, S., 2002. Long-Term training induces specific adaptations on the physique of rhythmic sports and female artistic gymnasts. *European Journal of Sport Science* 2, 1–14.
<https://doi.org/10.1080/17461390200072304>
- Edriss, S., Romagnoli, C., Maurizi, M., Caprioli, L., Bonaiuto, V., Annino, G., 2025. Pose estimation for pickleball players' kinematic analysis through MediaPipe-based deep learning: A pilot study. *Journal of Sports Sciences* 43, 1860–1870.
<https://doi.org/10.1080/02640414.2025.2524283>
- Enoka, R.M., 2015. *Neuromechanics of human movement*, 5th ed. ed. Human kinetics, Champaign.
- Farina, D., Merletti, R., Enoka, R.M., 2014. The extraction of neural strategies from the surface EMG: an update. *Journal of Applied Physiology* 117, 1215–1230.
<https://doi.org/10.1152/jappphysiol.00162.2014>
- Fenn, W.O., 1924. The relation between the work performed and the energy liberated in muscular contraction. *The Journal of Physiology* 58, 373–395.
<https://doi.org/10.1113/jphysiol.1924.sp002141>
- Ferber, R., Macdonald, S., 2014. *Running mechanics and gait analysis*. Human Kinetics, Champaign, IL.
- Filli, L., Sutter, T., Easthope, C.S., Killeen, T., Meyer, C., Reuter, K., Lörinicz, L., Bolliger, M., Weller, M., Curt, A., Straumann, D., Linnebank, M., Zörner, B., 2018. Profiling walking dysfunction in multiple sclerosis: characterisation, classification and progression over time. *Sci Rep* 8, 4984.
<https://doi.org/10.1038/s41598-018-22676-0>
- Finni, T., Komi, P.V., Lepola, V., 2000. In vivo human triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump. *European Journal of Applied Physiology* 83, 416–426.
<https://doi.org/10.1007/s004210000289>
- Fleming, B.C., Oksendahl, H., Beynnon, B.D., 2005. Open- or Closed-Kinetic Chain Exercises After Anterior Cruciate Ligament Reconstruction?: Exercise and Sport Sciences Reviews 33, 134–140.
<https://doi.org/10.1097/00003677-200507000-00006>
- Fritz, C.O., Morris, P.E., Richler, J.J., 2012. Effect size estimates: Current use, calculations, and interpretation. *Journal of Experimental Psychology: General* 141, 2–18.
<https://doi.org/10.1037/a0024338>
- Gafner, S.C., Allet, L., Hilfiker, R., Bastiaenen, C.H.G., 2021. Reliability and Diagnostic Accuracy of Commonly Used Performance Tests Relative to Fall History in Older Persons: A Systematic Review. *CIA Volume* 16, 1591–1616.
<https://doi.org/10.2147/CIA.S322506>
- Gollhofer, A., Komi, P.V., Fujitsuka, N., Miyashita, M., 1987. Fatigue during stretch-shortening cycle exercises. II. Changes in neuromuscular activation patterns of human skeletal muscle. *Int J Sports Med* 8 Suppl 1, 38–47.
- Grooms, D.R., Page, S.J., Nichols-Larsen, D.S., Chaudhari, A.M.W., White, S.E., Onate, J.A., 2017. Neuroplasticity Associated With Anterior Cruciate Ligament

- Reconstruction. *Journal of Orthopaedic & Sports Physical Therapy* 47, 180–189. <https://doi.org/10.2519/jospt.2017.7003>
- Hallett, M., Lebieadowska, M.K., Thomas, S.L., Stanhope, S.J., Denckla, M.B., Rumsey, J., 1993. Locomotion of Autistic Adults. *Archives of Neurology* 50, 1304–1308. <https://doi.org/10.1001/archneur.1993.00540120019007>
- Hamill, J., Knutzen, K.M., Derrick, T.R., 2021. Biomechanics: 40 Years On. *Kinesiology Review* 10, 228–237. <https://doi.org/10.1123/kr.2021-0015>
- Helme, M., Tee, J., Emmonds, S., Low, C., 2021. Does lower-limb asymmetry increase injury risk in sport? A systematic review. *Physical Therapy in Sport* 49, 204–213. <https://doi.org/10.1016/j.ptsp.2021.03.001>
- Henriksson, M., Ledin, T., Good, L., 2001. Postural Control after Anterior Cruciate Ligament Reconstruction and Functional Rehabilitation. *Am J Sports Med* 29, 359–366. <https://doi.org/10.1177/03635465010290031801>
- Henrique, J.R., Ramirez-Campillo, R., Afonso, J., Silva, R., Moran, J., Clemente, F.M., 2023. Plyometric training programs in handball: a systematic scoping review. *Kinesiology (Zagreb, Online)* 55, 298–336. <https://doi.org/10.26582/k.55.2.13>
- Hermens, H.J., Freriks, B., Disselhorst-Klug, C., Rau, G., 2000. Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology* 10, 361–374. [https://doi.org/10.1016/S1050-6411\(00\)00027-4](https://doi.org/10.1016/S1050-6411(00)00027-4)
- Hershler, C., Milner, M., 1980a. Angle-angle diagrams in the assessment of locomotion. *Am J Phys Med* 59, 109–125.
- Hershler, C., Milner, M., 1980b. Angle-angle diagrams in above-knee amputee and cerebral palsy gait. *Am J Phys Med* 59, 165–183.
- Hill, 1938. The heat of shortening and the dynamic constants of muscle. *Proc. R. Soc. Lond. B* 126, 136–195. <https://doi.org/10.1098/rspb.1938.0050>
- Hoffman, M., Schrader, J., Kocejka, D., 1999. An investigation of postural control in postoperative anterior cruciate ligament reconstruction patients. *J Athl Train* 34, 130–136.
- Horak, F.B., 2006. Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? *Age and Ageing* 35, ii7–ii11. <https://doi.org/10.1093/ageing/afl077>
- Horak, F.B., Nashner, L.M., 1986. Central programming of postural movements: adaptation to altered support-surface configurations. *Journal of Neurophysiology* 55, 1369–1381. <https://doi.org/10.1152/jn.1986.55.6.1369>
- Hortobágyi, T., Granacher, U., Fernandez-del-Olmo, M., Howatson, G., Manca, A., Deriu, F., Taube, W., Gruber, M., Márquez, G., Lundbye-Jensen, J., Colomer-Poveda, D., 2021. Functional relevance of resistance training-induced neuroplasticity in health and disease. *Neuroscience & Biobehavioral Reviews* 122, 79–91. <https://doi.org/10.1016/j.neubiorev.2020.12.019>
- Hortobágyi, T., Rider, P., Gruber, A.H., DeVita, P., 2016. Age and muscle strength mediate the age-related biomechanical plasticity of gait. *Eur J Appl Physiol* 116, 805–814. <https://doi.org/10.1007/s00421-015-3312-8>
- Hortobágyi, T., Vetrovsky, T., Balbim, G.M., Sorte Silva, N.C.B., Manca, A., Deriu, F., Kolmos, M., Kruuse, C., Liu-Ambrose, T., Radák, Z., Vácz, M., Johansson, H., Dos Santos, P.C.R., Franzén, E., Granacher, U., 2022. The impact of aerobic and resistance training intensity on markers of

- neuroplasticity in health and disease. *Ageing Research Reviews* 80, 101698. <https://doi.org/10.1016/j.arr.2022.101698>
- Horwood, A., Chockalingam, N., 2023. *Clinical Biomechanics in Human Locomotion: Gait and Pathomechanical Principles*, 1st ed. Elsevier Science & Technology, San Diego.
- Hosseinzadeh, S., Barzegari, A., Taghipour, M., Mehr Aein, R., Gholinia, H., 2020. Changes of the Patellar Tendon Moment and Length in Different Knee Angles: A Biomechanical in Vivo Study. *Arch Bone Jt Surg* 8, 641–645. <https://doi.org/10.22038/abjs.2020.42551.2158>
- Howells, B.E., Arden, C.L., Webster, K.E., 2011. Is postural control restored following anterior cruciate ligament reconstruction? A systematic review. *Knee Surg Sports Traumatol Arthrosc* 19, 1168–1177. <https://doi.org/10.1007/s00167-011-1444-x>
- Hulleck, A.A., Menoth Mohan, D., Abdallah, N., El Rich, M., Khalaf, K., 2022. Present and future of gait assessment in clinical practice: Towards the application of novel trends and technologies. *Front. Med. Technol.* 4, 901331. <https://doi.org/10.3389/fmedt.2022.901331>
- Iyengar, V., Santos, M.J., Ko, M., Aruin, A.S., 2009. Grip Force Control in Individuals With Multiple Sclerosis. *Neurorehabil Neural Repair* 23, 855–861. <https://doi.org/10.1177/1545968309338194>
- Jarvis, P., Turner, A., Read, P., Bishop, C., 2022. Reactive Strength Index and its Associations with Measures of Physical and Sports Performance: A Systematic Review with Meta-Analysis. *Sports Med* 52, 301–330. <https://doi.org/10.1007/s40279-021-01566-y>
- Kapreli, E., Athanasopoulos, S., Gliatis, J., Papathanasiou, M., Peeters, R., Strimpakos, N., Van Hecke, P., Gouliamos, A., Sunaert, S., 2009. Anterior Cruciate Ligament Deficiency Causes Brain Plasticity: A Functional MRI Study. *Am J Sports Med* 37, 2419–2426. <https://doi.org/10.1177/0363546509343201>
- Knezevic, O.M., Mirkov, D.M., Kadija, M., Milovanovic, D., Jaric, S., 2012. Alternating Consecutive Maximum Contraction as a Test of Muscle Function in Athletes Following ACL Reconstruction. *J Hum Kinet* 35, 5–13. <https://doi.org/10.2478/v10078-012-0074-9>
- Komi, P.V., 2000. Stretch-shortening cycle: a powerful model to study normal and fatigued muscle. *Journal of Biomechanics* 33, 1197–1206. [https://doi.org/10.1016/S0021-9290\(00\)00064-6](https://doi.org/10.1016/S0021-9290(00)00064-6)
- Komi, P.V., Bosco, C., 1978. Utilization of stored elastic energy in leg extensor muscles by men and women. *Med Sci Sports* 10, 261–265.
- Konishi, Y., Aihara, Y., Sakai, M., Ogawa, G., Fukubayashi, T., 2007. Gamma loop dysfunction in the quadriceps femoris of patients who underwent anterior cruciate ligament reconstruction remains bilaterally. *Scandinavian Med Sci Sports* 17, 393–399. <https://doi.org/10.1111/j.1600-0838.2006.00573.x>
- Kouvelioti, V., Kellis, E., Kofotolis, N., Amiridis, I., 2015. Reliability of Single-leg and Double-leg Balance Tests in Subjects with Anterior Cruciate Ligament Reconstruction and Controls. *Research in Sports Medicine* 23, 151–166. <https://doi.org/10.1080/15438627.2015.1005292>
- La Greca, S., Antonacci, G., Marinelli, S., Cifelli, P., Di Giminiani, R., 2024. The acute effect of verbal instructions on performance and landing when dropping from different heights: the ground reaction force-time profile of drop vertical jumps in female volleyball athletes. *Front. Sports Act. Living* 6, 1474537.

- <https://doi.org/10.3389/fspor.2024.1474537>
- La Greca, S., Marinelli, S., Farina, F., Di Giminiani, R., 2026. Postural control and leg muscle activation in athletes who underwent anterior cruciate ligament reconstruction and healthy controls: A preliminary study on acute responses to whole-body vibration. *Clinical Biomechanics*, 137, 106856. <https://doi.org/10.1016/j.clinbiomech.2026.106856>
- La Greca, S., Marinelli, S., Farina, F., Di Giminiani, R., 2025a. An Evaluation of Postural Balance in Single-leg and Double-leg Conditions in Patients with Reconstruction of the Anterior Cruciate Ligament and Healthy Controls. *Acta kinesiol.* 67–79. <https://doi.org/10.51371/issn.1840-2976.2025.19.3.8>
- La Greca, S., Marinelli, S., Totaro, R., Pistoia, F., Di Giminiani, R., 2025b. Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. *Applied Sciences* 15, 8351. <https://doi.org/10.3390/app15158351>
- La Greca, S., Rapali, M., Ciapriani, G., Russo, L., Vinciguerra, M.G., Di Giminiani, R., 2022. Acute and Chronic Effects of Supervised Flexibility Training in Older Adults: A Comparison of Two Different Conditioning Programs. *IJERPH* 19, 16974. <https://doi.org/10.3390/ijerph192416974>
- Latash, M.L., 2008. *Synergy*. Oxford University Press, Oxford; New York.
- Latash, M.L., Singh, T., 2024. *Neurophysiological Basis of Motor Control*, Third edition. ed. Human Kinetics, Champaign.
- Lees, A., 2002. Technique analysis in sports: a critical review. *Journal of Sports Sciences* 20, 813–828. <https://doi.org/10.1080/026404102320675657>
- Lehmann, T., Paschen, L., Baumeister, J., 2017. Single-Leg Assessment of Postural Stability After Anterior Cruciate Ligament Injury: a Systematic Review and Meta-Analysis. *Sports Med - Open* 3, 32. <https://doi.org/10.1186/s40798-017-0100-5>
- Lepley, A.S., Grooms, D.R., Burland, J.P., Davi, S.M., Kinsella-Shaw, J.M., Lepley, L.K., 2019. Quadriceps muscle function following anterior cruciate ligament reconstruction: systemic differences in neural and morphological characteristics. *Exp Brain Res* 237, 1267–1278. <https://doi.org/10.1007/s00221-019-05499-x>
- Lord, C., Rutter, M., DiLavore, P., Risi, S., Gotham, K., Bishop, S., 2012. *Autism diagnostic observation schedule–2nd edition (ADOS-2)*. Los Angeles, CA: Western Psychological Corporation 284, 474–478.
- Luft, A.R., Smith, G.V., Forrester, L., Whittall, J., Macko, R.F., Hauser, T., Goldberg, A.P., Hanley, D.F., 2002. Comparing brain activation associated with isolated upper and lower limb movement across corresponding joints. *Human Brain Mapping* 17, 131–140. <https://doi.org/10.1002/hbm.10058>
- MacKenzie, E.G., Snow, N.J., Chaves, A.R., Reza, S.Z., Ploughman, M., 2024. Weak grip strength among persons with multiple sclerosis having minimal disability is not related to agility or integrity of the corticospinal tract. *Multiple Sclerosis and Related Disorders* 88, 105741. <https://doi.org/10.1016/j.msard.2024.105741>
- Maffiuletti, N.A., Aagaard, P., Blazevich, A.J., Folland, J., Tillin, N., Duchateau, J., 2016. Rate of force development: physiological and methodological considerations. *Eur J Appl Physiol* 116, 1091–1116. <https://doi.org/10.1007/s00421-016-3346-6>
- Mancini, M., Horak, F.B., 2010. The relevance of clinical balance

- assessment tools to differentiate balance deficits. *Eur J Phys Rehabil Med* 46, 239–248.
- Marinelli, S., La Greca, S., Mazzaferro, D., Russo, L., Di Giminiani, R., 2024. A focus on exercise prescription and assessment for a safe return to sport participation following a patellar tendon reconstruction in a soccer player. *Acta kinesiol.* <https://doi.org/10.51371/issn.1840-2976.2024.18.2.3>
- Masedu, F., Angelozzi, M., Di Giminiani, R., Valenti, M., 2013. The use of fractal dimension methods in clinical epidemiology: an application for postural assessment. *EBPH* 10, 9. <https://doi.org/10.2427/8735>
- Mayer, S.W., Queen, R.M., Taylor, D., Moorman, C.T., Toth, A.P., Garrett, W.E., Butler, R.J., 2015. Functional Testing Differences in Anterior Cruciate Ligament Reconstruction Patients Released Versus Not Released to Return to Sport. *Am J Sports Med* 43, 1648–1655. <https://doi.org/10.1177/0363546515578249>
- McGill, S., 2016. Low back disorders: evidence-based prevention and rehabilitation, Third edition. ed. Human Kinetics, Champaign, IL.
- McGinnis, P.M., 2020. Biomechanics of sport and exercise, Fourth edition. ed. Human Kinetics, Champaign, IL.
- McGrath, R., Lang, J.J., Ortega, F.B., Chaput, J.-P., Zhang, K., Smith, J., Vincent, B., Piñero, J.C., Garcia, M.C., Tomkinson, G.R., 2022. Handgrip strength asymmetry is associated with slow gait speed and poorer standing balance in older Americans. *Archives of Gerontology and Geriatrics* 102, 104716. <https://doi.org/10.1016/j.archger.2022.104716>
- Medved, V. (Ed.), 2021. Measurement and Analysis of Human Locomotion, Series in Biomedical Engineering. Springer International Publishing, Cham. <https://doi.org/10.1007/978-3-030-79685-3>
- Meiners, K.M., Loudon, J.K., 2020. Dynamic and Static Assessment of Single-Leg Postural Control in Female Soccer Players. *Journal of Sport Rehabilitation* 29, 174–178. <https://doi.org/10.1123/jsr.2018-0072>
- Merletti, R., Parker, P. (Eds.), 2004. Electromyography: Physiology, Engineering, and Noninvasive Applications, 1st ed. Wiley. <https://doi.org/10.1002/0471678384>
- Mirkov, D.M., Knezevic, O.M., Maffiuletti, N.A., Kadija, M., Nedeljkovic, A., Jaric, S., 2017. Contralateral limb deficit after ACL-reconstruction: an analysis of early and late phase of rate of force development. *Journal of Sports Sciences* 35, 435–440. <https://doi.org/10.1080/02640414.2016.1168933>
- Montoro-Bombú, R., Sarmiento, H., Buzzichelli, C., Moura, N.A., Gonzáles Badillo, J.J., Santos, A., Rama, L., 2023. Methodological considerations for determining the volume and intensity of drop jump training. A systematic, critical and prepositive review. *Front. Physiol.* 14, 1181781. <https://doi.org/10.3389/fphys.2023.1181781>
- Motl, R.W., Cohen, J.A., Benedict, R., Phillips, G., LaRocca, N., Hudson, L.D., Rudick, R., Multiple Sclerosis Outcome Assessments Consortium, 2017. Validity of the timed 25-foot walk as an ambulatory performance outcome measure for multiple sclerosis. *Mult Scler* 23, 704–710. <https://doi.org/10.1177/135245817690823>
- Muybridge, E., 1979. Muybridge's Complete human and animal locomotion: all 781 plates from the 1887 Animal locomotion. Dover Publications, New York.
- Nashner, L.M., 1982. Adaptation of human movement to altered environments. *Trends in Neurosciences* 5, 358–361. [https://doi.org/10.1016/0166-2236\(82\)90204-1](https://doi.org/10.1016/0166-2236(82)90204-1)
- Nashner, L.M., 1977. Fixed patterns of rapid postural responses among leg

- muscles during stance. *Exp Brain Res* 30. <https://doi.org/10.1007/BF00237855>
- Nashner, L.M., McCollum, G., 1985. The organization of human postural movements: A formal basis and experimental synthesis. *Behav Brain Sci* 8, 135–150. <https://doi.org/10.1017/S0140525X00020008>
- Newell, K.M., 1986. Constraints on the Development of Coordination, in: Wade, M.G., Whiting, H.T.A. (Eds.), *Motor Development in Children: Aspects of Coordination and Control*. Springer Netherlands, Dordrecht, pp. 341–360. https://doi.org/10.1007/978-94-009-4460-2_19
- Nicol, C., Avela, J., Komi, P.V., 2006. The Stretch-Shortening Cycle: A Model to Study Naturally Occurring Neuromuscular Fatigue. *Sports Medicine* 36, 977–999. <https://doi.org/10.2165/00007256-200636110-00004>
- Noamani, A., Riahi, N., Vette, A.H., Rouhani, H., 2023. Clinical Static Balance Assessment: A Narrative Review of Traditional and IMU-Based Posturography in Older Adults and Individuals with Incomplete Spinal Cord Injury. *Sensors (Basel)* 23, 8881. <https://doi.org/10.3390/s23218881>
- Noble, J.W., Eng, J.J., Boyd, L.A., 2014. Bilateral motor tasks involve more brain regions and higher neural activation than unilateral tasks: an fMRI study. *Exp Brain Res* 232, 2785–2795. <https://doi.org/10.1007/s00221-014-3963-4>
- Paillard, T., Noé, F., 2015. Techniques and Methods for Testing the Postural Function in Healthy and Pathological Subjects. *BioMed Research International* 2015, 1–15. <https://doi.org/10.1155/2015/891390>
- Pamukoff, D.N., Pietrosimone, B.G., Ryan, E.D., Lee, D.R., Blackburn, J.T., 2017. Quadriceps Function and Hamstrings Co-Activation After Anterior Cruciate Ligament Reconstruction. *Journal of Athletic Training* 52, 422–428. <https://doi.org/10.4085/1062-6050-52.3.05>
- Parkinson, A.O., Apps, C.L., Morris, J.G., Barnett, C.T., Lewis, M.G.C., 2021. The Calculation, Thresholds and Reporting of Inter-Limb Strength Asymmetry: A Systematic Review. *jsportscimed* 594–617. <https://doi.org/10.52082/jssm.2021.594>
- Passacantando, E., Marinelli, S., La Greca, S., Farina, F., Cariati, I., Cifelli, P., Di Giminiani, R., 2025. Drop vertical jump performance and asymmetries in young rhythmic gymnasts. *Sports Biomechanics* 1–17. <https://doi.org/10.1080/14763141.2025.2539705>
- Paterno, M.V., Schmitt, L.C., Ford, K.R., Rauh, M.J., Hewett, T.E., 2013. Altered postural sway persists after anterior cruciate ligament reconstruction and return to sport. *Gait & Posture* 38, 136–140. <https://doi.org/10.1016/j.gaitpost.2012.11.001>
- Pauk, J., Zawadzka, N., Wasilewska, A., Godlewski, P., 2017. GAIT DEVIATIONS IN CHILDREN WITH CLASSIC HIGH-FUNCTIONING AUTISM AND LOW-FUNCTIONING AUTISM. *J. Mech. Med. Biol.* 17, 1750042. <https://doi.org/10.1142/S0219519417500427>
- Payton, C., Burden, A., British Association of Sport and Exercise Sciences (Eds.), 2018. *Biomechanical evaluation of movement in sport and exercise: the British Association of Sport and Exercise Sciences guide*, 2nd edition. ed. Routledge, Taylor & Francis Group, London; New York.
- Petró, B., Papachatzopoulou, A., Kiss, R.M., 2017. Devices and tasks involved in the objective assessment of standing dynamic balancing – A systematic literature review. *PLoS ONE* 12, e0185188.

- <https://doi.org/10.1371/journal.pone.0185188>
- Przysucha, E., Zerpa, C., Vidotto, E., Puskas, L., Goodwina, C., 2024. Postural Sway Characteristics in Static Balance Control of Youth Female Soccer Players. *Acta kinesiol.* 61–68. <https://doi.org/10.51371/issn.1840-2976.2024.18.3.09>
- Rhea, C.K., Kiefer, A.W., Haran, F.J., Glass, S.M., Warren, W.H., 2014. A new measure of the CoP trajectory in postural sway: Dynamics of heading change. *Medical Engineering & Physics* 36, 1473–1479. <https://doi.org/10.1016/j.medengphys.2014.07.021>
- Rizzato, A., Paoli, A., Marcolin, G., 2025. Exploring the relationship between static and dynamic balance performance through the same center-of-pressure parameters. *BMC Sports Sci Med Rehabil* 17, 221. <https://doi.org/10.1186/s13102-025-01251-x>
- Roe, C., Jacobs, C., Hoch, J., Johnson, D.L., Noehren, B., 2022. Test Batteries After Primary Anterior Cruciate Ligament Reconstruction: A Systematic Review. *Sports Health: A Multidisciplinary Approach* 14, 205–215. <https://doi.org/10.1177/19417381211009473>
- Sahoo, P.K., Sahu, M.M., 2021. Quantitative assessment of postural balance in patients with chronic anterior cruciate ligament injury- A controlled study. *Journal of Clinical Orthopaedics and Trauma* 23, 101645. <https://doi.org/10.1016/j.jcot.2021.101645>
- Shumway-Cook, A., Woollacott, M.H., 2007. *Motor control: translating research into clinical practice*, 3rd ed. ed. Lippincott Williams & Wilkins, Philadelphia.
- Sibley, K.M., Beauchamp, M.K., Van Ooteghem, K., Straus, S.E., Jaglal, S.B., 2015. Using the Systems Framework for Postural Control to Analyze the Components of Balance Evaluated in Standardized Balance Measures: A Scoping Review. *Archives of Physical Medicine and Rehabilitation* 96, 122–132.e29. <https://doi.org/10.1016/j.apmr.2014.06.021>
- Slobounov, S., Hallett, M., Cao, C., Newell, K., 2008. Modulation of cortical activity as a result of voluntary postural sway direction: An EEG study. *Neuroscience Letters* 442, 309–313. <https://doi.org/10.1016/j.neulet.2008.07.021>
- Solomonow, M., Baratta, R., Zhou, B.H., Shoji, H., Bose, W., Beck, C., D'Ambrosia, R., 1987. The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *Am J Sports Med* 15, 207–213. <https://doi.org/10.1177/036354658701500302>
- Stensdotter, A.-K., Tengman, E., Häger, C., 2016. Altered postural control strategies in quiet standing more than 20 years after rupture of the anterior cruciate ligament. *Gait & Posture* 46, 98–103. <https://doi.org/10.1016/j.gaitpost.2016.02.020>
- Stergiou, N., Decker, L.M., 2011. Human movement variability, nonlinear dynamics, and pathology: Is there a connection? *Human Movement Science* 30, 869–888. <https://doi.org/10.1016/j.humov.2011.06.002>
- Sutherland, D.H., 2005. The evolution of clinical gait analysis part III – kinetics and energy assessment. *Gait & Posture* 21, 447–461. <https://doi.org/10.1016/j.gaitpost.2004.07.008>
- Travers, B.G., Powell, P.S., Klinger, L.G., Klinger, M.R., 2013. Motor Difficulties in Autism Spectrum Disorder: Linking Symptom Severity and Postural Stability. *J Autism Dev Disord* 43, 1568–1583. <https://doi.org/10.1007/s10803-012-1702-x>
- Vaishya, R., Misra, A., Vaish, A., Ursino, N., D'Ambrosi, R., 2024. Hand grip strength as a proposed new vital sign

- of health: a narrative review of evidences. *J Health Popul Nutr* 43, 7. <https://doi.org/10.1186/s41043-024-00500-y>
- Vernazza-Martin, S., Martin, N., Vernazza, A., Lepellec-Muller, A., Rufo, M., Massion, J., Assaiante, C., 2005. Goal Directed Locomotion and Balance Control in Autistic Children. *J Autism Dev Disord* 35, 91–102. <https://doi.org/10.1007/s10803-004-1037-3>
- Virgile, A., Bishop, C., 2021. A Narrative Review of Limb Dominance: Task Specificity and the Importance of Fitness Testing. *Journal of Strength and Conditioning Research* 35, 846–858. <https://doi.org/10.1519/JSC.00000000000003851>
- Weiss, M.J., Moran, M.F., Parker, M.E., Foley, J.T., 2013. Gait analysis of teenagers and young adults diagnosed with autism and severe verbal communication disorders. *Front. Integr. Neurosci.* 7. <https://doi.org/10.3389/fnint.2013.00033>
- Wiggins, A.J., Grandhi, R.K., Schneider, D.K., Stanfield, D., Webster, K.E., Myer, G.D., 2016. Risk of Secondary Injury in Younger Athletes After Anterior Cruciate Ligament Reconstruction: A Systematic Review and Meta-analysis. *Am J Sports Med* 44, 1861–1876. <https://doi.org/10.1177/0363546515621554>
- Wilson, G.J., Murphy, A.J., 1996. The Use of Isometric Tests of Muscular Function in Athletic Assessment: *Sports Medicine* 22, 19–37. <https://doi.org/10.2165/00007256-199622010-00003>
- Winter, D., 1995. Human balance and posture control during standing and walking. *Gait & Posture* 3, 193–214. [https://doi.org/10.1016/0966-6362\(96\)82849-9](https://doi.org/10.1016/0966-6362(96)82849-9)
- Winter, D.A., 2009. Biomechanics and motor control of human movement, 4th ed. ed. Wiley, Hoboken, N.J.
- Winter, D.A., Prince, F., Frank, J.S., Powell, C., Zabjek, K.F., 1996. Unified theory regarding A/P and M/L balance in quiet stance. *Journal of Neurophysiology* 75, 2334–2343. <https://doi.org/10.1152/jn.1996.75.6.2334>
- Wojtys, E.M., Huston, L.J., 1994. Neuromuscular Performance in Normal and Anterior Cruciate Ligament-Deficient Lower Extremities. *Am J Sports Med* 22, 89–104. <https://doi.org/10.1177/036354659402200116>
- Young, W.B., 2006. Transfer of Strength and Power Training to Sports Performance. *International Journal of Sports Physiology and Performance* 1, 74–83. <https://doi.org/10.1123/ijsp.1.2.74>
- Zatsiorsky, V.M., 2002. Kinetics of human motion. Human Kinetics, Champaign, IL.
- Zeni, J.A., Richards, J.G., Higginson, J.S., 2008. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait & Posture* 27, 710–714. <https://doi.org/10.1016/j.gaitpost.2007.07.007>
- Zult, T., Gokeler, A., Van Raay, J.J.A.M., Brouwer, R.W., Zijdwind, I., Hortobágyi, T., 2017. An anterior cruciate ligament injury does not affect the neuromuscular function of the non-injured leg except for dynamic balance and voluntary quadriceps activation. *Knee Surg Sports Traumatol Arthrosc* 25, 172–183. <https://doi.org/10.1007/s00167-016-4335-3>

List of publications

Di Giminiani, R.; Di Lorenzo, D.; **La Greca, S.**; Russo, L.; Masedu, F.; Totaro, R.; Padua, E. (2022). Angle-Angle Diagrams in the Assessment of Locomotion in Persons with Multiple Sclerosis: A Preliminary Study. *Applied Science*. 12(14), 7223. DOI. 10.3390/app12147223

La Greca, S., Rapali, M., Ciaprini, G., Russo, L., Vinciguerra, M. G., & Di Giminiani, R. (2022). Acute and Chronic Effects of Supervised Flexibility Training in Older Adults: A Comparison of Two Different Conditioning Programs. *International journal of environmental research and public health*. 19(24), 16974. DOI. 10.3390/ijerph192416974

Di Giminiani, R.; Marinelli, S.; **La Greca, S.**; Di Blasio, A.; Angelozzi, M.; Cacchio, A. (2023). Neuromuscular Characteristics of Unilateral and Bilateral Maximal Voluntary Isometric Contractions following ACL Reconstruction. *Biology*. 12(9), 1173. DOI. 10.3390/biology12091173

Marinelli, S.; **La Greca, S.**; Mazzaferro, D.; Russo, L.; Di Giminiani, R. (2024). A focus on exercise prescription and assessment for a safe return to sport participation following a patellar tendon reconstruction in a soccer player. *Acta Kinesiologica*. 18(2), 13-23. DOI. 10.51371/issn.1840-2976.2024.18.2.3

Di Giminiani, R., **La Greca, S.**, Marinelli, S., Attanasio, M., Masedu, F., Mazza, M., Valenti, M. (2024). Locomotion and Postural Control in Young Adults with Autism Spectrum Disorders: A Novel Kinesiological Assessment. *Journal of Functional Morphology and Kinesiology*. 9(4), 185. DOI. 10.3390/jfmk9040185

La Greca, S., Antonacci, G., Marinelli, S., Cifelli, P. and Di Giminiani, R. (2024). The acute effect of verbal instructions on performance and landing when dropping from different heights: the ground reaction force-time profile of drop vertical jumps in female volleyball athletes. *Frontiers in Sports and Active Living*. 6:1474537. doi: 10.3389/fspor.2024.1474537

La Greca, S., Marinelli, S., Totaro, R., Pistoia, F., & Di Giminiani, R. (2025). Effect of Individualized Whole-Body Vibration Exercise on Locomotion and Postural Control in a Person with Multiple Sclerosis: A 5-Year Case Report. *Applied Sciences*, 15(15), 8351. <https://doi.org/10.3390/app15158351>

Passacantando, E., Marinelli, S., **La Greca, S.**, Farina, F., Cariati, I., Cifelli, P., & Di Giminiani, R. (2025). Drop vertical jump performance and asymmetries in young rhythmic gymnasts. *Sports Biomechanics*, 1-17. <https://doi.org/10.1080/14763141.2025.2539705>

S. La Greca, S. Marinelli, F. Farina, R. Di Giminiani (2025). An evaluation of postural balance in single-leg and double-leg conditions in patients with reconstruction of the anterior cruciate ligament and healthy controls. *Acta Kinesiologica*. 19(3), 67-79. DOI. 10.51371/issn.1840-2976.2025.19.3.8

S. La Greca, S. Marinelli, F. Farina, & R. Di Giminiani. (2026). Postural control and leg muscle activation in athletes who underwent anterior cruciate ligament reconstruction and healthy controls: A preliminary study on acute responses to whole-body vibration. *Clinical Biomechanics*, 137, 106856. <https://doi.org/10.1016/j.clinbiomech.2026.106856>

Acknowledgements

I would like to first and foremost thank Professor Di Giminiani for inspiring in me a deep and enduring passion for the world of research, and for giving me the opportunity to pursue this PhD. I owe a great deal to his guidance, trust, and support throughout this work.

I am also sincerely grateful for the friendship and collaboration of my laboratory colleagues, Stefano and Francesco, with whom I share the daily challenges, discussions, and achievements of research.

I would like to thank my partner, Paolina, for her constant presence, understanding, and support in everyday life.

Finally, I wish to express my deepest gratitude to my family for always being close to me. My brothers, Alexandre and Aandrea, and my parents, Florence and Antonino, whose unwavering support has been fundamental throughout my entire journey.