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Kinematic and Muscular Analysis in Clinical and Sport Contexts

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Abstract

Exploring the kinematics of human movement holds significant clinical relevance, offering essential insights into limb functionality under both physiological and pathological conditions, insights that are crucial for accurate diagnosis, effective treatment planning, and rehabilitation monitoring. Traditionally, non-wearable systems have been regarded as the gold standard for motion analysis; however, their high cost, limited accessibility, and dependence on controlled laboratory environments restrict their practical application. In contrast, wearable technologies, such as Inertial Measurement Units (IMU) and surface Electromyography (sEMG), provide portable, non-invasive, and flexible solutions that enable continuous monitoring and the acquisition of objective, quantitative data across a variety of real-world settings. This thesis explores the application of IMU and sEMG in human movement analysis, with a focus on two main situations: (1) the development of an IMU-based protocol for assessing fall risk in the elderly, and (2) the investigation of shoulder kinematics and muscular activity in overhead sports. Through these studies, the work aims to advance the implementation of wearable technologies as reliable tools for objective movement assessment, bridging the gap between laboratory-based research and clinical or athletic practice.

The first part of the thesis presents a novel methodology to make objective the application of the Tinetti Fall Risk Assessment Scale, the most widely used clinical tool for evaluating fall risk in the elderly population. The proposed protocol tries to overcome the main limitations of the clinical scale by adopting the use of sensors and of objective kinematic parameters derived from angular rotations. Validation was initially performed on a healthy young cohort to establish normative reference values, followed by application to an elderly population with varying levels of fall risk. The findings suggest age-related differences in the parameters, in particular in spectral power during standing balance, highlighting measurable distinctions in postural control. Although correlations between kinematic parameters and clinical scores were generally weak, consistent trends emerged in step duration and amplitude during the 360° turning task, suggesting its high sensitivity in capturing gait irregularities. Overall, these findings highlight the potential of wearable sensor technologies to

complement and enhance conventional clinical scales, enabling more objective, reliable, and data-driven evaluations of balance and mobility in older adults.

The second part focuses on shoulder biomechanics across three overhead sports, swimming, climbing, and water polo, where extensive mobility and repetitive overhead movements often lead to joint instability and overuse injuries. Using IMU and sEMG, sport-specific kinematic patterns and neuromuscular activation profiles were quantified, alongside an evaluation of the effects of a physiotherapy intervention. Differences were identified between healthy and pathological athletes, with the rehabilitation protocol demonstrating notable improvements in muscle activation balance, proprioceptive control, and overall movement efficiency. Despite the relatively small sample sizes, the findings validate the applicability of wearable technologies as effective tools for monitoring and supporting targeted physiotherapy, individualised rehabilitation, and injury prevention strategies.

Overall, this thesis contributes to the growing body of evidence advocating the validity of wearable IMU and sEMG technologies in both clinical and sports contexts. By enabling objective, quantitative, and context-specific analysis of human movement, these systems bridge the gap between laboratory research and practical application, paving the way toward data-driven healthcare, personalised rehabilitation, and performance optimisation.

Sommario

Lo studio della cinematica del movimento umano ha assunto una rilevanza clinica significativa poiché consente di fornire informazioni sugli arti sia in condizioni fisiologiche sia patologiche; informazioni fondamentali per una diagnosi accurata, una pianificazione terapeutica efficace e un monitoraggio oggettivo della riabilitazione. Tradizionalmente i sistemi "non wearable" sono stati considerati il gold standard per l'analisi del movimento; tuttavia, il loro elevato costo, la limitata accessibilità e la necessità di ambienti di laboratorio controllati, ne riducono l'applicabilità pratica. Al contrario, le tecnologie "wearable", come le unità di misura inerziali (IMU) e l'elettromiografia di superficie (sEMG), offrono soluzioni portatili, non invasive e flessibili, che permettono un monitoraggio continuo e la raccolta di dati quantitativi e oggettivi in una vasta gamma di contesti.

La presente tesi esplora l'applicazione di IMU e sEMG nell'analisi del movimento focalizzandosi su due ambiti principali: (1) lo sviluppo di un protocollo basato su IMU per la valutazione del rischio di caduta nella popolazione anziana, e (2) lo studio della cinematica e dell'attività muscolare della spalla negli sport "overhead". Attraverso questi studi, il lavoro ha lo scopo di promuovere l'impiego di tecnologie "wearable" come strumenti affidabili per una valutazione oggettiva del movimento, colmando così il divario tra la ricerca in laboratorio e la pratica clinica o sportiva.

La prima parte della tesi presenta una nuova metodologia per oggettivare l'applicazione della scala del rischio di caduta Tinetti, lo strumento clinico maggiormente diffuso per la valutazione del rischio di caduta nella popolazione anziana. Il protocollo proposto mira a superare le principali limitazioni della scala tradizionale attraverso l'impiego di sensori e parametri cinematici oggettivi derivati dalle rotazioni angolari. La validazione è stata inizialmente condotta su un gruppo di giovani adulti sani, al fine di stabilire valori di riferimento normativi, per poi essere applicata a una popolazione anziana con differenti livelli di rischio di caduta. I risultati evidenziano differenze legate all'età nei parametri analizzati, in particolare nella potenza spettrale durante il mantenimento dell'equilibrio, sottolineando differenze misurabili nel controllo posturale. Sebbene le correlazioni tra parametri cinematici e punteggi clinici risultino in generale deboli, sono emerse tendenze coerenti nella

durata e nell’ampiezza del passo durante la prova di rotazione a 360°, suggerendo un’elevata sensibilità di tale compito nel rilevare irregolarità del cammino. Complessivamente, i risultati ottenuti evidenziano il potenziale delle tecnologie indossabili nel completare e migliorare le scale cliniche convenzionali, consentendo valutazioni più oggettive, affidabili e basate su dati quantitativi dell’equilibrio e della mobilità negli anziani.

La seconda parte si concentra sulla biomeccanica della spalla in tre sport overhead, nuoto, arrampicata e pallanuoto, nei quali l’ampia mobilità articolare e i movimenti ripetitivi sopra la testa possono determinare instabilità articolare e lesioni da sovraccarico. Mediante l’utilizzo combinato di IMU e sEMG sono stati quantificati i pattern cinematici e i profili di attivazione neuromuscolare specifici per ciascuna disciplina, insieme alla valutazione degli effetti di un intervento fisioterapico mirato. Sono state identificate differenze tra atleti sani e atleti con patologie della spalla, con il protocollo riabilitativo che ha evidenziato miglioramenti significativi nell’equilibrio di attivazione muscolare, nel controllo propriocettivo e nell’efficienza complessiva del movimento. Nonostante le dimensioni relativamente ridotte dei campioni, i risultati confermano l’applicabilità delle tecnologie indossabili come strumenti efficaci per il monitoraggio e il supporto della fisioterapia mirata, della riabilitazione personalizzata e delle strategie di prevenzione degli infortuni.

Nel complesso, la tesi contribuisce a supportare la validità delle tecnologie wearable basate su sensori IMU ed EMG superficiale in contesto clinico e sportivo. A partire da un’analisi oggettiva, quantitativa e specifica del movimento umano, questi sistemi colmano il divario tra ricerca di laboratorio e applicazione pratica, aprendo la strada verso una sanità basata sui dati, una riabilitazione personalizzata e un’ottimizzazione delle prestazioni motorie.

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List of abbreviations

ANOVA	Mixed-factors Analysis of Variance
ApEn	Approximate Entropy
CorrDim	Correlation Dimension
CV	Coefficient of Variation
Dm	Radial Displacement
ER	External Rotation
FD	Fractal Dimension
HR	Heart Rate
IMU	Inertial Measurement Units
IR	Internal Rotation
Lyap	Largest Lyapunov Exponent
m.LD	Latissimus Dorsi
m.LT	Lower Trapezius
m.MD	Middle Deltoid
m.MT	Middle Trapezius
m.PM	Pectoralis Major
m.S	Scalene
m.SA	Serratus Anterior
m.SCM	Sternocleidomastoid
m.UT	Upper Trapezius
MVC	Maximum Voluntary Contraction
MVIC	Maximum Voluntary Isometric Contraction
NRS	Numeric Rating Scale
POMA	Performance Oriented Mobility Assessment
PSD	Power Spectral Density
ROA	Romberg Open Eyes
ROC	Romberg Closed Eyes
RMS	Root-Mean-Square
RPE	Rate of Physical Exertion
sEMG	Surface Electromyography
T_c	Contraction Time
T_d	Delay Time
TMG	Tensiomyography
T_r	Half-relaxation Time
T_s	Sustain Time
Var	Variance

Chapter 1. INTRODUCTION

1.1 Preface

Exploring the kinematics of human movement is clinically significant and fundamental for assessing limb functionality under both normal and pathological conditions, serving as a critical component in accurate diagnosis and effective treatment planning [1].

Various non-invasive methodologies have been developed to objectively analyse kinematics by means of systems, broadly classified as wearable and non-wearable. While non-wearable systems are regarded as the gold standard, they exhibit several limitations that wearable systems are designed to overcome [2], [3]. In fact, owing to their portability, the wearable systems facilitate data acquisition across various settings, including hospitals, outpatient clinics, and gyms [4], and devices such as IMU are increasingly employed to capture human movement and provide objective, quantitative data. Moreover, wearable systems enable the monitoring of rehabilitation and the management of neurological and musculoskeletal disorders.

A direct application of wearable systems is the assessment of fall risk in the elderly population. Several studies have demonstrated the potential of these technologies to detect critical conditions by analysing kinematic parameters extracted directly from sensor signals [5], [6], [7], [8]. Until now, fall risk among older adults has been primarily evaluated using clinical scales, typically administered in ambulatory or long-term care settings. These assessments typically consist of physical performance tests and/or questionnaires addressing daily living activities. The Tinetti Mobility Test, also known as the Performance-Oriented Mobility Assessment (POMA), is the most used test, it comprises two sections evaluating gait and balance [9]. Transforming its inherently visual and subjective assessments into objective, quantifiable parameters could address two major limitations: the reliance on healthcare professionals, which introduces inter-operator variability in scoring, and the restricted accessibility of these assessments in clinical and hospital settings.

Integrating the analysis of human movement kinematics through IMU with muscular activity offers valuable insights, providing a more comprehensive understanding of underlying mechanisms in specific conditions. In particular, the sEMG is a non-invasive tool that allows for the registration of the electrical activity of muscles [10]. The combination of IMU and sEMG is widely employed in sports settings to enhance performance and detect altered movement patterns that may lead to injury [11], [12]. This integrated approach is particularly relevant for the upper limb, where complex neuromuscular coordination and dynamic joint interactions play a critical role in functional stability and performance. In particular, the shoulder joint is the most mobile in the body, and its wide mobility results in reduced joint stability, making the shoulder more susceptible to injury and dislocations. Shoulder injuries are widespread, especially among athletes who perform overhead movements with high intensities and extreme ranges of motion, such as those in swimming, climbing, and water polo [13], [14].

This thesis examines the application of IMU and sEMG tools, as well as their signal processing, to develop innovative protocols for analysing human movement. This work aims to contribute to the ongoing advancement and broader adoption of such systems in clinical practice. Ultimately, the research results support the transition from traditional observational assessments to objective, data-driven approaches that enhance diagnosis, treatment planning, and rehabilitation outcomes in musculoskeletal and neurological disorders.

1.2 Thesis objectives

The present PhD thesis is structured into two main parts. The first part (Chapter 2) presents the development and application of a protocol designed to analyse and assess the tasks of the Tinetti Fall Risk Assessment Scale using IMU. The second part (Chapter 3) focuses on three studies investigating the kinematic patterns and muscular activity of the shoulder joint across three overhead sports, providing insights into sport-specific movement strategies and potential injury mechanisms.

1.2.1 Evaluation of fall risk through sensor-based protocol

According to the World Health Organisation, 28–35% of individuals aged 65 and older experience a fall each year, with this rate rising to 32–42% among those aged 70 and above. As a result, the demand for hospitalisation increases significantly, placing considerable physical, psychological, and financial burdens on both patients and healthcare systems. Furthermore, due to age-related decline and increased frailty, many elderly individuals who suffer fall-related injuries remain hospitalised for the remainder of their lives [15].

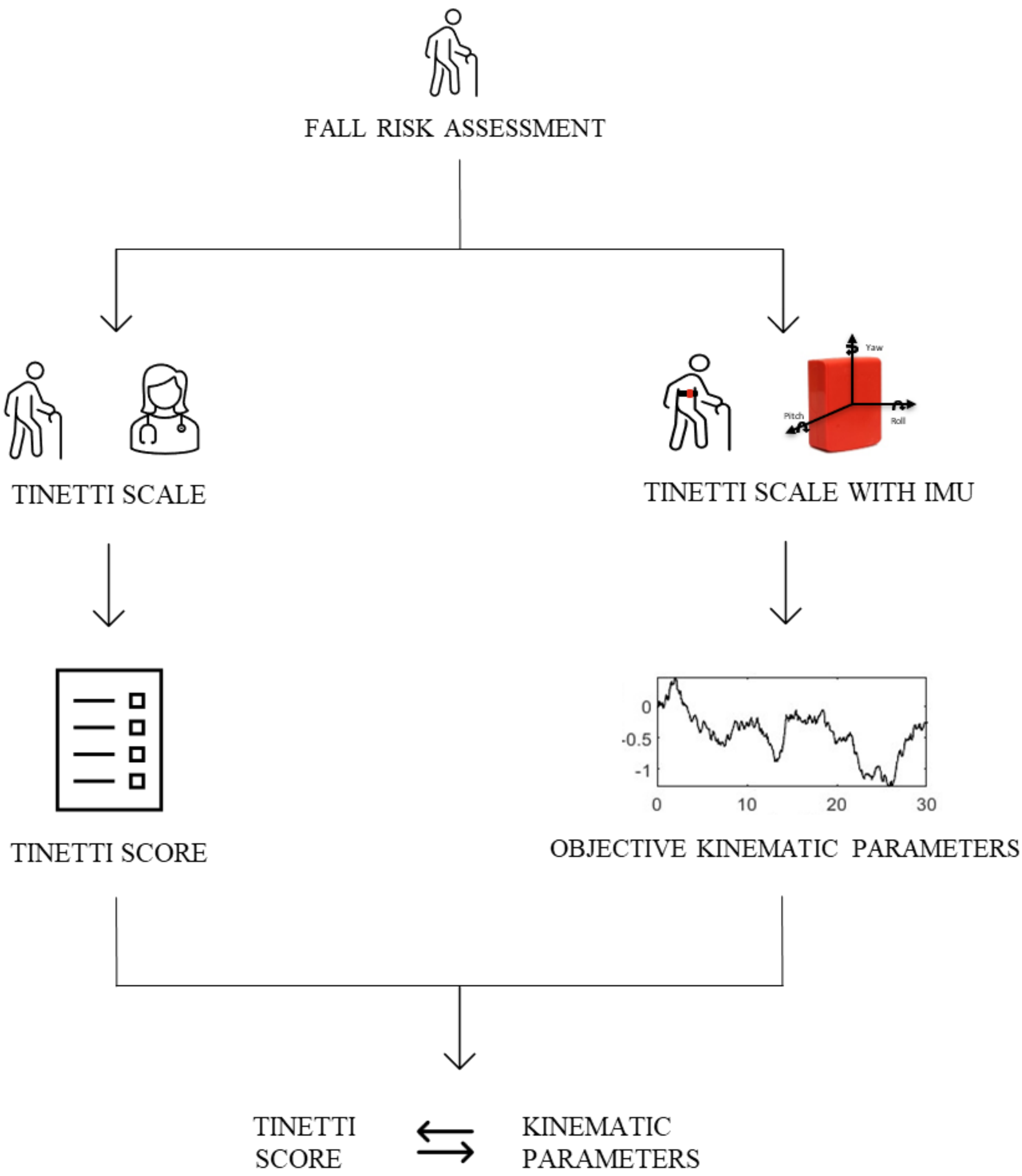
The use of clinical scales represents a valid and widely accepted approach for assessing fall risk. However, despite their ease of administration, these tools have notable limitations, particularly in terms of subjectivity and sensitivity. Recent literature [16], [17], [18] highlights the potential of wearable devices, such as accelerometers and gyroscopes, as a promising alternative. These technologies enable the collection of objective, quantitative data without the need for continuous supervision by specialised personnel. [6], [7]. Capturing movement data in terms of joint angles, acceleration, and velocity enables the extraction of quantitative parameters that could be correlated with exercise performance scores, thereby facilitating the identification of individuals at high or low risk of falling.

1.2.2 A novel approach to studying the shoulder joint in overhead sports

Shoulder dysfunctions rank as the third most common musculoskeletal disorder and are a leading cause of musculoskeletal pain and disability. These dysfunctions can range from well-defined conditions with clear diagnostic criteria and pathophysiology to vague, hard-to-diagnose issues lacking a precise clinical definition [19], [20], [21]. This condition is particularly common in sports that heavily involve the upper extremities, due to the demanding mechanics associated with rapid shoulder elevation, abduction, and external rotation. For instance, shoulder injuries account for approximately 12% to 19% of all injuries in baseball, and between 23% and 38% in swimming, within a single year [14], [22].

To properly understand the kinematics and muscular activity of the shoulder joint during typical overhead movements, such as those performed in sports like swimming, climbing, and water polo, researchers are increasingly turning to portable systems, including IMU and sEMG which allow to capture high-resolution data on joint angles, accelerations, and muscle activation patterns in real-world or sport-specific environments. This, in turn, supports the development of more targeted and individualised protocols for injury prevention, performance optimisation, and rehabilitation. By objectively identifying dysfunctional movement patterns or asymmetries early on, these tools can aid clinicians and coaches in designing interventions that reduce the risk of overuse injuries and enhance shoulder stability. As such, wearable technology is not only advancing the field of sports science but also contributing to a broader shift toward personalised, data-driven approaches in clinical and athletic settings.

Chapter 2. EVALUATION OF FALL RISK THROUGH SENSOR-BASED PROTOCOL



2.1 Introduction

The risk of falling is defined as a reduced ability to maintain balance while standing, walking, or sitting. Falls and their resulting injuries represent a major public health concern, frequently requiring medical intervention. They are responsible for 20–30% of mild to severe injuries and account for approximately 10–15% of all emergency department visits [14]. The most common causes of fall-related hospital admissions include hip fractures, traumatic brain injuries, and upper limb injuries. As individuals age and experience increased frailty, the likelihood of long-term hospitalisation following a fall rises significantly, and approximately 20% of individuals who sustain a hip fracture die within one year of the incident [15].

Beyond the physical consequences, falls often lead to post-fall syndrome, a condition characterised by dependence, reduced autonomy, confusion, immobilisation, and depression, which further limits an individual's ability to perform daily activities.

Age-related biological changes contribute to an exponential rise in fall risk, particularly in individuals over the age of 80. Without the implementation of effective prevention strategies, the number of fall-related injuries is expected to double by 2030 [7], [16]. Thus, addressing these evolving needs requires the adoption of innovative strategies within the global healthcare system, establishing a framework that prioritises preventive measures to avoid the worsening of conditions that could have been easily treated in their early stages.

In response to this growing issue, a variety of clinical scales have been developed to assess fall risk, particularly in geriatric care settings, as well as for individuals with gait and balance impairments. In particular, for assessing fall risk, the literature proposes different scales such as Timed Up and Go test (to assess functional mobility and dynamic balance), Berg Balance Scale (to determine balance performance), Functional Reach Test (to evaluate dynamic balance performance), and Tinetti Mobility Test (to assess gait and balance performance) [5], [23]. These tools have proven to be flexible enough to evaluate not only the healthy elderly population, but also patients with different underlying pathological conditions, like, for example, Parkinson’s disease [23], patients receiving hemodialysis [24], and systemic sclerosis [25].

However, it is important to note that these tools are not designed to prevent falls directly; rather, they serve to identify individuals at high risk and guide appropriate intervention strategies [17], [18].

While clinical scales are quick and easy to administer, they rely heavily on the presence of trained healthcare professionals and are subject to evaluator bias, which limits their objectivity; thus, it is necessary to implement a system independent of the healthcare provider. Furthermore, wearable technologies offer a promising solution by enabling the precise, objective measurement of mobility parameters. These devices enhance the accuracy of fall risk assessments and support the development of more targeted, data-driven prevention strategies [18].

The research progressed through three stages. First, several Tinetti tasks were characterised in a young, healthy population (Section 2.4) to establish a reference benchmark for subsequent analyses. Next, in Section 2.5, an elderly population was compared with the young group to identify statistically significant differences in the extracted parameters and to understand the biomechanical changes associated with ageing. Finally, a study focusing exclusively on the elderly population was conducted to investigate correlations between kinematic parameters and Tinetti scores, with the aim of determining which parameters best discriminate between different fall risk conditions (high versus low).

2.2 Fall risk evaluation: clinical scales

Clinical scales are widely used tools for assessing fall risk, particularly in geriatric settings involving older adults (aged 65 and above) and individuals with conditions that impair gait and balance. Generally, clinical scales involve questionnaires or functional assessments of gait, balance, and risk factors [26] in order to classify people at low, moderate, or high fall risk.

The Conley Scale, the Tinetti Performance-Oriented Mobility Assessment, the Stratify Scale, the Berg Balance Scale, and the Timed Up and Go test are the most commonly employed assessment tools.

The Stratify scale is a clinical tool used to assess fall risk in hospitalised patients. It consists of five items. When patients do not present cognitive deficits, the first item is answered by the patient, and the remaining items are completed by the nurse. The scale evaluates factors such as history of falls, psychological status, motor function, and other relevant body functions. The maximum total score is 5, with a score of 2 or higher indicating a high risk of falling. Owing to its high sensitivity, as well as its simplicity and speed of administration, the Stratify scale is considered the gold standard for fall-risk assessment at the time of patient admission [5], [27].

The Conley scale aims to assess the risk of falls, and it is composed of two sections of 3 questions each to investigate previous falls and cognitive impairments. This scale is generally administered upon admission of the patient to the ward. Each response is assigned a score, and the cumulative sum determines the overall risk level: a total score between 0 and 1 indicates low risk, whereas a score between 2 and 10 corresponds to a higher risk level [28].

The Berg balance scale is composed of 14 items to evaluate simple movement of everyday life as the ability to maintain or change position and execute tasks in unstable positions. Each task is scored from 0 to 4 according to the quality of the performance. The total score, obtained by summing the individual items, reflects overall balance ability: lower scores indicate a higher risk of falling. Specifically, a total score below 45 out of 56 is associated with an increased risk of falls [29].

The Timed up and go test evaluates mobility, balance, and fall risk in the elderly population. The test assesses the time spent to stand up from a chair, walk for 3 meters, turn 360 degrees, walk 3 meters in the chair direction, and sit down. The total time defines the level of fall risk [30].

2.2.1 Tinetti scale for fall risk assessment

The Tinetti test, also called POMA, was developed by Mary Tinetti [9]. This scale, which can be utilised only by trained healthcare professionals, is composed of a series of scored exercises aimed at assessing balance and gait. In the first part, focused on balance (ranged from 0 to 16 points), the patients are asked to sit down, arise from the chair, stand first with eyes open, then with eyes closed, withstand a nudge on the chest, turn 360°, and sit down. While the second part, targeted on gait (ranged from 0 to 12 points), consists of letting the patient walk and observing step parameters (i.e. step height, step length, continuity symmetry, walking stance, amount of trunk sway, path deviation). The sum of the scores obtained in each of the two parts defines the overall numeric evaluation of fall risk: if below 20, the fall risk is high; otherwise, the risk is classified as low [31].

2.3 Fall risk evaluation through wearable devices

In order to make this scale available to a growing number of older people outside the hospital, increasing the accessibility of the test, and reducing the high inter-operator bias in scoring the exercises, it is necessary to implement a system independent of the healthcare provider. The use of specific devices, such as wearable accelerometers/gyroscopes, can be a solution to this problem, thanks to the possibility of gathering a large amount of data related to movements produced during the fall risk evaluation test [7]. The literature reports other studies about the use of inertial sensors for fall risk assessment, which, however, show great variability in the results, mainly due to the placement of the sensors, the features extracted from them, and the test used [6].

Wearable sensors are a practical technology that can be easily used in clinical, laboratory settings and in common point-of-care environments. Moreover, they are portable, low-cost, and allow the monitoring of a great range of movements [32], [33]. These devices gain information about angular velocity, acceleration of body segments, from which other kinematic parameters can be computed. Literature reports other attempts to use inertial sensors to monitor human activity in order to detect falls [34], [35], [36], [37], [38], [39].

Moreover, clinical scale tasks are typically performed in supervised conditions; therefore, the patient's behaviour is not representative of daily living in a real-world environment [40]. As a result, there is a growing need to evaluate fall risk in uncontrolled, everyday environments. Wearable sensors address this gap by enabling continuous monitoring during routine functional activities, offering a more realistic representation of an individual's fall risk, and facilitating the timely detection of fall-related events [41], [42].

2.4 Parameters extraction from wearable devices

In this study, signals were collected using IMU sensors that measure three-dimensional Euler angles. Euler angles are used to represent the orientation of a rigid body in three-dimensional space through a sequence of three successive rotations about specified axes. The interpretation of these angles depends on the chosen rotation convention, which defines both the order of rotations and whether they are intrinsic (about the body's moving axes) or extrinsic (about the fixed reference axes) [43]. A known limitation of this representation is the presence of gimbal lock, a kinematic singularity in which two rotation axes become aligned, and one degree of freedom is lost. Furthermore, Euler-angle representations are convention-dependent and non-unique, as the same orientation may correspond to different angle triplets depending on the chosen rotation order [44], [45].

From the Euler angles, several parameters were extracted for the different exercises of the Tinetti scale. For the arising from a chair and sitting down exercise, starting from the trunk orientation, the duration of the rotational movement, the maximum angular amplitude along the three axes, and the final angular position were analysed. These parameters provide information on the degree of forward trunk flexion during and at the end of the movement, as well as on motor control [46], [47].

For the balance exercises, nonlinear parameters were selected, including fractal dimension (FD), correlation dimension (CorrDim), largest Lyapunov exponent (Lyap), and approximate entropy (ApEn), to quantify postural oscillations and to identify differences within the population [48], [49]. In addition, power spectral density (PSD) and variance (Var), similarly to the other features, were used to characterise the oscillatory behaviour and the underlying postural control strategies [50].

Finally, for the gait exercises, spatiotemporal parameters were extracted, specifically step duration, step amplitude, and step velocity, as these tasks require dynamic balance and continuous adjustment of the base of support. These parameters provide insight into temporal stability, movement efficiency, and compensatory strategies employed by the participant [51], [52].

2.5 Kinematic characterisation of movements in young subjects during the Tinetti test [53]

2.5.1 Aim of the study

The use of new technologies leads to the need for new standard data to refer to. The aim of this study was to evaluate the possibility of automatic extraction of one or more parameters from a magnetic-inertial sensor that could be directly correlated with the execution of exercises included in the Tinetti test, thus supporting non-expert operators to assess the falling risk. This preliminary study can be used as a benchmark for future evaluation of individuals at risk of falling.

2.5.2 Materials and Methods

2.5.2.1 Subjects and Tasks

A group of 30 healthy young subjects, 15 males and 15 females (aged 21-31, mean 22.7 ± 2.3 years), participated to the study. The subjects, presenting a Tinetti score of 28, performed four different exercises of the Tinetti test: arising from a chair (1), standing balance with open eyes (2) and with closed eyes (3), and sitting down (4). In order to calibrate the system, before starting the first and the second task, the subject was asked to maintain the resting position for few seconds, sitting with the back leaning against the backrest or standing with arms along sides, respectively. All the enrolled subjects signed the informed consent.

2.5.2.2 Acquisition and Analysis

For the data acquisition, a wireless magneto-inertial sensor from the MTw Awinda Development Kit (Xsens Technology) capable of detecting 3D Euler angles was used. The sensor was placed on the chest of the subject by means of Velcro body strap so that pitch, roll, and yaw angles were the rotations around respectively anteroposterior, right-left, and top-down directions (Figure 1).

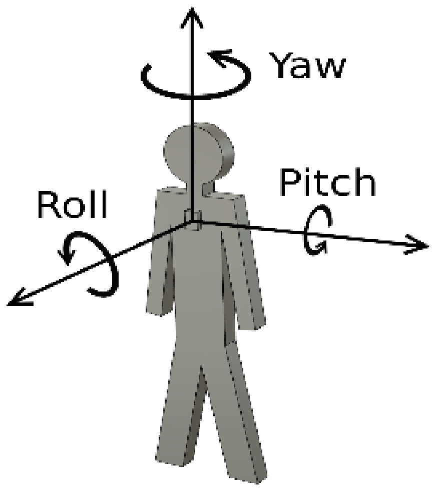


Figure 1. Schematic representation of Pitch, Roll, and Yaw angles.

A specific software (MT Manager) allows the operator to link the devices to the host and to acquire the data at a sample frequency of 100 Hz. The collected data were processed using a program written in MATLAB®.

For the exercises (1) and (4), the evaluated parameters were the duration of the rotation movement and the maximum angular amplitude along the three directions, while for (2) and (3), the parameters evaluated were the FD, calculated by the Higuchi algorithm [54], and the total spectral power extracted from the PSD estimated by using Welch's method. These values were averaged over all subjects, and the coefficient of variation (CV) was evaluated to estimate the intersubject variability.

2.5.3 Results

Figure 2 shows the trends of the median curves of roll, pitch, and yaw angles together with the twenty-fifth and seventy-fifth percentiles during the execution of exercises (1) and (4). Since the direction of rotation (right/left, clockwise/counterclockwise) is not relevant for the evaluation of the exercise, the angles measured by roll and yaw are referred to their absolute values. The roll and yaw angle graphs have been time-aligned to the pitch angles by setting the instant corresponding to the maximum pitch angle to zero.

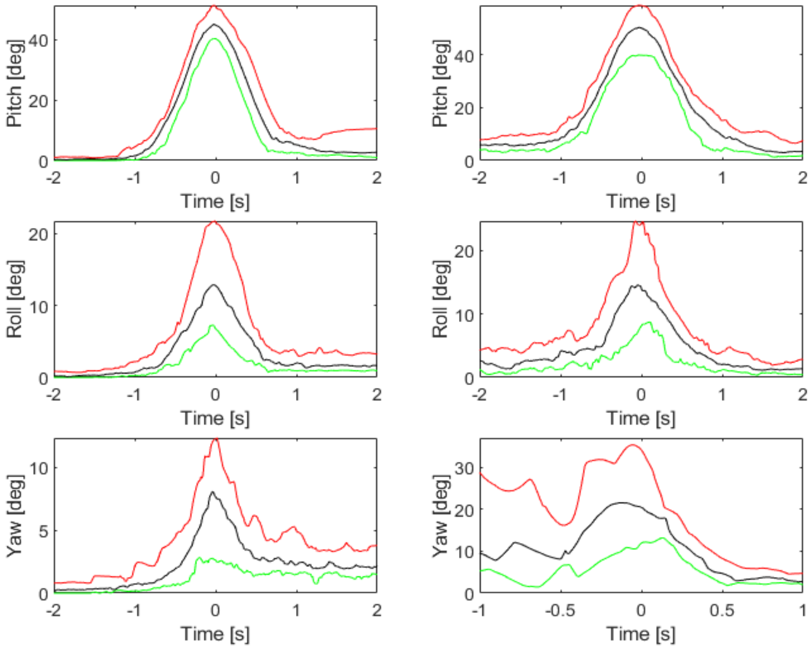


Figure 2. Trends of median (black), 25th (green), and 75th (red) percentile of absolute values of roll, pitch, and yaw for exercise (1), left side, and exercise (4), right side.

As expected, the rotation along the anteroposterior direction (pitch) presents, in both exercises, a very large peak value (till 45° and about 50° for the exercises (1) and (4), respectively), much greater than those in the other directions. Moreover, Table 1 shows a quite low inter-subject variability for both peak and duration values measured by mean of CV in the exercises (1) and (4).

Figure 3 shows a typical angle rotation behavior around the three directions during the exercises (2) and (3) recorded in a subject. The amplitude is very small (always under 1°-7°) in all the directions, with some fluctuations and values similar in the two situations.

Table 1. Median, 25th, 75th percentile, and relative standard deviation (CV) of Peak and Duration values during exercise (1) and exercise (2).

		Pitch		Roll		Yaw	
		Peak [deg]	Duration [s]	Peak [deg]	Duration [s]	Peak [deg]	Duration [s]
Exercise (1)	25th percentile	40.4	2.1	2.3	0.1	1.9	0.1
	Median	45.0	2.4	8.1	1.1	4.4	0.5
	75th Percentile	51.0	2.6	15.7	2.2	11.0	1.2
	CV	0.27	0.18	1.16	0.93	1.04	1.14
Exercise (4)	25th percentile	40.0	2.1	5.0	0.0	28.7	0.89
	Median	50.2	2.5	11.5	1.1	16.5	1.1
	75th Percentile	58.3	2.7	16.3	1.9	12.7	1.5
	CV	0.34	0.25	1.08	1.06	0.48	0.72

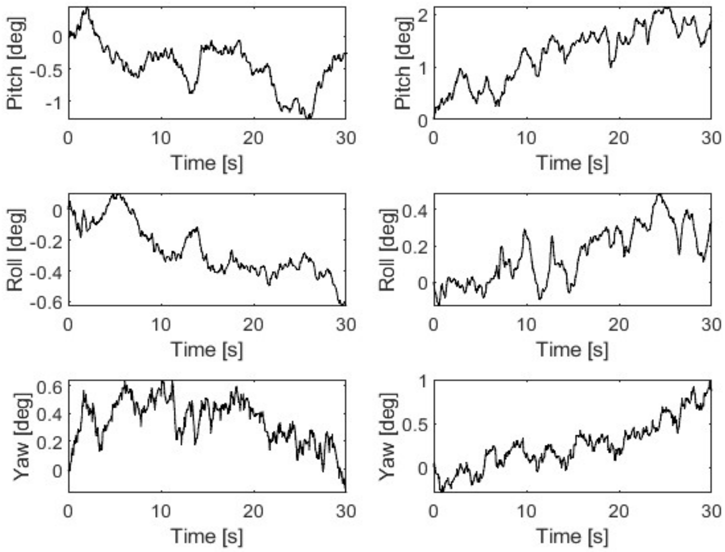


Figure 3. Example of angles of rotation during the Open (left side) and Closed Eyes (right side) exercises recorded in a subject.

Table 2. Median, 25th, and 75th percentile, CV of PSD and FD parameters in the balance exercises with open or closed eyes.

		Pitch		Roll		Yaw	
		SP [deg^2]	FD [-]	SP [deg^2]	FD [-]	SP [deg^2]	FD [-]
Exercise (2)	25th percentile	0.11	1.08	0.03	1.06	0.04	1.07
	Median	0.18	1.14	0.05	1.10	0.12	1.10
	75th Percentile	0.35	1.21	0.12	1.15	0.26	1.16
	CV	1.42	0.09	2.06	0.05	1.71	0.07
Exercise (3)	25th percentile	0.11	1.07	0.04	1.06	0.04	1.09
	Median	0.18	1.11	0.05	1.10	0.09	1.12
	75th Percentile	0.29	1.19	0.10	1.14	0.13	1.17
	CV	0.76	0.08	1.49	0.05	1.32	0.07

Table 1 reports median, 25th percentile, 75th percentile, and relative standard deviation (CV) values of spectral power and FD calculated in balance exercises with open eyes (exercise 2) and closed eyes (exercise 3). As in the exercises (1) and (4), also in these tests, the greatest median spectral power as well as FD values are shown by pitch, followed by yaw and roll. The CV presented a large relative variability for spectral power, especially in the exercise with open eyes. On the contrary, the CV of the FD of roll, pitch, and yaw is very low (0.05-0.09) in both exercises.

2.5.4 Discussion

According to the CV values in Table 1 and Table 2, all the parameters evaluated (peak and duration of rotation for exercise (1) and (4), PSD and FD for exercise (2) and (3)) present a low variability and comparable responses among subjects, demonstrating high repeatability of the exercises in young people.

The results show that in exercises (1) and (4), the prevailing rotation was, as expected, in the anteroposterior direction (pitch angle) with a peak of 45.0° and a duration of 2.4s for (1), and 50.2° and 2.5s for (2), respectively. Roll and yaw presented a lower angular rotation and duration, as shown in Table 1, confirming that in both sitting down and standing up movements, the displacement of the trunk laterally or around the vertical axis is small compared to the anteroposterior one.

We would compare these results to those recently reported by Rivolta et al. [7] that used a triaxial accelerometer to register the linear acceleration along vertical, medio-lateral, and anterior-posterior axes from which amplitude and duration of the movement to execute some Tinetti exercise were evaluated. The sitting down and standing up exercises showed the highest amplitude variation, along the anterior-posterior axis, with a duration, respectively of 2.31s and 1.47s for high-risk subjects and of 1.95s and 1.30s for low-risk subjects. In our study, the movement duration seems much longer than that of Rivolta et al., even in subjects considered to be at high risk. However, we must consider the fact that in our case, the durations are established on the basis of the profiles of the angular rotations which present different amplitudes according to the three rotation axes (Table 1). In the case of Rivolta et al. instead, the durations are calculated from a single acceleration profile that can therefore justify the differences found.

A particular note concerns the yaw angle behavior during exercise (4) as depicted in Figure 2: it is evident that some subjects began the sitting movement turning around its vertical axis before the operator gave the command to sit, so influencing the initial trait of the signal. However, the amplitude of the movement was correctly estimated, discarding the first part of the movement.

The results related to spectral power and FD evaluated for exercise (2) and (3) show that spectral power of pitch and roll is the same for both exercises (0.18 pitch, 0.05 roll), while for yaw there is a slight difference (0.12 open eyes, 0.09 closed eyes), presenting the exercise with open eyes a higher value. The FD presents, for all the angles in both conditions, a median value slightly greater than 1 (between 1.10 and 1.14). Since FD is a measure of the complexity of a time series with value between 1 (smooth fluctuations like those recorded in our subjects) and 2 (abrupt fluctuations), it could be compared with sample entropy (another parameter utilized to measure complexity) that was used in [7]. In the latter paper, during standing balance tasks the sample entropy was normalized to the theoretical sample entropy value of a white Gaussian noise with the same variance and the values obtained for open eyes and closed eyes task were of 0.975 (high risk subjects) and 0.992 (low risk subjects), 0.985 (high risk subjects) and 0.994 (low risk subjects), respectively. Since greater values can be associated with greater signal complexity, we expect that in subjects with a greater risk of falling, the value of the FD will increase as well as entropy.

In conclusion, the results show that some parameters can well characterize the movement during the exercises in a population of young healthy subjects. In particular, the peak and the duration of movement in the anterior-posterior direction for exercise (1) and exercise (4), and the spectral power and the FD for exercise (2) and (3) seem to be good candidates showing a low variability among subjects. Moreover, since our system is based on a 3D magneto-inertial sensor, it allows to detect in a precise and accurate way the angular rotations, distinguishing the movement components along the three orthogonal directions. However, this is a preliminary study that can be used in the future as a benchmark for evaluations of individuals at risk of falling. The next step will concern the application of the same protocol to elderly people.

2.6 Comparison of kinematic parameters between young and elderly subjects [55]

2.6.1 Aim of the study

This paper is an extended study of [56], in which only young subjects were considered, with the aim to compare the kinematic characterization of movement between young and elderly groups during the execution of three tasks belonging to the Tinetti scale by using a wireless magneto-inertial sensor.

2.6.2 Materials and Methods

2.6.2.1 Subjects and Tasks

52 participants were enrolled in the study, 28 healthy young people (15 males, aged 22.7 ± 2.3 years) and 24 elderly people (10 males, aged 78.2 ± 8.9 years) from the Geriatric Department of the Trieste Hospital. Elderly subjects included had the age of 65 or older and were able to execute the tasks. All the subjects performed three different tasks of the Tinetti scale: (1) arising from a chair, standing balance with open (2) and closed (3) eyes in Romberg position [57]. The protocol provided for two calibration phases: before starting the first task sitting with the back leaning against the backrest, and before the two balance tasks standing with arms along sides. According to the Tinetti score, which was assigned by trained personnel, the elderly subjects, following the protocol used in [58], were divided into two groups: High Risk of falling (9 subjects, 3 males, Tinetti score ≤ 18) and Low Risk of falling (15 subjects, 7 males, Tinetti score > 18).

All the participants signed informed consent before being enrolled in the study.

2.6.2.2 Acquisition and Analysis

The data were acquired by means of a wireless magneto-inertial sensor from MTw Awinda Development Kit (Xsens Technology) placed on the chest of the subjects by means of Velcro body strap (Figure 4). The sensor registered the 3D Euler angles, pitch, roll and yaw, representing the rotation around anteroposterior, right-left and top-down directions. A specific software (MT Manager), linked to the device, enabled the operator to gather data at a sampling frequency of 100 Hz. For task (1), we evaluated the duration of the movement and the maximum angular amplitude along the three directions, while for tasks (2) and (3), the parameters estimated were the FD using the Higuchi algorithm [54], the CD, the Lyap, and the ApEn. Moreover, in these tasks we evaluated also the Var and the total spectral power extracted from the PSD estimated by using Welch's method. To identify significant differences between each pair of the three groups, we compared the parameters calculated on each task by using the rank sum Wilcoxon test with the Bonferroni correction for multiple comparisons.

The data analysis was performed by using a proprietary program written in MATLAB®.

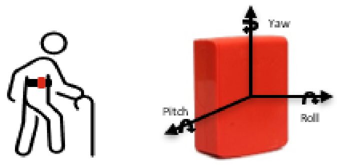


Figure 4. Schematic representation of the sensor placed on the subject's chest showing the Pitch, Roll, and Yaw angles.

2.6.3 Results

Table 1 presents the mean values and the standard deviation of the maximum angular amplitude (deg) and duration (s) for roll, pitch, and yaw rotations during task (1). Generally, the rotation along the pitch angle (anteroposterior direction) shows a very large amplitude with respect to roll and yaw in all the three groups. Comparing the amplitude values, the Young group exhibits the greatest amplitude, followed by Low Risk and High Risk groups with no significant differences between each pair of groups.

As well as for the amplitude, the duration of the movement for each angular rotation is quite similar among the groups. Only roll and yaw show a significant duration difference ($p=0.004$ and $p=0.002$, respectively) between Low Risk and Young group, even if the differences are only of about half a second. Furthermore, a large variability in High Risk group is present, almost double with respect to the Low Risk and Young groups, highlighting a higher variability in movement among High Risk group subjects.

Table 3. Mean values (± 1SD) of angular amplitude (deg) and duration (s) for each angular rotation during task (1) for Low Risk, High Risk, and Young group.

	Amplitude (mean ± SD) [deg]			Duration (mean ± SD) [s]		
	High Risk	Low Risk	Young	High Risk	Low Risk	Young
Pitch	44.4 ± 8.9	41.5 ± 19.1	47.9 ± 12.4	2.7 ± 1.5	2.3 ± 0.6	2.7 ± 0.8
Roll	14.5 ± 11.5	21.1 ± 31.9	21 ± 15.3	2.5 ± 1.4	1.7 ± 0.4	2.4 ± 0.6
Yaw	12.5 ± 6.5	18.2 ± 32.1	10.3 ± 4.6	2.3 ± 1.7	2.1 ± 0.7	1.6 ± 0.4

For tasks (2) and (3) the typical angular rotations behavior in elderly and Young groups is shown in Figure 5. In particular, the amplitude of the three angular rotations is greater and slower in elderly group (A and B graphs) compared to Young group (C and D graphs) in which low amplitude and high variability is present. To consider these differences, we subdivided the spectral power in three bands: 0.1-0.35 Hz, 0.35-0.7 Hz, and 0.7-5.0 Hz.

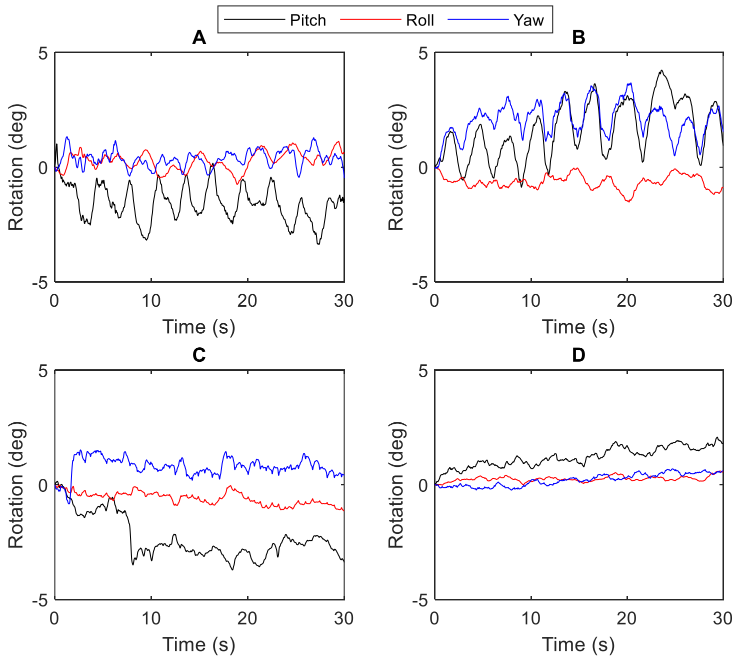


Figure 5. Angle rotation (deg) typical behaviour along the three directions during standing balance with open eyes (A and C) and with closed eyes (B and D), for the elderly group (A and B) and the Young group (C and D).

Table 4. Mean values (± 1SD) of parameters for task (2) in each group (High Risk, Low Risk, Young) together with p-values of the differences between Young and each of the two elderly groups. The parameters are: Spectral Power in 0.1-0-35 Hz, 0.35-0.7 Hz, and 0 Hz bands, Var, FD, ApEn, CorrDim, and the Lyap.

		High Risk	Low Risk	Young	High Risk vs Young	Low Risk vs Young
Pitch	0.1-0.35Hz [deg^2]	0.40 ± 0.40	0.37 ± 0.72	0.10 ± 0.21	0.007	0.027
	0.35-0.7Hz [deg^2]	0.13 ± 0.12	0.10 ± 0.20	0.03 ± 0.08	n.s.	n.s.
	0.7-5.0Hz [deg^2]	0.03 ± 0.04	0.02 ± 0.03	0.02 ± 0.03	n.s.	n.s.
	Var [deg^2]	1.10 ± 1.62	0.95 ± 1.52	0.53 ± 0.66	n.s.	n.s.
	FD [-]	1.19 ± 0.07	1.19 ± 0.06	1.29 ± 0.12	0.018	0.001
	ApEn [-]	0.58 ± 0.11	0.46 ± 0.14	0.49 ± 0.22	n.s.	n.s.
	Corr Dim [-]	0.31 ± 0.20	0.25 ± 0.19	0.18 ± 0.13	n.s.	n.s.
	Lyap [s^{-1}]	4.68 ± 1.98	2.41 ± 2.53	2.75 ± 3.23	n.s.	n.s.
Roll	0.1-0.35Hz [deg^2]	0.33 ± 0.79	0.13 ± 0.16	0.04 ± 0.08	0.002	0.010
	0.35-0.7Hz [deg^2]	0.06 ± 0.10	0.04 ± 0.05	0.01 ± 0.01	0.001	0.010
	0.7-5.0Hz [deg^2]	0.01 ± 0.03	0.01 ± 0.01	0.01 ± 0.01	n.s.	n.s.
	Var [deg^2]	0.77 ± 1.72	0.49 ± 0.52	0.16 ± 0.23	n.s.	0.048
	FD [-]	1.18 ± 0.07	1.21 ± 0.08	1.30 ± 0.10	0.008	0.025
	ApEn [-]	0.56 ± 0.07	0.51 ± 0.14	0.56 ± 0.13	n.s.	n.s.
	Corr Dim [-]	0.18 ± 0.23	0.18 ± 0.15	0.07 ± 0.09	n.s.	0.050
	Lyap [s^{-1}]	4.39 ± 2.52	2.95 ± 2.74	3.37 ± 2.43	n.s.	n.s.
Yaw	0.1-0.35Hz [deg^2]	0.21 ± 0.37	0.08 ± 0.10	0.07 ± 0.13	n.s.	n.s.
	0.35-0.7Hz [deg^2]	0.11 ± 0.23	0.03 ± 0.05	0.03 ± 0.09	0.018	n.s.
	0.7-5.0Hz [deg^2]	0.03 ± 0.03	0.02 ± 0.02	0.03 ± 0.08	0.046	n.s.
	Var [deg^2]	0.85 ± 0.71	0.53 ± 0.53	0.41 ± 0.45	n.s.	n.s.
	FD [-]	1.26 ± 0.10	1.31 ± 0.09	1.39 ± 0.17	n.s.	n.s.
	ApEn [-]	0.58 ± 0.17	0.56 ± 0.16	0.58 ± 0.26	n.s.	n.s.
	Corr Dim [-]	0.28 ± 0.19	0.18 ± 0.14	0.16 ± 0.14	n.s.	n.s.
	Lyap [s^{-1}]	4.72 ± 2.68	1.64 ± 2.89	3.19 ± 2.61	n.s.	n.s.

Table 5. Mean values (± 1SD) of parameters for task (3) in each group (High Risk, Low Risk, Young) together with p-values of the differences between Young and each of the two elderly groups. The parameters are: Spectral Power in 0.1-0-35 Hz, 0.35-0.7 Hz, and 0.7-5.0 Hz bands, Var, FD, ApEn, CorrDim and Lyap.

		High Risk	Low Risk	Young	High Risk vs Young	Low Risk vs Young
Pitch	0.1-0.35Hz [deg²]	0.94 ± 1.32	0.40 ± 0.84	0.09 ± 0.08	0.050	0.05
	0.35-0.7Hz [deg²]	0.35 ± 0.57	0.05 ± 0.08	0.02 ± 0.03	n.s.	n.s.
	0.7-5.0Hz [deg²]	0.05 ± 0.08	0.01 ± 0.02	0.01 ± 0.01	n.s.	n.s.
	Var [deg²]	1.73 ± 2.69	0.95 ± 1.62	0.34 ± 0.38	n.s.	n.s.
	FD [-]	1.18 ± 0.06	1.17 ± 0.07	1.25 ± 0.13	n.s.	0.014
	ApEn [-]	0.55 ± 0.08	0.50 ± 0.16	0.54 ± 0.17	n.s.	n.s.
	Corr Dim [-]	0.30 ± 0.25	0.24 ± 0.22	0.15 ± 0.11	n.s.	n.s.
	Lyap [s⁻¹]	3.23 ± 3.24	3.41 ± 3.37	2.98 ± 2.65	n.s.	n.s.
Roll	0.1-0.35Hz [deg²]	0.28 ± 0.56	0.13 ± 0.18	0.04 ± 0.06	0.021	0.050
	0.35-0.7Hz [deg²]	0.10 ± 0.17	0.06 ± 0.09	0.01 ± 0.01	0.026	0.023
	0.7-5.0Hz [deg²]	0.03 ± 0.04	0.01 ± 0.01	0.00 ± 0.01	n.s.	n.s.
	Var [deg²]	0.77 ± 1.39	0.42 ± 0.39	0.11 ± 0.15	n.s.	0.031
	FD [-]	1.19 ± 0.06	1.19 ± 0.07	1.27 ± 0.08	0.050	0.011
	ApEn [-]	0.60 ± 0.11	0.52 ± 0.14	0.58 ± 0.16	n.s.	n.s.
	Corr Dim [-]	0.17 ± 0.22	0.17 ± 0.16	0.04 ± 0.06	n.s.	0.023
	Lyap [s⁻¹]	3.92 ± 2.47	3.29 ± 2.92	3.96 ± 2.14	n.s.	n.s.
Yaw	0.1-0.35Hz [deg²]	0.16 ± 0.21	0.07 ± 0.13	0.03 ± 0.05	n.s.	n.s.
	0.35-0.7Hz [deg²]	0.12 ± 0.19	0.03 ± 0.05	0.01 ± 0.01	0.050	n.s.
	0.7-5.0Hz [deg²]	0.07 ± 0.10	0.01 ± 0.01	0.01 ± 0.01	0.002	n.s.
	Var [deg²]	0.82 ± 1.24	0.29 ± 0.29	0.19 ± 0.27	n.s.	n.s.
	FD [-]	1.34 ± 0.09	1.33 ± 0.07	1.42 ± 0.15	n.s.	n.s.
	ApEn [-]	0.69 ± 0.14	0.63 ± 0.21	0.70 ± 0.26	n.s.	n.s.
	Corr Dim [-]	0.21 ± 0.22	0.13 ± 0.12	0.08 ± 0.08	n.s.	n.s.
	Lyap [s⁻¹]	4.45 ± 2.30	3.23 ± 3.55	3.14 ± 2.63	n.s.	n.s.

Table 4 reports the values of parameters extracted for task (2) and Table 5 for task (3). No parameters show significant difference between Low Risk and High Risk groups. The spectral power in the 0.1-0.35 Hz band presents a progressive reduction passing from High Risk to Low Risk and Young group for each angular rotation. The reduction is significant only between Young and each of elderly groups for pitch and roll rotations. The spectral power in the 0.35-0.7 Hz band shows a similar behavior but the reduction is significant only between High Risk and Low Risk vs Young group in roll rotation. The spectral power in 0.7-5.0 Hz band presents negligible values comparable among the three groups.

Regarding the Var, High Risk group presents the greatest values, followed by Low Risk and Young groups. Due to the large variability between the subjects, the differences are significant only between Low Risk and Young groups.

FD has a higher value for Young group with significant differences between this group and both the High Risk and Low Risk groups in pitch and roll, while in yaw the differences are lower and not significant. The values of ApEn are comparable between the three groups with no significant differences for the three angular rotations. Concerning CorrDim, Young group presents lower values while High Risk presents the highest. However, only between Young and Low Risk groups in roll angular rotation the significant difference is present. The largest Lyap exponent is generally highest in the High Risk group without significant differences between group pairs.

2.6.4 Discussion

The aim of the study was to compare the kinematic characterization of movement between young and elderly groups during the execution of three tasks belonging to the Tinetti scale by using a wireless magneto-inertial sensor. Almost all parameters considered show differences between Low Risk and High Risk groups, but, due to the high variability and the low number of subjects, they are not significant. Thus, it is not possible to distinguish Low Risk and High Risk groups through the parameters examined in the tasks of arising from a chair, standing balance with open and closed eyes. However, the angular rotations confirm that some differences between High Risk and Low Risk group exist as reported in the only paper [7] which tried to quantify, by using an accelerometer, the Tinetti test score. Unfortunately, it is not possible to directly compare the results of our study with those of [58] due to the very different parameters.

On the other hand, the comparison between Young group and both High Risk and Low Risk groups exhibits some significant differences depending on the task, rotation angle and parameter.

Regarding task (1), the greatest amplitude is shown by pitch (which could be considered the main angular rotation for this task) followed by roll and yaw rotations not only in the Young group, as previously found in [56], but also in both High Risk and Low Risk groups. In pitch and roll rotations, the values in the Young group display a higher amplitude than in Low Risk and High Risk groups, sign of a greater dynamism of movement in young people. The only significant differences found in this task are the durations of roll and yaw rotations between Low Risk and Young groups. However, the total duration of the pitch, main angular rotation in the task of arising from a chair, is comparable in the three groups, being about 2.5 seconds. In addition, some subjects of the High Risk group show slower movement than other subjects probably due to the uncertainty in the execution of the task that is reflected in a great inter-subject variability.

In tasks (2) and (3), several parameters are found to be statistically different between Young group and elderly groups. In particular, the spectral power in 0.1-0.35 Hz and 0.35-0.7 Hz bands for pitch and roll rotations shows values significantly lower in Young group than in High Risk and Low Risk groups, with the latter presenting the lowest values. This is mainly due to the greater amplitude of angular rotation in anteroposterior and right-left directions with a longer duration of oscillation (lower frequency) in elderly groups than in Young group (see example in Figure 5). On the contrary, the Young group presents a negligible amplitude of movements at a higher frequency probably linked to a continuous quick control of the stability producing a very small spectral power and a FD significantly greater than in elderly groups. In yaw angle, the spectral powers are significantly greater in High Risk group in bands below 0.7 Hz which could be a sign of a wider undulation of the subject, while result is negligible in the other two groups. The CorrDim, measuring the complexity, is greater in High Risk group and significantly different only between the Low Risk and Young groups in roll rotation, probably due to the greatest angular amplitudes.

In conclusion, the low number of participants did not allow for highlighting any significant difference between High Risk and Low Risk groups in the three considered tasks. Thus, a higher number of enrolled subjects is necessary to verify if the detected differences could become significant. Despite the differences found in spectral powers in tasks (2) and (3) suggest a different balance regulation mechanism between young and elderly groups, with the latter showing a greater anteroposterior swing, a larger number of participants is required to confirm this hypothesis.

Moreover, the three considered tasks are not the most sensitive tasks to evaluate the risk of falling and do not seem to allow a precise distinction between subjects presenting different Tinetti scores. Extending the research to other tasks of the Tinetti test and increasing the number of subjects, it might be possible to use a device based on a magneto-inertial sensor to have a more reliable estimate of the risk of falling.

2.7 Quantitative identification of fall risk levels based on kinematic data

2.7.1 Aim of the study

This study aims to investigate the relationship between kinematic parameters, obtained from movements performed during the balance tasks of the Tinetti scale using wireless magneto-inertial sensors, and both the total Tinetti score and the balance subscore. Establishing such correlations could help identify quantitative parameters capable of objectively describing different levels of fall risk and balance impairment as defined by the clinical scale.

2.7.2 Materials and Methods

2.7.2.1 Participants

A total of 56 participants, aged between 58 and 93 years (mean age: 77.4 ± 8.1), were recruited from the Geriatric Department of Trieste Hospital. Eligibility criteria required participants to be over 55 years of age and to have been assessed as suitable for participation by a geriatrician. All participants received a detailed explanation of the study procedures and provided written informed consent prior to enrollment.

2.7.2.2 Study protocol

The study was conducted in the outpatient clinic of the Geriatric Department at Trieste Hospital. Each participant completed all tasks of the Tinetti scale, which were evaluated and scored by an experienced clinician. For the purposes of this analysis, six tasks were selected: rising from a chair (1), sitting down (2), standing in the Romberg position [42] with eyes open (3) and closed (4) for 30 seconds, maintaining a standing balance for 30 seconds (5), and turning 360 degrees (6). Calibration was performed before each task: prior to the first task, participants were seated with their backs resting against the chair's backrest, whereas for the subsequent tasks, calibration was carried out in an upright position with the arms relaxed along the sides.

2.7.2.3 Acquisition and Analysis

Four wireless magneto-inertial sensors (MTw Awinda Development Kit, Xsens Technology, Enschede, The Netherlands) were positioned on the chest and on both tibia. Data were acquired at a sampling frequency of 100 Hz. The wireless sensors, managed via dedicated software (MT Manager), enabled the collection of 3D Euler angles, pitch, roll, and yaw, corresponding to rotations around the anteroposterior, mediolateral, and vertical axes, respectively.

For tasks (1) and (2), the analysis included the total movement duration, the maximum angular amplitude along the three axes, the total angular displacement, and the final angular positions. For tasks (3), (4), and (5), several nonlinear and frequency-domain features were computed over the entire recording: FD using the Higuchi algorithm [33], CorrDim [59], Lyap [60], ApEn [61], [62]. Additionally, the Var and PSD were estimated. The total spectral power was derived from the PSD, calculated via Welch's method, and was analysed across three frequency bands (0.1–0.35 Hz, 0.35–0.7 Hz, and 0.7–5.0 Hz), as defined in previous work [55].

In task (6), analysis focused on spatiotemporal gait parameters, specifically the number of steps, mean and standard deviation of step duration, step amplitude, step velocity, stance duration, stance amplitude, and stance velocity.

All data processing and analyses were performed using a proprietary program developed in MATLAB®.

2.7.3 Results

The balance-related tasks from the Tinetti Scale were analysed in this study to explore potential correlations between kinematic parameters, computed from the behaviour of Euler angles, and the scores assigned, both total and balance subscores. All results are presented as scatter plots to visually highlight possible relationships between each extracted parameter and the corresponding clinical score.

For task (1), rising from a chair, and task (2), sitting down, only data from the sensor positioned on the chest were analysed. Among the three angular directions, only the pitch component was considered, as it was expected to capture the primary forward–backward movement characteristic of these tasks. In contrast, roll and yaw were excluded due to their minimal and less informative contributions. The parameters extracted for these tasks included the maximum angular amplitude, movement duration, total angular displacement, and final angular position. Figure 6Figure 7 illustrate the relationship between each parameter and the corresponding score. However, no clear trend or strong correlation emerged, and the parameters did not allow a reliable distinction between different risk levels as defined by the Tinetti scale.

To assess the complexity and variability of body movements during the standing balance tasks (tasks 3, 4, and 5), a set of nonlinear and frequency-domain features was computed [63]. Among the three sensors used, only the one positioned on the chest was analysed, as it is the most sensitive to subtle body oscillations compared to the tibial sensors [7]. Additionally, the yaw component, corresponding to rotation around the vertical axis, was excluded due to its limited contribution to postural sway analysis. Figure 8Figure 10Figure 12 presents the result for pitch angle, while Figure 9Figure 11,Figure 13 for roll angle, respectively, in tasks 3, 4, and 5. From a general observation, the computed parameters did not show a clear correlation with the Tinetti total or balance scores. No consistent trend was identified across participants, suggesting high inter-individual variability in the kinematic measures.

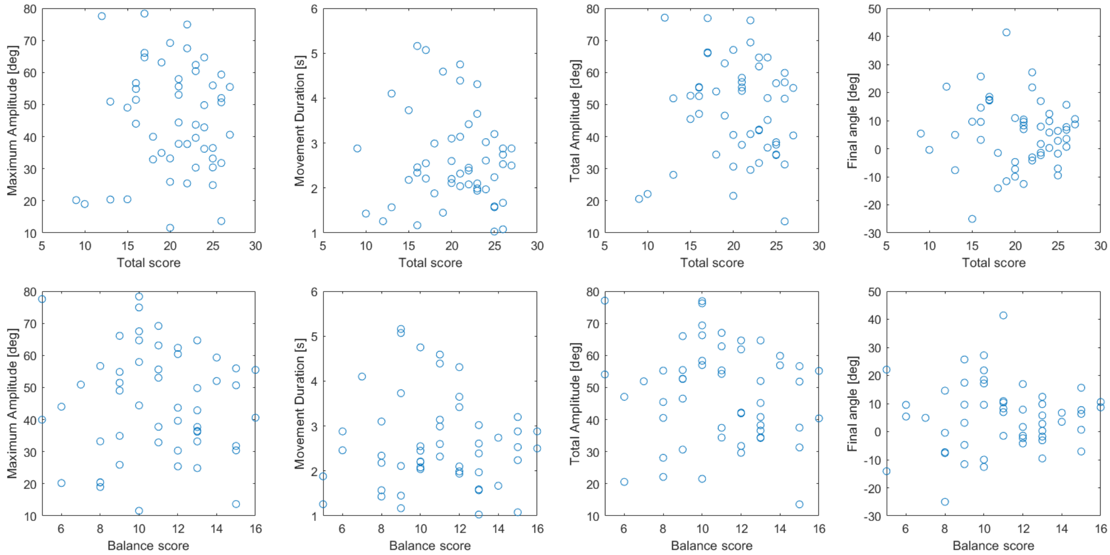


Figure 6. Scatter plots illustrating the Maximum Amplitude, Movement Duration, Total Amplitude, and Final Angle for Task (1), rising from a chair. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

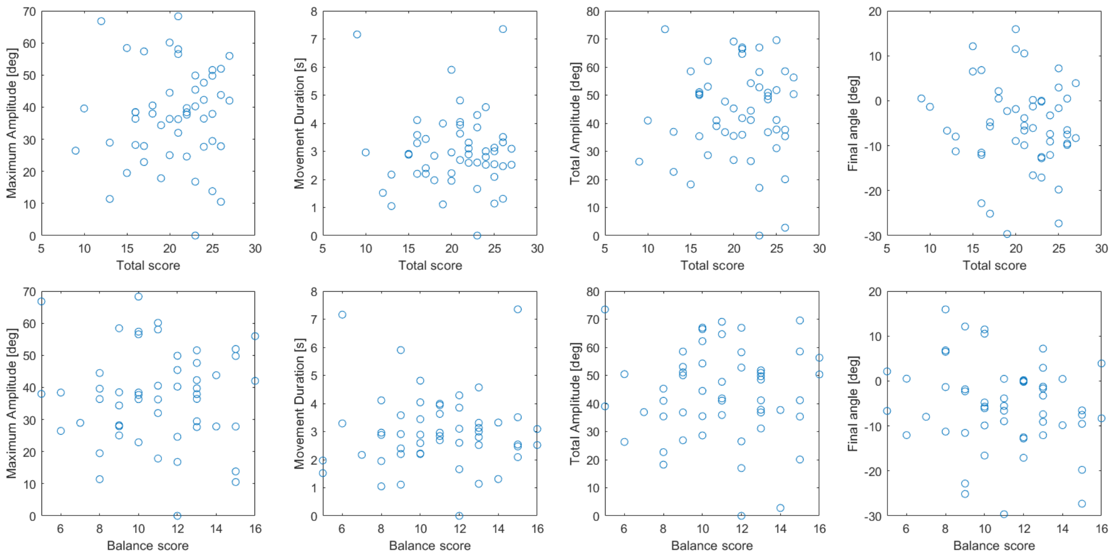


Figure 7. Scatter plots illustrating the Maximum Amplitude, Movement Duration, Total Amplitude, and Final Angle for Task (2), sitting down. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

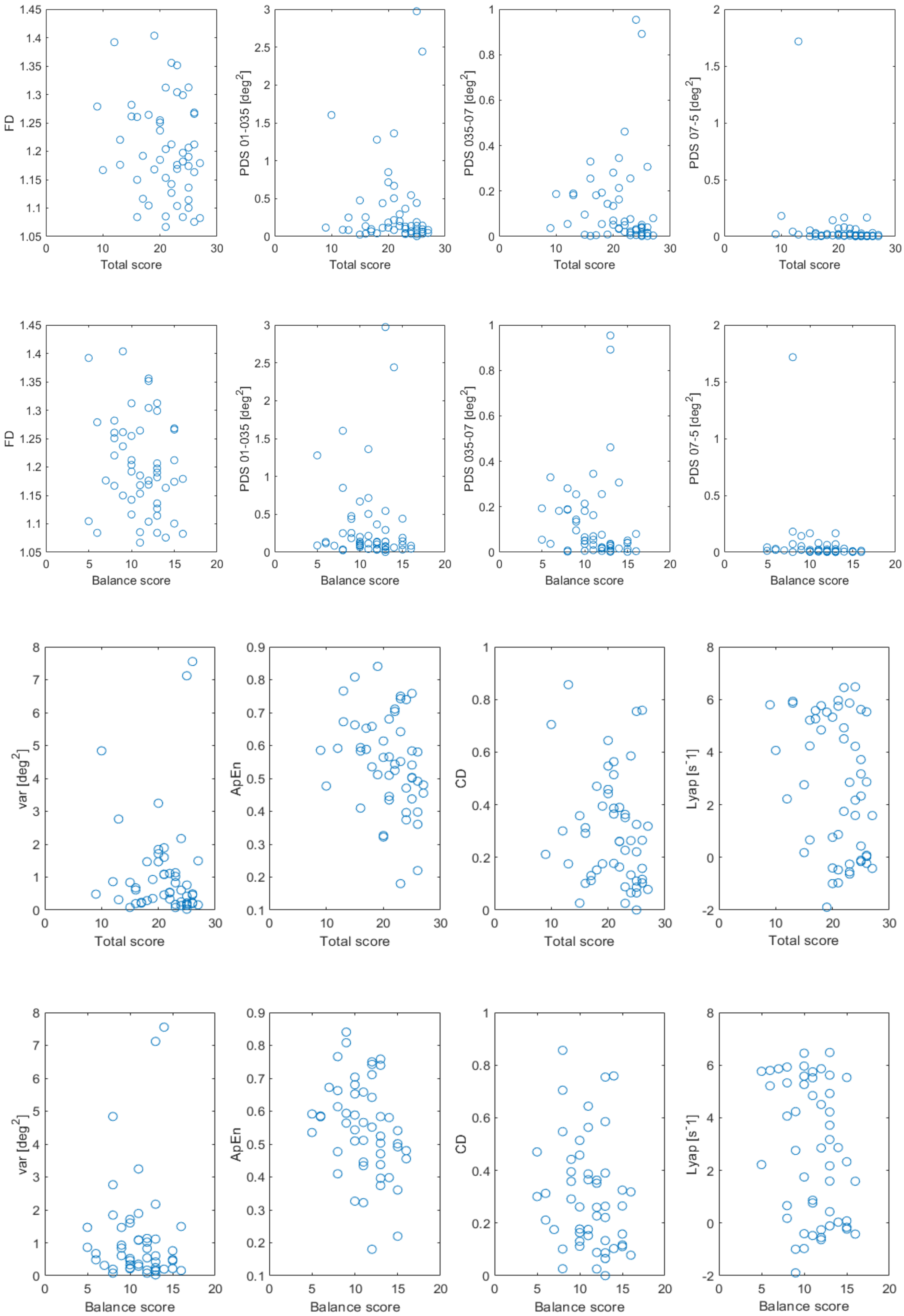


Figure 8. Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (3), Romberg with open eyes, pitch component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

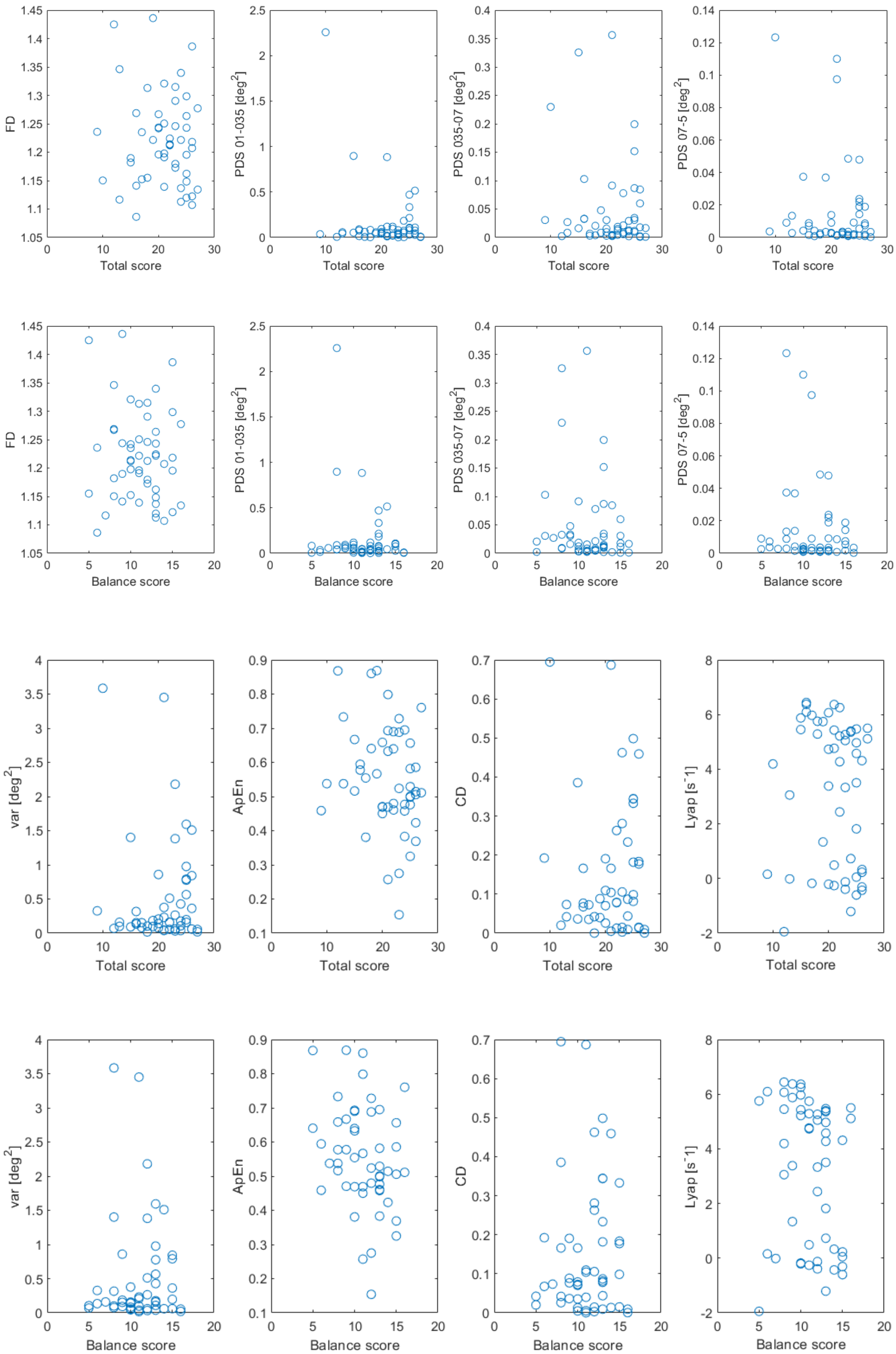


Figure 9. Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (3), Romberg with open eyes, roll component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

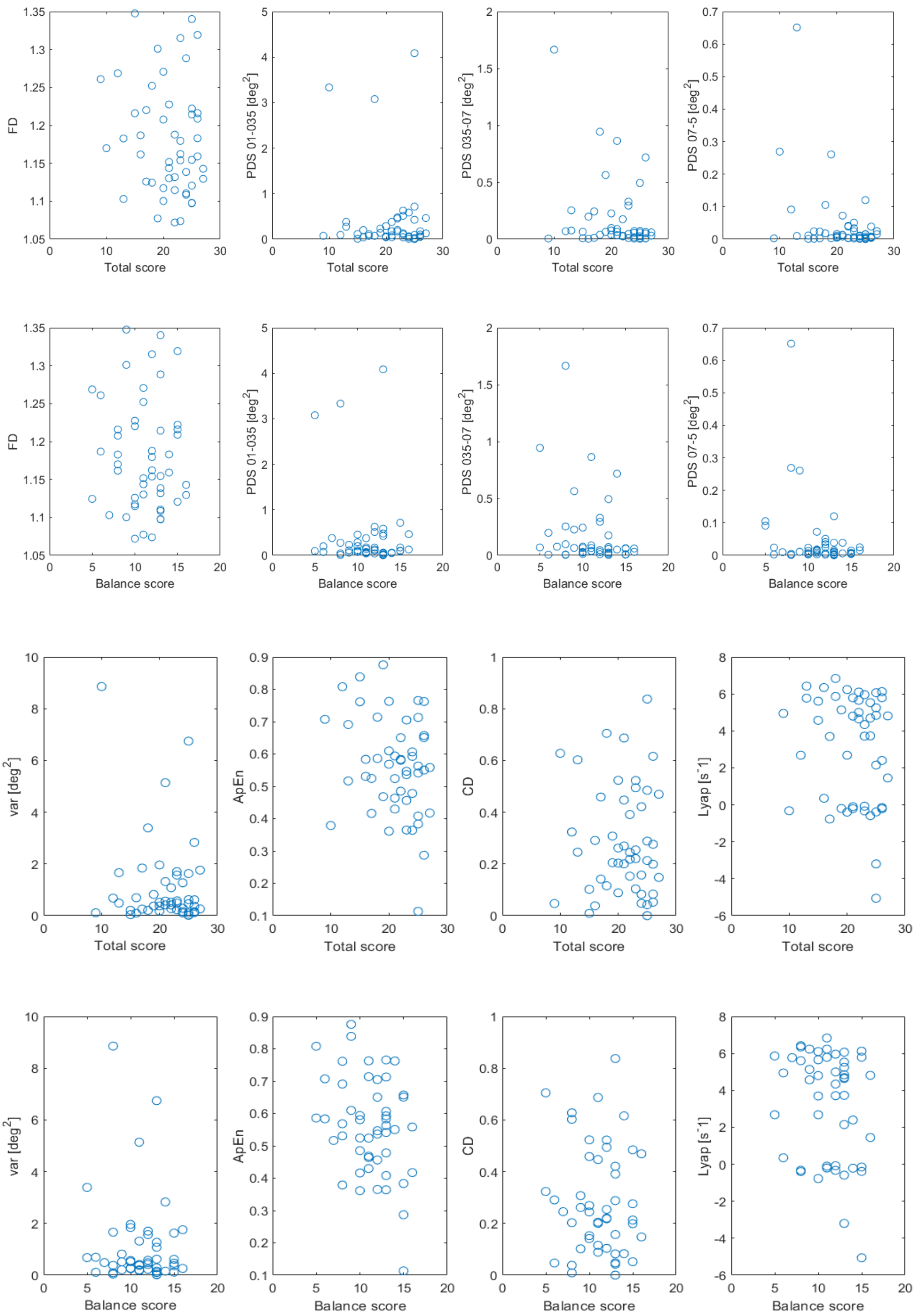


Figure 10. Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (4), Romberg with open eyes, pitch component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

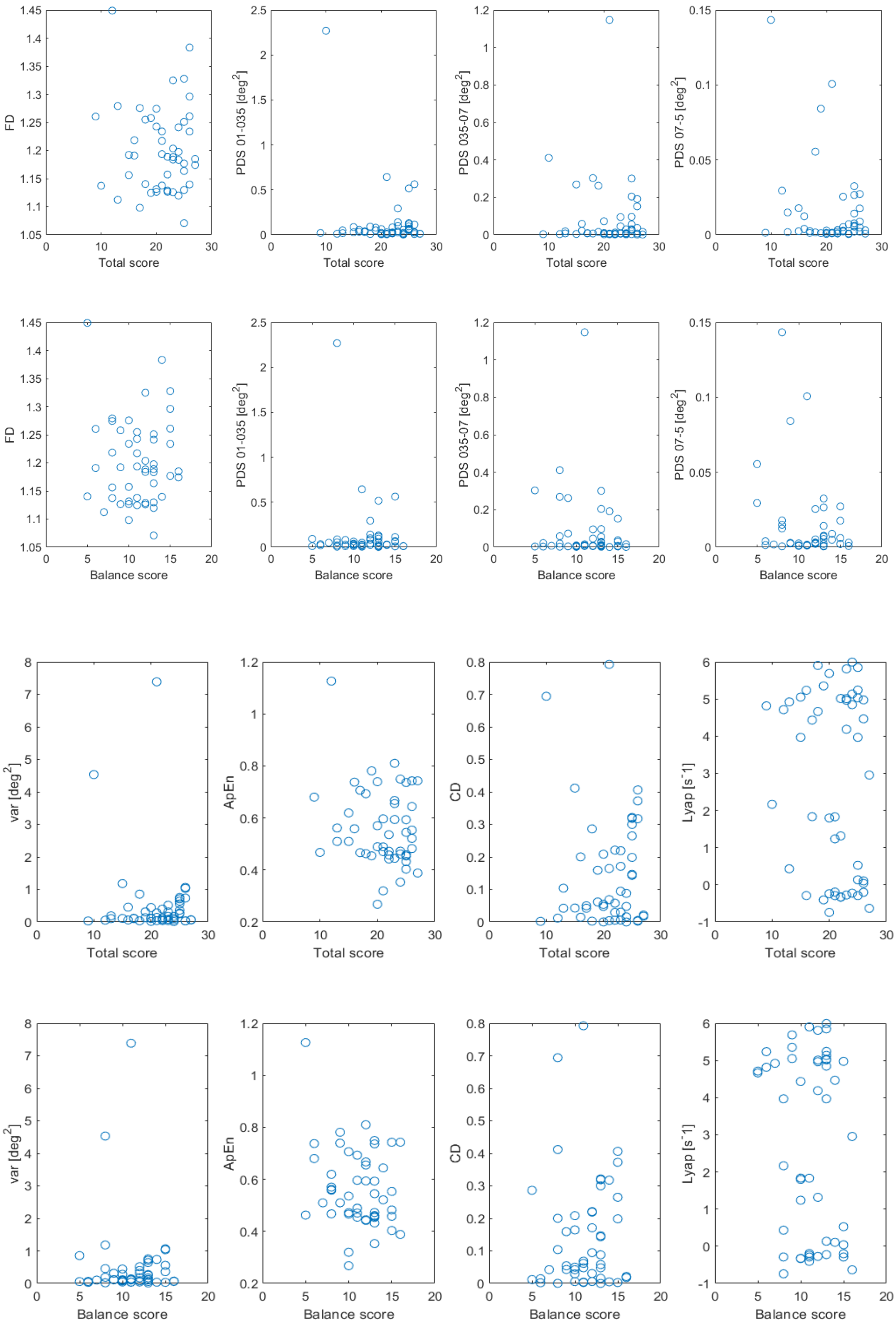


Figure 11. Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (4), Romberg with open eyes, roll component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

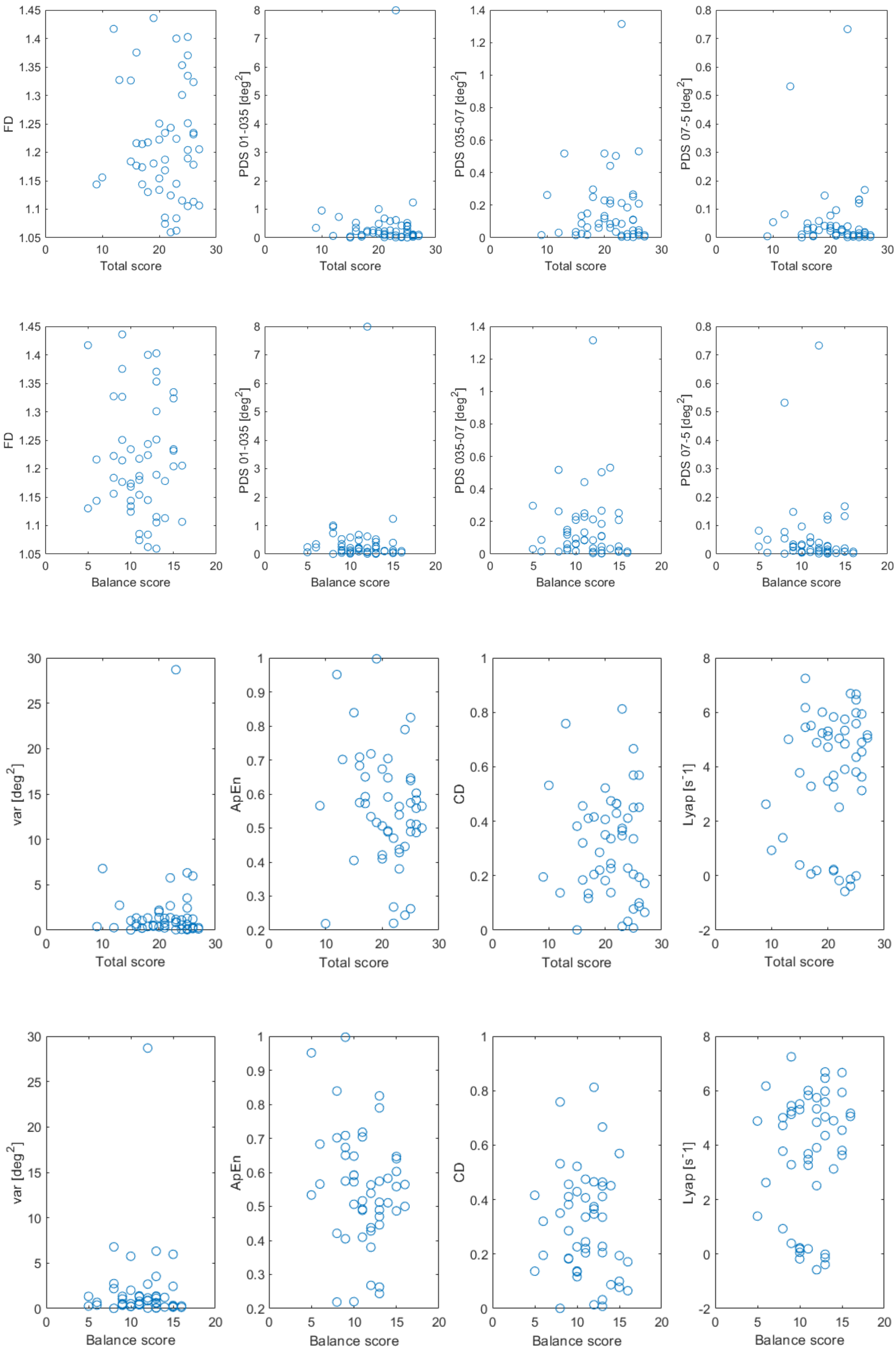


Figure 12. Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (5), Romberg with open eyes, pitch component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

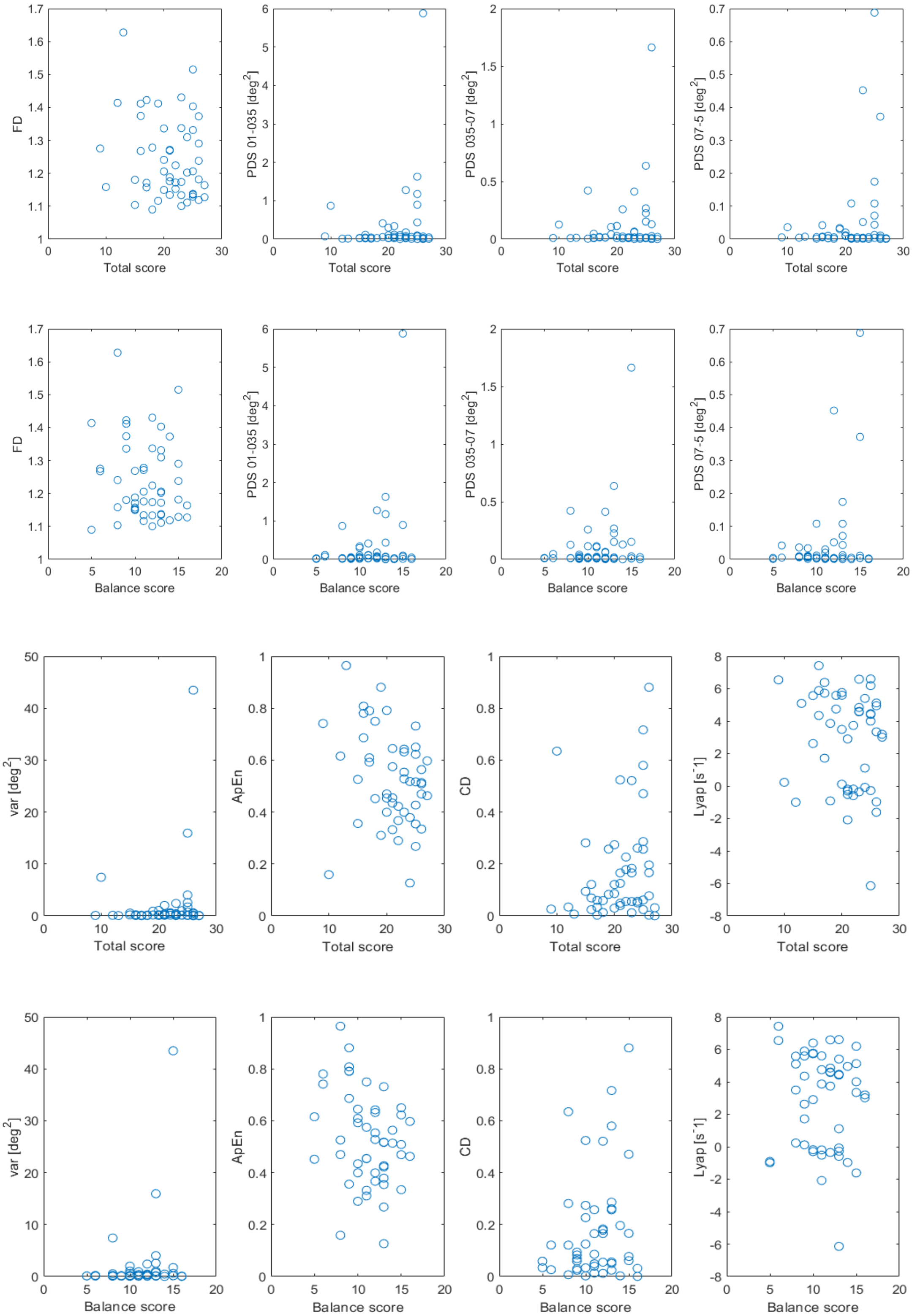


Figure 13. *Scatter plots illustrating the FD, Var, ApEn, CorrDim, Lyap, and PSD in 0.1-.035 Hz, 0.35-0.7 Hz, 0.7-5.0 Hz for Task (5), Romberg with open eyes, roll component. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).*

For task (6), turning 360 degrees, spatiotemporal gait parameters were derived from the sensors positioned on both tibiae. The analysis focused on the pitch angle, as it represents the primary angular rotation associated with stepping movements and enables accurate identification of gait cycles. From this signal, parameters such as the number of steps, mean and standard deviation of step duration, step amplitude, step velocity, stance duration, stance amplitude, and stance velocity were computed to characterise gait performance during the turning movement. Figure 14 illustrates the relationship between the extracted gait parameters and the Tinetti scores. Although a clear correlation is not evident, a trend can be observed for the Median Step Duration and Mean Step Amplitude, both showing a partial association with the total and balance scores.

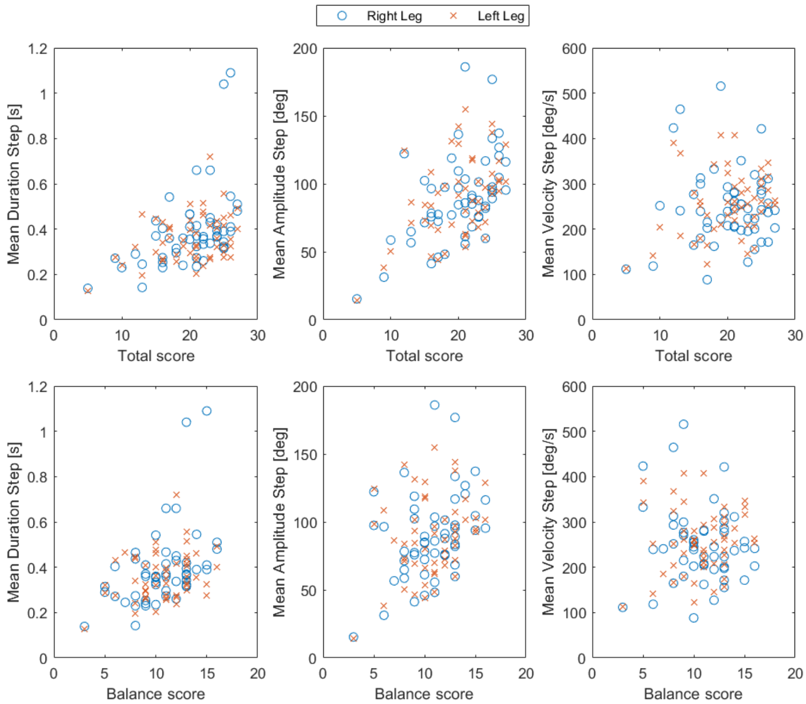


Figure 14. Scatter plots illustrating the Mean Step Duration, Mean Step Amplitude, and Mean Step Velocity, for Task (6), turning 360 degrees. The parameters are shown in relation to the total Tinetti score (top row) and the balance subscore (bottom row).

2.7.4 Discussion

The combined analysis of balance and fall risk using the Tinetti Scale, together with three-dimensional angular rotation data obtained from IMU sensors, provides a valuable framework for the quantitative assessment of balance performance. Previous studies have investigated the integration of objective kinematic measures to complement or validate clinical scales such as the Tinetti Test [64], [7]. However, direct comparison with the present work is limited by methodological differences, including sensor placement, data type, and the specific kinematic parameters extracted.

This study specifically investigates the correlation between kinematic parameters, primarily derived from pitch and roll angle measurements, and both the total and balance subscores of the Tinetti test. The results provide important insights into the relationship between balance performance and fall risk across different tasks and populations.

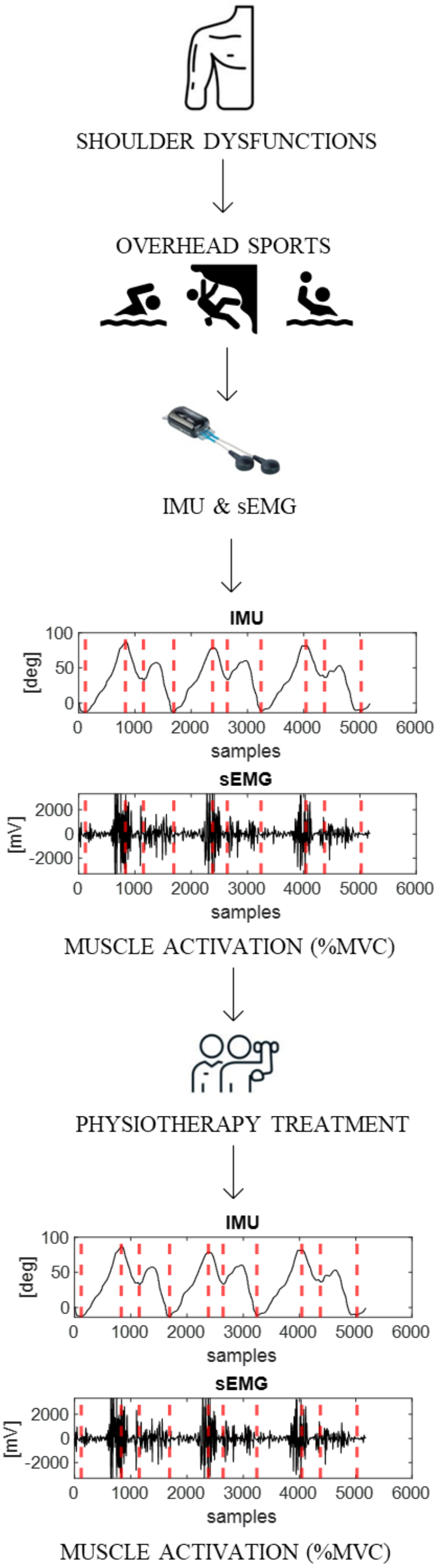
In examining tasks such as rising from a chair (Task 1) and sitting down (Task 2), only the pitch component from the IMU sensor mounted on the chest was considered due to its relevance to the anticipated forward-backward movement characteristic of these activities. The collected parameters, including maximum angular amplitude, movement duration, total angular displacement, and final angular position, were analysed, but could not establish a clear trend or strong correlation with Tinetti scores. This observation aligns with research suggesting that traditional clinical assessments like the Tinetti Scale may not fully encapsulate the complexities of postural control as assessed through kinematic tools [65]. Therefore, while IMU-based measurements offer detailed quantitative descriptions of movement dynamics, their ability to discriminate between different fall risk levels appears limited within the scope of this dataset, which seems to reflect differences in individual strategies, strength, and confidence rather than balance control alone.

For the standing balance in Tasks 3, 4, and 5, a nonlinear and frequency-domain features analysis from the chest-mounted IMU was performed. The exclusion of the yaw component, which measures rotation around the vertical axis, was justified as it contributes minimally to postural sway assessment [66]. Even if the parameters considered evaluate balance, stability, control, and adaptability, they do not detect any greater change in the execution of the tasks. Despite an increased number of elderly participants compared with [42], the data demonstrated greater inter-individual variability as well, which further complicated the establishment of consistent correlations with Tinetti scores.

Task 6, turning 360 degrees, yielded spatiotemporal gait parameters extracted from tibial sensors, focusing on the pitch angle. Parameters such as mean step duration, mean step amplitude, and mean step velocity were analysed against Tinetti scores. Although a significant correlation was not established, trends emerged for specific parameters like median step duration and mean step amplitude, suggesting that gait regularity and smoothness during turning could be related to balance performance. Turning movements are complex tasks that challenge postural control, and even small deviations in timing or amplitude may reflect reduced stability. As expected, subjects at higher risk of falls performed shorter and smaller steps. These findings resonate with earlier studies linking gait stability and temporal features with balance assessments, indicating potential applications for IMU data in predicting fall risks [67].

The potential use of IMU technology to enhance clinical assessments is supported by research where kinematic data has provided valuable insights into evaluating balance and gait [65], [68]. While the present analysis did not yield strong correlations across the evaluated tasks with Tinetti scores, the findings suggest that the integration of IMU sensors provides valuable insights into movement dynamics. It highlights the need for further exploration into how device-derived kinematic data can complement traditional clinical assessments to better predict fall risk and enhance patient outcomes.

Chapter 3. A NOVEL APPROACH TO STUDYING THE SHOULDER JOINT IN OVERHEAD SPORTS



3.1 Introduction

The shoulder is a ball-and-socket joint formed by the integration of bones, tendons, and muscles. It exhibits the greatest mobility of all human joints, enabling a wide range of movements, including flexion and extension, abduction and adduction, internal and external rotation, as well as circumduction, a composite motion encompassing multiple planes [69], [70], [71]. However, this remarkable mobility comes at the expense of stability, rendering the shoulder particularly vulnerable to injuries and dislocations. Thus, the functional integrity of the shoulder depends on a delicate balance between mobility and stability.

Shoulder dysfunctions rank as the third most common musculoskeletal disorder and are one of the leading causes of musculoskeletal pain and disability worldwide. These conditions range from disorders with clearly defined diagnostic criteria and pathophysiology to more ambiguous presentations that lack definitive diagnostic markers [19], [20], [21].

The prevalence of shoulder injuries is especially high among athletes, particularly those engaged in overhead sports requiring repetitive, high-intensity movements at extreme ranges of motion. Such explosive and large-amplitude actions significantly increase the mechanical loading of the joint, thereby elevating the risk of dysfunction. The characterization of shoulder kinematics is therefore fundamental not only for clinical purposes, such as diagnosis, treatment evaluation, and recognition of pathological movement patterns, but also for the early detection of subtle biomechanical alterations [20], [21], [72]. In the context of sports, the study of shoulder kinematics further contributes to performance optimization and injury prevention through the identification of compensatory or maladaptive movement strategies [73].

The following chapter reports three studies conducted in collaboration with the Department of Physiotherapy of the University of Trieste involving athletes from overhead sports disciplines. For each sport examined, a specific acquisition protocol and an associated physiotherapy treatment plan were developed to address the unique biomechanical and functional demands of each activity. Sections 3.4 to 3.8 focus on studies related to the swimmer’s shoulder, exploring kinematic patterns, muscular activation, and rehabilitation approaches. Sections 3.9 and 3.10 examine climbing, analyzing shoulder function under load-bearing and overhead movement conditions. Finally, Sections 3.11 and 3.12 investigate water polo players, assessing shoulder performance and recovery strategies specific to this high-intensity aquatic sport.

3.2 Shoulder joint through wearable devices

To objectively study shoulder kinematics and movement behaviour across different contexts, the literature describes non-invasive assessment methods classified into wearable and non-wearable systems. Non-wearable devices, such as ultrasound-based motion analysis systems [74], stereophotogrammetry [75], and optoelectronic systems [20], are widely regarded as the gold standard because of their high reliability, accuracy, and precision [4]. However, these approaches are associated with notable limitations: they are costly, require extensive data processing, and are restricted to controlled laboratory environments. Wearable systems have emerged as a practical and portable alternative, enabling the assessment of kinematic parameters outside the laboratory and supporting long-term monitoring of human movement [2], [76]. In recent years, their role has expanded considerably, with applications in clinical diagnostics, rehabilitation monitoring, and the management of neurological and musculoskeletal disorders [4].

The most common wearable devices are IMU, which typically integrate a triaxial accelerometer, a gyroscope, and a magnetometer within a single sensor. These units rely on sensor fusion algorithms to reduce measurement errors and improve accuracy [77]. Magnetic measurement systems, in particular, allow precise estimation of the three-dimensional position and orientation of rigid bodies [4], [78] and have demonstrated utility in assessing both upper- and lower-limb joints [79], [80], [81], [82].

From a biomechanical perspective, the upper limb can be modeled as a kinematic chain composed of rigid segments, thorax, upper arm, forearm, and hand, connected by joints that permit relative motion. The shoulder joint contributes three primary degrees of freedom: abduction/adduction, internal/external rotation, and flexion/extension. These rotations are commonly described using Euler angles, which map the anatomical axes of motion to roll, pitch, and yaw [4]. Methodologies for ambulatory assessment of upper-limb kinematics have been proposed [83]. Notably, Cutti et al. [78] developed a protocol to quantify scapulothoracic, humerothoracic, and elbow kinematics in ambulatory conditions, particularly for simple and slow movements [84], [85].

Shoulder joint stability is also maintained by active muscular control and a very lax joint capsule. In addition to motion tracking, superficial electromyography (sEMG) represents a valuable non-invasive technique for evaluating shoulder function [86]. Wearable sEMG systems, typically consisting of surface electrodes placed over specific muscles, have advanced considerably in terms of portability and wireless connectivity.

In addition to sEMG and IMU-based assessments, researchers have also explored the use of tensiomyography (TMG) and handheld dynamometry to evaluate shoulder muscle function. TMG is a non-invasive technique that quantifies the mechanical contractile properties of skeletal muscles, providing valuable information about muscle tone, stiffness, contraction time, and fatigue-related adaptations [87], [88]. Previous studies have demonstrated that TMG is highly sensitive in detecting fatigue-induced changes in upper-limb muscles, particularly in the biceps brachii, following high-volume and high-load resistance exercises [89]. Furthermore, shoulder strength, as measured by handheld dynamometry, is known to decrease under fatigue, which can alter proprioceptive control and range of motion, thereby increasing the risk of overuse shoulder injuries [90]. These complementary assessment tools contribute to a more comprehensive understanding of neuromuscular performance and fatigue mechanisms, offering potential value for both clinical evaluation and sports performance monitoring.

3.3 Shoulder joint in overhead sports

3.4 Swimming

Shoulder injuries are highly prevalent in swimming, with an annual incidence estimated between 23% and 38% [14]. The swim stroke requires complex, highly coordinated musculoskeletal sequences that expose the shoulder to substantial multidirectional stresses and repetitive loading. These mechanical demands frequently lead to pain or discomfort in the shoulder region. Over time, swimmers may develop adaptive structural and functional changes that facilitate the execution of sport-specific movements. However, such adaptations often disrupt normal glenohumeral biomechanics, thereby increasing the risk of various pathological conditions [91], [92].

The term swimmer's shoulder was originally introduced by Kennedy and Hawkins to describe supraspinatus tendon impingement beneath the coracoacromial arch. Contemporary research, however, emphasizes that shoulder pain in swimmers is multifactorial in nature. Contributing factors include subacromial impingement syndrome, repetitive overuse, muscular fatigue, scapular dyskinesis, and increased joint laxity or instability [72], [93]. In addition, disruptions in the coordinated activation of scapular and glenohumeral muscles can lead to abnormal kinematics. For example, swimmers with impingement syndrome have been shown to exhibit reduced scapular upward rotation, a maladaptation that may compromise shoulder stability and increase injury risk [93].

Given these risks, the assessment of shoulder kinematics in swimming holds considerable importance for both performance optimization and clinical practice. Motion analysis facilitates the identification of altered movement patterns that may predispose athletes to injury, provide early indicators of pathological change, or reflect the residual effects of previous lesions [73].

Moreover, identifying the muscles that are more implicated in swimming and that might be affected by fatigue or alterations of their activity could help to identify those swimmers with a higher risk of developing shoulder pain.

3.5 Influence of fatigue in swimmers suffering from swimmer shoulder pain [94]

3.5.1 Aim of the study

The aim of our work is to study the effects of fatigue on the kinematic of the shoulder joint between thorax and arm in swimmers suffering from *swimmer shoulder* compared to healthy swimmers by using an IMMS.

3.5.2 Materials and Methods

3.5.2.1 Participants

11 young swimmers (7 male), aged between 15 and 27 (mean 20.18 ± 3.5 years), recruited from local swimming clubs, took part in the study. Inclusion criteria required participants to have a minimum of 3 years of swimming training and maintain a training volume of at least 4.5 h per week, with the front crawl being their primary swimming style. Six subjects were identified as healthy as they did not present a history of chronic or acute muscular or joint diseases affecting the tested shoulder or upper limb and five individuals were considered affected by *swimmer shoulder* pathology by an expert physiotherapist team.

Participants and their legal guardians were properly informed about the protocol and signed informed consent before being enrolled in the study that received approval from the local university ethics committee (122/2022) and adhered to the principles outlined in the Declaration of Helsinki.

3.5.2.2 Study Protocol

The protocol used for the movement acquisitions is part of a more extended protocol within which tensiomiography and shoulder strength measurements were also carried out and not considered for this study [95].

The testing session took place in a local swimming pool and consisted of three parts: during the first and the third part concerning the execution of the exercise, the movement was measured through inertial sensors, while during the second part the fatiguing protocol was carried out. In the first and third part, the participants were asked to perform 40s of dry front crawl exercise using only one arm, lying prone on a physiotherapist's bed, with a metronome set at a reasonable frequency of beats per minute (70) to synchronize the movements. At each beat the arm had to be over the head or along the body. Thus, each subject performed 23 full strokes (one every two beats). Before the start and after the end of the exercise, the subject kept resting position with arm along the body for 10 s. The fatiguing procedure consisted of 30 min front crawl swimming in water at different incremental intensities as the protocol described in [95].

3.5.2.3 Acquisition and Analysis

Since the aim of the study was to characterize the movement of the shoulder, a reduced version of the protocol proposed by Magalhaes et al. [96] was adopted. Two wireless magneto inertial sensors were used to acquire data from MTw Awinda Development Kit (Xsens Technology) able to detect 3D Euler angles: one sensor was placed on the subject's chest (on the flat portion of the sternum), while another sensor on the subject's arm (above the centre of the homerus and posteriorly) by means of Velcro body strap. This setup allowed for the study of the arm movement relative to the thorax. For the calibration phase, the subject was asked to keep the tested arm along the body for some seconds. The devices are linked to the host through the dedicated software (MT Manager) which allows the operator to collect data at a sample frequency of 100 Hz during the first and the third phase of the protocol. For each stroke and for each Euler angle, maximum and minimum values were identified, and the peak-to-peak amplitude of the stroke was calculated, excluding the first and the last stroke. As the stroke duration was strictly linked to the metronome synchronization, the analysis focused only on the amplitude of the movement according to the three angle rotations. In order to estimate the mean shape of stroke movement of each subject, for each Euler angle, all the strokes were synchronized at the times in which the minimum amplitude peaks were reached and then averaged. To highlight the existence of significant differences in stroke amplitude either in healthy and pathological group, in this preliminary analysis, we compared the amplitude values of the three angular rotations before and after the fatiguing session in healthy and pathological subject. The significance of the differences was assessed with the sign rank test for paired group in the comparison before/after fatiguing exercise and with the rank sum Wilcoxon test in other comparison. The analysis was performed using a program written in MATLAB®.

3.5.3 Results

Figure 1 reports the average trends of stroke amplitude for each movement rotation in healthy and pathological subjects before and after the fatiguing protocol.

Internal/External rotation and Adduction/Abduction rotation present a similar behavior with slight difference between healthy and pathological group and before and after the exercise, while the Flexion/Extension rotation presents a marked difference between healthy and pathological group both before and after the fatiguing exercise. Noticeably, the movement trends are similar both between the two groups and before and after exercise separately in each of the three Euler angles (Figure 15).

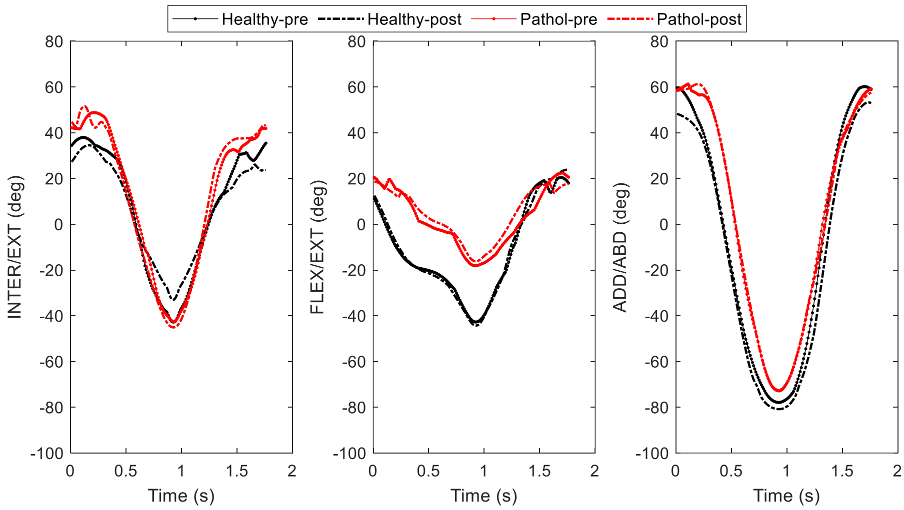


Figure 15. Mean amplitude trend of the three movements rotation in healthy (black) and pathological (red) subjects before (solid line) and after (dashed line) the fatiguing protocol.

Table 6 shows mean values (± 1SD) of stroke amplitude calculated on subjects belonging to the two groups before and after the fatiguing protocol for the three movement rotations. The results correspond to what shown in the graph of Figure 5 highlighting a large variability (SD greater than 10 degs) among the subjects in most of all the situations.

Because of the low number of subjects, we report in

Figure 16 the boxplot presenting median values, 75th and 25th percentile of the three rotation angles in the two groups before and after exercise. As expected, only Flexion/Extension shows (Table 6) a large difference before and after the fatigue in both healthy and pathological group while the Adduction/Abduction rotations present a reduction of the amplitude after the fatiguing task both in healthy and pathological subjects. Finally, the Internal/External rotations show an increase only in pathological subjects after exercise. However, due to the large variability among the responses of the subjects and the low number of participants, only the Flexion/Extension rotations showed significant ($p=0.0043$) differences between healthy and pathological group both before and after the fatigue exercise.

Table 6. Mean values (± SD) of stroke amplitude (deg) before ('pre') and after ('post') fatiguing esercise for each of the three rotation angles (Internal/External, Flexion/Extension, Adduction/Abduction) in healthy and pathological subjects.

	Inter/Ext [°]	Flex/Ext [°]	Add/Abd [°]
Healthy-pre	87±11	77±13	144±10
Healthy-post	85±18	80±16	138±7
Pathol-pre	91±12	46±6	142±15
Pathol-post	96±12	40±6	141±16

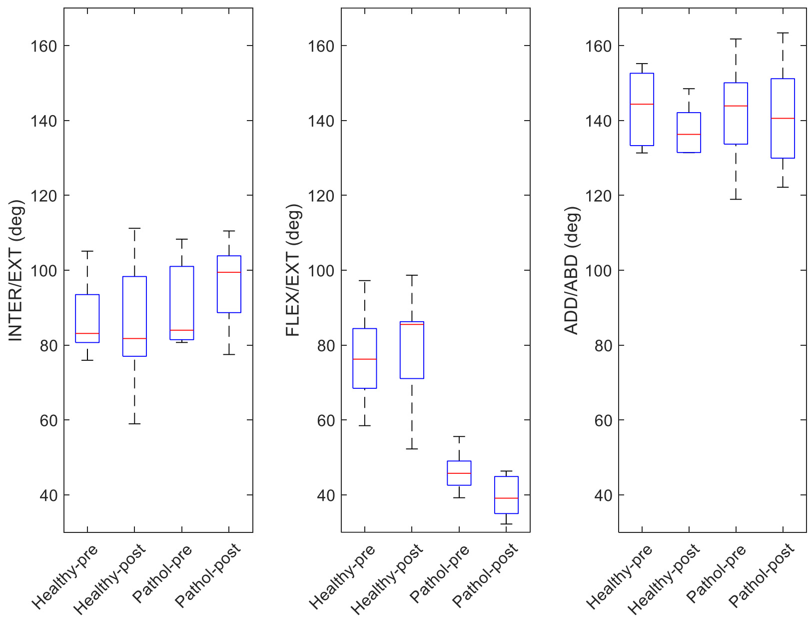


Figure 16. Box plots of the amplitude values. In red, the median values with the box delimiting the 25th and 75th percentiles. The results are displayed separately for each movement rotation in the two different conditions (before and after fatigue exercise) for healthy and pathological subjects.

3.5.4 Discussion

Swimmer shoulder represents one of the main disabilities in elite swimmer and some researchers demonstrated that fatigue could negatively affect strength, range of motion and proprioception thus highlighting a connection between fatigue and potential mechanisms leading to shoulder pathology in swimmers [91]. The aim of our work was to study the effects of fatigue on the kinematic of the shoulder joint in swimmers suffering from swimmer shoulder compared to healthy subjects.

The used of magneto inertial sensors allowed to verify the repeatability of the kinematic movement and to quantify the rotation angles. The results showed some differences after fatiguing exercise in Internal/External rotation with an increase of the amplitude in pathological subject and a slight decrease of the amplitude after the fatigue in Adduction/Abduction in both healthy and pathological groups. However, despite these differences were not significant mainly due to the low number of subjects and a large standard deviation (see Table 6 and Fig.2), the results found confirm that fatigue has an influence on the kinematics of the shoulder as reported in [91], but we cannot directly compare the results because of the different methodology used.

Finally, the only significant difference we found is between healthy and pathological groups in the Flexion/Extension rotation, both before and after the fatigue exercise with greater values in healthy subjects. These differences are ascribable to the pathology, and our methodology allowed us to highlight which component of the movement is affected, enabling a more targeted treatment in order to recover the incorrect motor response due to the pathological condition.

3.6 Impact of physiotherapy on shoulder kinematics with swimmer's shoulder pain [97]

3.6.1 Aim of the study

Since swimmer's shoulder can be treated with different rehabilitation protocols [92], [98], [99], and to the best of the authors' knowledge, kinematic differences have been scarcely used as an outcome for these treatments. In this preliminary work, we aimed to describe the effects of physiotherapy treatment and the influence of fatigue due to swimming on shoulder kinematics using IMU. Furthermore, this is an extension of [94], in which we explore only fatigue's influence in healthy subjects and those suffering from swimmer's shoulder pain.

3.6.2 Materials and Methods

3.6.2.1 Participants

Male swimmers, aged between 18 and 40 years, from local swimming clubs were recruited in this study. The inclusion criteria specified a minimum of three years of swim training, an average weekly training volume of at least 4.5 h, and a primary specialisation in the front crawl stroke. A team of expert physiotherapists deemed the subjects to have swimmer's shoulder pathology, and they were excluded if the presence of primary impingement, tissue pathology, or high irritability was found. In particular, the recommendations suggested by Tovin [100] were adopted, focusing on the impairments that are associated with the onset of symptoms including glenohumeral hypermobility or instability, impaired posture, impaired rotator cuff strength, altered scapulohumeral rhythm or poor neuromuscular control, or a tight posterior capsule. Indeed, all the subjects who first reported a subjective feeling of pain in the shoulder district while swimming or immediately after received a dedicated evaluation. After the subjective assessment, which collected information about the area, symptoms description, behavior, and intensity, a physiotherapy clinical examination checked for postural abnormalities, reduced shoulder range of motion with pain, altered scapulohumeral rhythm, and impingement tests.

All participants were provided comprehensive information regarding the study's purpose, methodology, and potential risks and benefits. The study protocol received ethical approval from the local university ethics committee (reference number 122/2022). Written informed consent was obtained from each participant before study enrolment, in accordance with ethical guidelines for human subject research.

3.6.2.2 Study Protocol

To register the movement, we followed the same protocol as in a previous study from our group [101]; in more detail, the experimental protocol, performed in a local swimming pool, consisted of two testing tasks, separated by a standardised in-water fatiguing session (Figure 17).

In particular, during the two testing tasks, inertial sensors were applied, and participants executed a 40 s dry front crawl exercise using the pathological arm while lying prone on a physiotherapist's bed with a metronome set at 70 beats per minute. The arm had to go over the head or along the body at each beat to perform one stroke every two beats. Moreover, the protocol provides for one calibration phase before the start of the exercise, with the subject maintaining the resting position with the arm along the body for a few seconds.

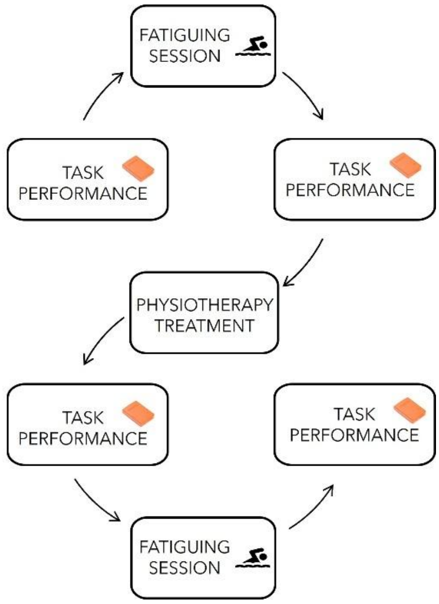


Figure 17. Schematic diagram representing the phases of the protocol implemented in the study.

The fatiguing session performed between the two testing tasks was conducted with 30 min of front crawl swimming in water at different incremental intensities following the protocol outlined in [95] and led by an expert swimming trainer. To assess that swimming intensity was the same for all the subjects, heart rate monitors and subjective ratings of perceived exertion were used to guide the exercise bouts.

Each participant was tested before and after physiotherapy treatment with ten sessions lasting about 1 h each, based on targeted physical dry exercises designed to reduce shoulder joint pain and strengthen the relevant muscles according to previous literature [100]. Manual therapy techniques were first used to address pain. Then, active exercise was promoted to correct postural deviations; improve anterior chest musculature; address hypomobility of the thoracic spine, loss of joint mobility, or excessive joint mobility; stretch the posterior capsule; and improve strength and endurance of the rotator cuff and scapular stabilisers.

In particular, the endurance capacity of the pectoralis major muscle was addressed, as it was found to be impaired in previous research [95]. This approach aims to correct compensatory techniques that participants may have adopted in their sports movements due to pain or dysfunction related to the swimmer's shoulder condition.

3.6.2.3 Acquisition and Analysis

The movements were recorded using two wireless magnetoinertial sensors from the MTw Awinda Development Kit (Xsens Technology, Enschede, The Netherlands) capable of detecting 3D Euler angles. A modified version of the protocol proposed by Fantozzi et al. [73] was adopted: one sensor was placed on the flat portion of the sternum of the subject and used as a reference and the other one above the centre of the humerus and posteriorly using Velcro body straps. The sensors were connected to the host system via dedicated software (MT Manager 2022) allowing data collection at a sampling frequency of 100 Hz during the first and third phases of the protocol.

The objective of this study was to characterise shoulder kinematics, with a specific focus on analysing arm movements relative to the thorax. Given that stroke duration was synchronised with a metronome, our analysis concentrated on assessing the arm's movement amplitude in relation to the thorax according to flexion/extension, abduction/adduction, and internal/external rotation.

We identified and analysed individual strokes for each participant and type of rotation, excluding the initial and final strokes to avoid any potential inconsistencies. Since the temporal patterns of each rotation type were consistent across strokes, we calculated the mean amplitude by averaging it over the entire exercise period. This approach allowed us to obtain a representative measure of movement amplitude for each rotation type throughout the exercise.

To examine potential kinematic differences, we compared the mean amplitude values of the three angular rotations before and after the physiotherapy treatment, as well as before and after the fatiguing protocol. Statistical significance was assessed using the Wilcoxon signed-rank test for paired samples, allowing us to determine whether any observed changes were statistically meaningful.

All kinematic data were analysed using a proprietary software program developed in MATLAB® (R2024a).

This methodological approach aimed to provide a clearer understanding of how physiotherapy and fatigue affect shoulder movement mechanics, ultimately contributing valuable insights into rehabilitation and performance optimisation.

3.6.3 Results

Among the eight participants who gave their initial availability to participate in the study and physiotherapy treatments, three were excluded as primary impingement was suspected and they were referred to the orthopaedic surgeon for further evaluations. As such, five subjects entered the study protocol, aged between 21 and 27 years, training in swimming from 10 to 15 years, competing at the national level, and with a training volume between 10 and 12 h per week. During the subjective evaluation, participants reported suffering from shoulder pain from 1 to 4 years, with pain usually presenting during swimming or immediately after, usually resolving before the following day, and with a pain intensity ranging from 4 to 7. After the physiotherapy protocol, all the participants reported an improvement in overall symptoms; two of them reported a complete resolution of symptoms, while three of them reported a decrease both in the frequency of pain and its intensity (maximum 2 on 10 on a numeric rating of pain).

Figure 18 shows the mean amplitude trend of the three rotational movements in participants, both before and after physiotherapy treatment and the fatiguing protocol.

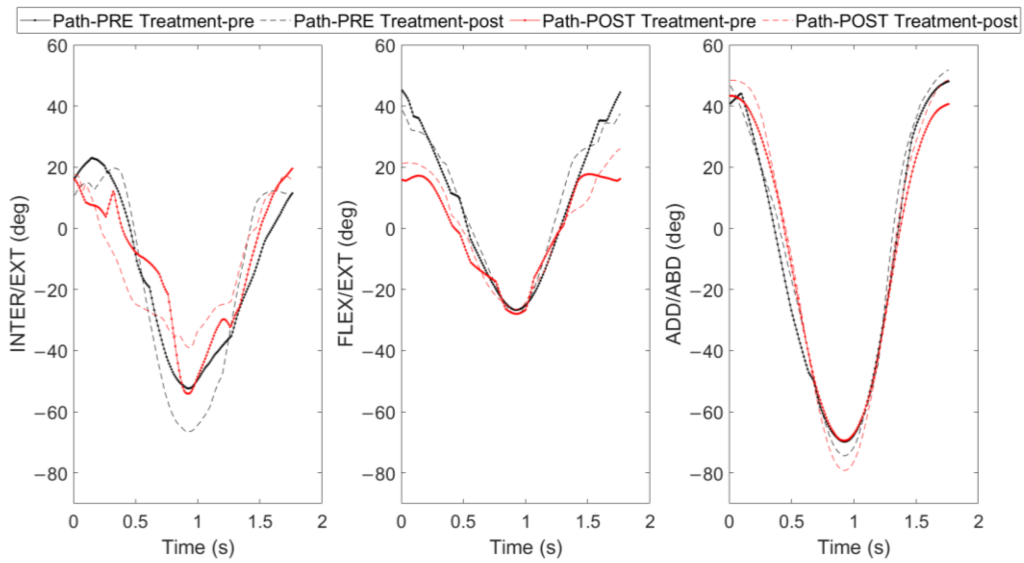


Figure 18. Mean amplitude trend of internal/external rotation, flexion/extension, and adduction/abduction before physiotherapy treatment (black) and pathological subjects after physiotherapy treatment (red), before (solid line), and after (dashed line) fatiguing protocol.

When comparing pre-treatment and post-treatment conditions before the fatiguing protocol, all three angular rotations exhibit similar patterns. However, a decrease in amplitude is noted in flexion/extension rotation. In contrast, when comparing the pre-treatment post-treatment conditions after the fatiguing protocol, flexion/extension rotation remains relatively consistent, adduction/abduction rotation shows a slight variation, and internal/external rotation presents a pronounced difference.

To confirm the observations above, Table 7 presents the mean values (±1 SD) of stroke amplitude for each angular rotation, both before and after the fatiguing protocol and physiotherapy treatment. The values underline a large variability among the participants.

Table 7. Mean values (±1 SD) of stroke amplitude [°] before and after physiotherapy and fatiguing exercise for each of the three rotations.

		Internal/External [°]	Flexion/Extension [°]	Abduction/Adduction [°]
Before Treatment	Before fatigue	80 ± 5	72 ± 28	124 ± 15
	After fatigue	92 ± 8	71 ± 34	131 ± 10
After Treatment	Before fatigue	80 ± 8	54 ± 22	126 ± 17
	After fatigue	81 ± 11	59 ± 21	134 ± 13

Figure 19 presents a box plot illustrating the distribution of amplitude values for each angular rotation, highlighting the median, 25th percentile, and 75th percentile values.

This visual comparison provides a clear overview of the changes in amplitude before and after the physiotherapy treatment as well as before and after the fatiguing protocol.

Each box plot captures the central tendency and variability for internal/external rotation, flexion/extension, and adduction/abduction amplitudes.

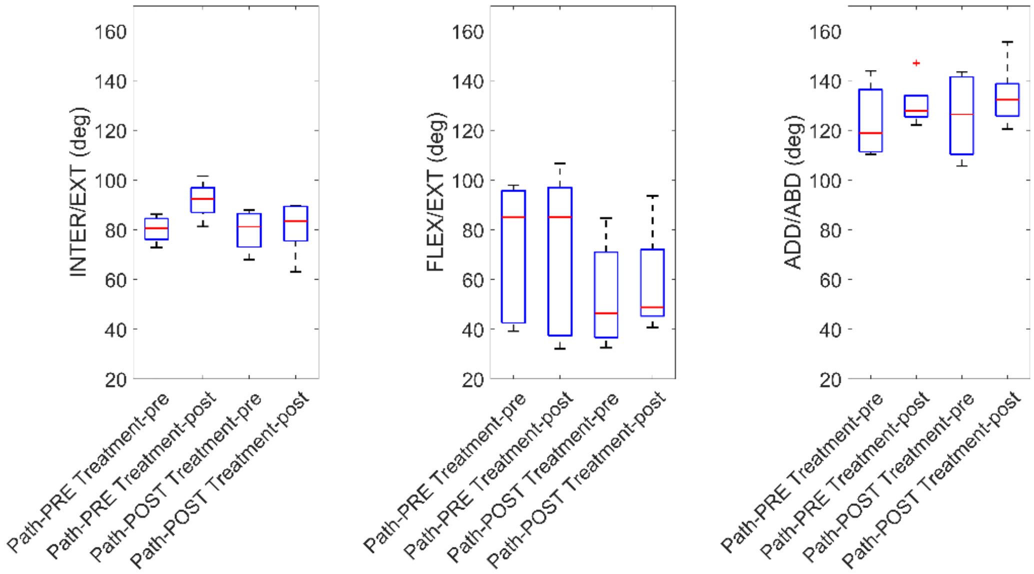


Figure 19. Box plots of the amplitude values for each angular rotation and condition (before and after the physiotherapy treatment and before and after the fatiguing protocol). The median values are in red with the box delimiting the 25th and 75th percentiles, and the red point indicates one outlier value.

3.6.4 Discussion

This preliminary study investigates the effects of a physiotherapy treatment and the impact of fatigue from swimming on shoulder kinematics, utilising two IMU placed on the sternum and humerus. The research focuses on individuals experiencing swimmer's shoulder pain, a prevalent and debilitating condition among elite swimmers. Swimmer's shoulder not only affects performance but also poses a significant risk to long-term shoulder health, highlighting the importance of developing effective diagnostic and rehabilitation strategies. These preliminary findings demonstrate that magneto-inertial sensors provide precise quantification of shoulder rotation angles and allow for detailed measurement of stroke amplitude during dry-land simulations of the front crawl exercise. This technology offers valuable insights into kinematic patterns, enabling a better understanding of how fatigue and physiotherapy interventions influence shoulder mechanics.

Swimming is a sport that requires high flexibility in terms of joint hypermobility, defined as the ability to move the joints beyond the typical range of motion [100]. From an anatomical perspective, this hypermobility is linked to impaired collagen, resulting in laxity of the connective tissue matrix. Such laxity compromises the stability of the joint capsule and increases the extensibility of ligaments, tendons, and skin, requiring a more intense stabilising muscular action. However, while joint hypermobility may confer certain advantages, such as enhanced flexibility and range of motion, it could lead to significant risks. Athletes with hypermobile joints are more susceptible to sport-related injuries due to potential deficits in muscle strength, greater muscular fatigue, and compromised joint stability [100]. Understanding these biomechanical dynamics is crucial for tailoring physiotherapy interventions aimed at minimising injury risks while optimising performance in elite swimmers [102], [103].

In particular, in these athletes, injuries are frequently attributed to specific adaptations in stabilising structures, resulting from the repetitive and intense demands of the sport. Initially, these adaptations seem to enhance the efficiency of the performance; however, over time, they can lead to alterations in shoulder biomechanics. It is important to note that such adaptations may appear not only after several years of practice but even after a single sports season [104]. Walker et al. [105] report that adolescent swimmers having an internal/external rotation of more than 100◦ could develop shoulder injuries, supporting the hypothesis that an optimal range of flexibility is necessary to swim without risking shoulder injury, and Behnam Liaghat et al. [106] found that young competitive swimmers with generalised hypermobility show both strength and fatigue deficits in medial rotation. The results of this study reveal a statistically significant increase ($p = 0.03$) in internal/external rotation amplitude in the pre-treatment condition when comparing values before and after the fatiguing exercise. Additionally, a reduction in amplitude was observed after the fatigue protocol when comparing the pre-treatment to post-treatment conditions. The findings about internal/external rotation are particularly intriguing, as these movements are markedly involved in front crawl swimming and swimmer’s shoulder [100]. The increase in internal/external rotation amplitude observed immediately after the fatiguing exercise may be attributed to the activation of muscle memory, developed through repetitive in-water training [107]. This muscle memory likely enables swimmers to maintain or even enhance movement efficiency during fatigue states. However, while such adaptations may initially aid performance, they could also reflect compensatory strategies that place additional strain on the shoulder joint, potentially increasing the risk of injury over time. Indeed, it has been previously suggested that a similar fatiguing task during swimming might result in altered muscular responses of the latissimus dorsi and pectoralis major muscles [95] that could, therefore, influence internal/external rotation amplitude; during the physiotherapy treatment, specific attention was directed to these muscles, trying to improve not only their strength and endurance capacity but also their pattern of activation. As such, it might be speculated that the reported differences might depend on the role of these muscles, how fatigue affects their activation, and how the treatment has improved their function and control.

The flexion/extension amplitude shows a large decrease after the fatiguing protocol, both before and after the treatment, with the post-treatment amplitude values being notably lower. Swimmers often exaggerate scapular protraction as a technique to achieve a wider and more powerful stroke and improve performance. While this adaptation may provide short-term benefits in the water, it places excessive strain on the shoulder complex, leading to pain and injuries. The primary goal of physiotherapy treatment in such cases is to target these biomechanical challenges by strengthening the stabilising muscles of the shoulder, helping to restore and maintain a healthy and functional range of motion. This approach not only helps to alleviate pain but also enhances movement efficiency, promoting proper biomechanics and significantly reducing the risk of future injuries.

The adduction/abduction amplitude values remain similar when comparing the pre-treatment and post-treatment measurements, with only a slight increase observed after the fatiguing protocol. Similar to the trends observed in internal/external rotation, this increase in amplitude could be attributed to muscle memory developed through the repetitive in-water movements characteristic of swimming.

The use of wearable sensors to assess shoulder kinematics has been proposed as a promising solution to provide the clinician and physiotherapist with reliable and quantitative data regarding the functional evaluation of the shoulder to support a better diagnosis and support injury prevention programs [108], representing an objective marker of rehabilitation efficacy as well [4]. Indeed, by providing such quantitative measures, it might be possible to better define the pathophysiological mechanisms underlying such a complex complaint that might manifest in a wide range of conditions and populations, from the elderly to the occupational setting and to elite athletes, with a great variety of symptoms.

Among them, sEMG, force sensors, IMU, accelerometers, fiber optic sensors, and strain sensors have been widely applied. With such techniques and their application within the “big data” context, machine learning might be implemented to further improve home rehabilitation by recognising the quality of performed physical exercises and possibly preventing disorders in patients’ movement [109].

As such, these devices might be useful to tailor the most optimal treatment to a clinical condition characterised by a heterogeneous clinical and symptomatic manifestation.

In swimming, the use of wearables has been implemented to provide a wide range of information, including athletic performance and training load, or swimming technique kinematics [110], [111], [112]; however, to the best of the authors’ knowledge, these are the first preliminary findings about the use of wearables, as IMU, to evaluate the efficacy of physiotherapy in the treatment of “swimmer shoulder” pain.

In conclusion, the use of magneto-inertial sensors allows for the identification of the specific movement components affected by the pathology, which could help provide a more targeted treatment. Due to the restricted sample size and the large variability, we found only one significant difference before and after the fatiguing exercise, indicating that fatigue impacts shoulder movement patterns. Additionally, a few differences approached statistical significance when comparing the pre- and post-physiotherapy treatment results.

Among the other limitations, the absence of a healthy control sample might have limited the interpretation of the current findings, as well as a comparison with the contralateral limb. Nevertheless, it should be reported that the proposed task is in line with some previous works addressing the “swimmer shoulder”. Moreover, to the best of the authors’ knowledge, this is the first study addressing the effect of a fatiguing task on these parameters and in this specific population, the latter being characterised by a selected form of shoulder pain.

Further investigation with larger sample sizes and more diverse fatigue protocols is needed to clarify the biomechanical implications of these subtle changes and to determine how physiotherapy interventions might optimise this aspect of shoulder kinematics in swimmers. Moreover, considering both sexes could determine if differences are present between male and female swimmers, as well as between different swimming techniques.

3.7 Shoulder tensiomyography and isometric strength in swimmers before and after a fatiguing protocol [95]

3.7.1 Aim of the study

The purpose of our study was to assess the effects of fatigue on shoulder muscles by concomitantly measuring isometric strength and TMG measures in healthy shoulders of young competitive swimmers before and after performing high-intensity swim training.

3.7.2 Methods

3.7.2.1 Participants

Volunteers were recruited from local swim clubs to participate in this cross-sectional study between January and April 2023. Of the 16 swimmers who volunteered, 14 swimmers (11 males and 3 females; age = 21 6 3 years [range, 17–26 years], height = 1.78 6 0.06 m, mass = 73.1 6 9.2 kg) who completed all assessments were included. Participant characteristics and training habits are reported in Table 8. Inclusion criteria were age between 16 and 35 years, participation in swimming for at least 3 years, a training volume of ≥ 4.5 h/wk, and use of the front crawl as the main swimming style. Participants were excluded if they had a history of acute or chronic muscular or joint disease of the tested shoulder or upper limb. All participants and their legal guardians provided written informed consent or assent, as appropriate, and the study was approved by the University of Trieste Ethics Committee (122/2022).

Table 8 Participant and training characteristics of the included sample (N=14)

Characteristic	Mean ± SD
Age, y	21 ± 3
Height, m	1.78 ± 0.06
Mass, kg	73.1 ± 9.2
Body mass index, kg/m^2	23.0 ± 1.6
Time training in swimming, y	13 ± 4
Training frequency, sessions/wk	6 ± 1
Training volume, h/wk	13 ± 2
	n (%)
Sex	
Female	3 (21)
Male	11 (79)
Competition level	
Regional	6 (43)
National or international	8 (56)

3.7.2.2 Study Protocol

All volunteers attended a first visit to the university laboratory, where we collected their characteristics, including information about training habits, and confirmed that they met the inclusion criteria. They familiarised themselves with the procedures of each assessment (ie, shoulder TMG and isometric strength evaluation). Training was repeated until participants were confident with the assessment techniques. They received a copy of the fatiguing protocol designed by the research team and an expert swimming trainer (A.C.), and they were recommended to try it in the following days. They were invited to the experimental session between 7 and 14 days after the first visit. The experimental session was conducted in a local swimming pool (depth = 25 m; water temperature range, 26°C–29°C). Measurements were collected in the morning at least 2 hours after participants woke. Participants were instructed to avoid strenuous training in the 48 hours before the study and to refrain from smoking and consuming alcohol the morning of the study. After arriving at the swim facility, they were asked to don their swimwear. The TMG and strength assessments were performed. Next, they performed the fatiguing protocol, consisting of 30 minutes of front-crawl swimming at different incremental intensities (Figure 1; see Supplemental Appendix, available online at https://dx.doi.org/10.4085/1062-6050-0265.23.S1); the overall exercise intensity was monitored with a waterproof heart rate monitor, with a heart rate of .70% the maximum for .90% of the training time and .80% for .30% of the training time. Participants were also asked to rate perceived exertion (rating of perceived exertion from 1 [least exertion] to 10 [greatest exertion]) during the different phases of the fatiguing protocol with respect to the training protocol, as reported in Supplemental Appendix and Figure 20. The TMG and isometric strength assessments were then repeated using the same measurement protocol within 20 minutes after completion of the fatiguing protocol. To reduce the influence of arm dominance, we randomly tested participants on their dominant or nondominant shoulder, and according to the absence of any shoulder pain symptoms. The dominant arm was defined as the arm usually used for throwing an object.

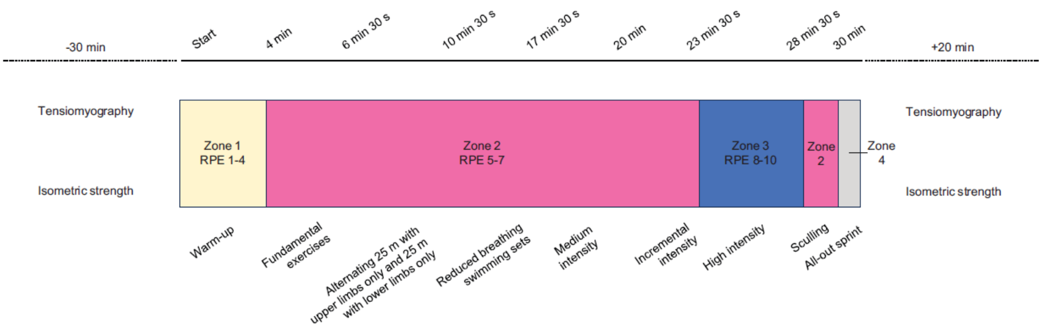


Figure 20. Representation of the swim-training protocol to induce fatigue. The outcomes assessment was performed before and after the 30-minute training protocol. Intensity zones from 1 to 4 and their corresponding rating of perceived exertion (RPE) are provided.

3.7.2.3 Instrumentation

TMG. The TMG is a mechanomyographic method that has been used in several studies and functions as a promising tool to assess in vivo skeletal muscle mechanical contractile properties; unlike other methods, such as myotonometry or shear-wave elastography, it is based on D_m after electrical stimulation, presenting an "active" response of the muscle [88], [113], [114]. A sensitive digital displacement sensor (TMG-BMC, Ltd) placed on the skin surface at the measuring site of the muscle of interest was used to record muscle-belly oscillations induced by a single 1-millisecond maximal monophasic electrical impulse. The stimulation amplitude was gradually increased from the minimum intensity to induce a recordable oscillation until the measured D_m during the twitch contraction (in millimeters) did not increase further, with electrical pulses ranging between 85 and 110 mA at a constant 30 V. To provide sufficient resting, we chose a 10- to 15-second interstimulation interval. The analysed measures were recorded from 2 maximal responses, averaged, and included in the final analysis. The standardized TMG-derived measures included the D^m (in millimeters), the time from electrical pulse to 10% of D_m (delay time [T_d]; in milliseconds), the time to contraction between 10% and 90% of D_m (contraction time

[T_c]; in milliseconds), the time when the response was .50% of D_m (sustain time [T_s]; in milliseconds), and the time between 90% and 50% of D_m during the muscle relaxation phase (half-relaxation time [T_r]; in milliseconds). All measures were extracted using TMG software (version 3.6.16) and used for offline analysis [114]. A smaller D_m has been suggested as an index of increased muscle stiffness, whereas a larger D_m implies lower muscle stiffness. Delay time represents a measure of muscle responsiveness. Contraction time represents the speed of twitch-force generation and has been reported to reflect muscle fiber type or tendon stiffness. Sustain time and Tr are the least studied measures, with Ts possibly being important for assessing muscle fiber fatigue status, although these last 2 measures require further investigation [115]. In particular, Martín-Rodríguez et al reported high reliability for T_c, T_d, and D_m and poor reliability for T_r [115] . Substantial evidence is available for the reliability of TMG in assessing exercise-induced muscle fatigue, although more studies are required to determine its accuracy and validity [116]. The following muscles were investigated according to their role in front-crawl swimming biomechanics and their validation with TMG according to the manufacturer's guidelines: anterior deltoid, medial deltoid, latissimus dorsi (mLD), pectoralis major (mPM), upper trapezius (mUT), middle trapezius, and lower trapezius (mLT) [117]. The participants were tested while sitting (anterior deltoid, medial deltoid, and mUT) and lying supine (mPM) and prone (mLD, middle trapezius, and mLT). The electrodes and sensor were placed by the same researcher (A.B.S.) with expertise in the use of TMG on the muscle area, and positions were marked with waterproof drawing ink to ensure the constant location of the electrodes on the skin over the repetitions.

Shoulder Strength. Isometric shoulder strength assessments were performed using a digital handheld dynamometer (model K-Pull; Kinvent) with participants in the supine position on a treatment bed. Shoulder flexion and extension were performed with the shoulder abducted to 140° in the scapular plane (30° anterior to the frontal plane), the elbow extended, and the forearm pronated. To assess external rotation and internal rotation, we positioned the arm in 90° of shoulder abduction with the forearm vertical and the elbow flexed to 90°. All starting positions were confirmed using a goniometer, and the dynamometer was aligned to be perpendicular to the forearm. These positions were chosen according to previous literature, suggesting that they are similar to the shoulder position during the relevant phases of the crawl swim stroke [118]. Swimmers were instructed to keep the trunk from moving during testing without any external stabilization; such a protocol and shoulder positions have been reported to have excellent intrarater reliability [118]. An experienced sport physiotherapist (M.M.) performed the strength measurements and instructed the participants on the assessment protocol. Before the measurements, all participants were allowed to perform a short warm-up and familiarize themselves with the positions and movements at submaximal effort. The different movements were assessed in a randomized order, and 2 repetitions of each strength test were performed with a rest period of 5 seconds between each repetition and 30 seconds between each test. Swimmers were asked to gradually build up to a maximal force, maintain this effort, then relax when instructed after a total of 5 seconds. Oral encouragement was provided to participants during testing to produce a maximal effort (make test), according to previous literature [118]. The maximal value (in newtons) was chosen and normalized by body mass.

3.7.2.4 Statistical Analysis

Outcomes are reported as mean, SD, count, and proportion (%), as appropriate. Two-tailed testing was performed. Normality testing using the Shapiro-Wilk test was performed for all datasets. Given that TMG T_r and T_s were not normally distributed in all the assessed muscles, we applied a log transformation for these variables. A repeated-measures analysis of variance was performed. Given the different muscles assessed with TMG, a 2-way repeated-measures analysis of variance was performed to examine the effect of time (before and after the protocol) and muscle (the 7 assessed muscles). A Greenhouse-Geisser correction was applied in case of lack of sphericity, and a Sidak correction was applied for post hoc analyses. The effect size was determined using partial η^2 (η_p^2). When we observed a main group effect, we computed simple main effects to compare each muscle independently. Finally, Pearson product-moment correlation coefficients were computed between the changes in isometric strength and TMG measures before and after the fatiguing protocol. We set the α level at .05. All statistical analyses were performed using SPSS (version 23; IBM Corp).

3.7.3 Results

3.7.3.1 Tensiomyography

An interaction effect of time and the tested muscle was found for Dm ($F_{6,78}$ = 2.504, P = .03, η_p^2 = 0.161). In particular, after the fatiguing protocol, reduced Dm was found in the mLD (-1.0 mm; 95% CI = -1.7, -0.3 mm; P = .007) and mPM (-1.4 mm; 95% CI=-2.4, -0.4 mm; P =.007). However, no differences were found for the other TMG derived measures (Figure 21; Table 10).

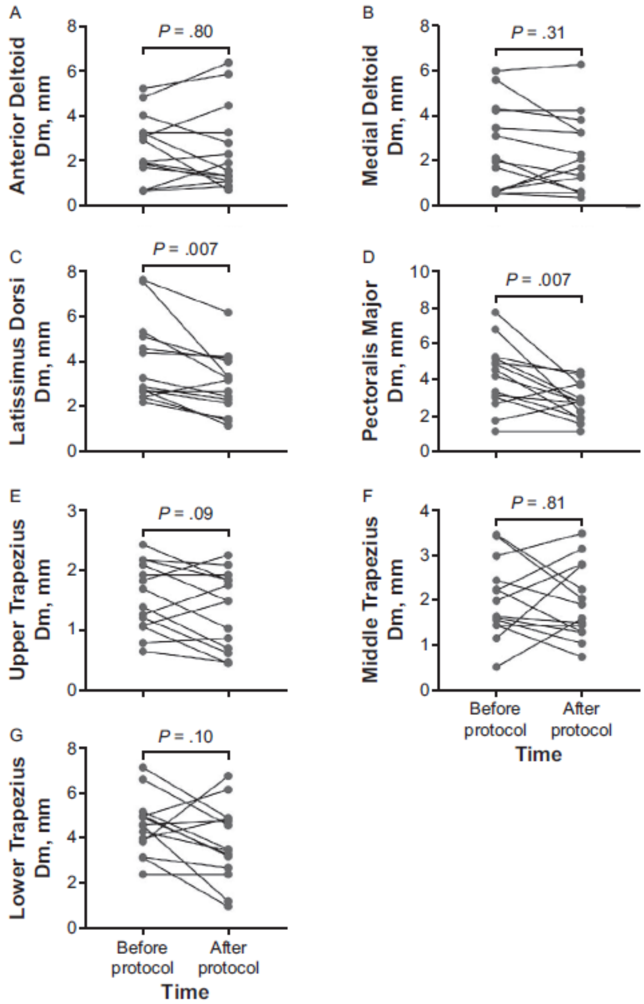


Figure 21. Radial displacement of the selected shoulder muscles before and after a fatiguing swimming protocol (N=14). We observed a time x muscle interaction ($F_{6,78}$ = 2.504, P =.03, η_p^2 = 0.161)

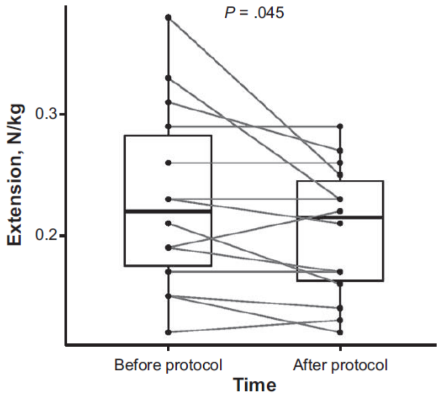


Figure 22. Isometric strength during shoulder extension before and after a fatiguing swimming protocol (N 5 14). We observed a time effect ($F_{1,13}$ = 4.936, P =.045, $\eta_p{}^2$ = 0.275).

3.7.3.2 Shoulder strength

Shoulder strength was affected by the fatiguing protocol, as a time effect was found during the extension task ($F_{1,13}$ = 4.936, P =.045, η^2 = 0.275). In particular, maximal isometric strength during extension was reduced by 0.03 N/kg (95% CI = 0.01, 0.05 N/kg; Figure 22). No effects were reported during flexion, external rotation, or internal rotation (Table 9).

Table 9. Shoulder Isometric Strength Before and After the Fatiguing Swimming Protocol (N=14; Mean ± SD)

Isometric Strength	Before Protocol, N/kg	After Protocol, N/kg	*P* Value
Flexion	0.22 ± 0.08	0.21 ± 0.08	.34
Extension	0.23 ± 0.08	0.20 ± 0.05	.045[a]
External rotation	0.28 ± 0.07	0.26 ± 0.06	.08
Internal rotation	0.32 ± 0.06	0.30 ± 0.07	.44

[a] Indicates difference ($P < .05$).

3.7.3.3 Correlation analysis

A correlation was found between the change in isometric strength during extension and mLD T_c (r = -0.544, P = .044), mLD D_m (r = 0.549, P = .042), and mUT T_c (r = 0.645, P = .01) and between isometric strength during flexion and mUT T_c (r = 0.683, P = .007).

Table 10. Muscle tensiomyography measures before and after the fatiguing swimming protocol (mean ± SD).

Measure	Before Protocol	After Protocol	*P* Value
Anterior deltoid			
Time of contraction, ms	16.4 ± 4.4	15.1 ± 2.2	.23
Time of delay, ms	17.8 ± 1.7	17.9 ± 1.5	.82
Time of relaxation, ms	34.3 ± 66.6	18.9 ± 16.7	.81
Maximal radial displacement, mm	2.6 ± 1.5	2.5 ± 1.8	.80
Time of sustain, ms	140.4 ± 255.1	171.6 ± 304.2	.92
Medial deltoid			
Time of contraction, ms	14.8 ± 3.2	14.5 ± 2.5	.69
Time of delay, ms	17.8 ± 1.9	17.4 ± 1.7	.21
Time of relaxation, ms	65.9 ± 150.1	15.3 ± 18.3	.19
Maximal radial displacement, mm	2.6 ± 1.9	2.3 ± 1.7	.32
Time of sustain, ms	140.3 ± 259.3	58.0 ± 111.7	.28
Latissimus dorsi			
Time of contraction, ms	42.1 ± 4.4	38.4 ± 6.0	.04[a]
Time of delay, ms	23.6 ± 3.5	22.8 ± 3.9	.60
Time of relaxation, ms	39.9 ± 15.6	31.3 ± 10.0	.01[a]
Maximal radial displacement, mm	4.0 ± 1.8	3.0 ± 1.4	.007[a]
Time of sustain, ms	105.2 ± 27.3	93.7 ± 15.4	.08
Pectoralis major			
Time of contraction, ms	24.9 ± 4.9	25.1 ± 5.6	.85
Time of delay, ms	25.2 ± 4.1	24.6 ± 5.7	.54
Time of relaxation, ms	35.4 ± 41.5	44.3 ± 56.8	.72
Maximal radial displacement, mm	4.2 ± 1.8	2.8 ± 1.0	.007[a]
Time of sustain, ms	67.6 ± 49.0	117.3 ± 195.2	.46
Upper trapezius			
Time of contraction, ms	20.5 ± 6.6	18.7 ± 3.0	.40
Time of delay, ms	19.6 ± 2.8	19.9 ± 3.7	.67
Time of relaxation, ms	69.3 ± 63.6	103.4 ± 199.6	.90
Maximal radial displacement, mm	1.5 ± 0.5	1.3 ± 0.6	.09
Time of sustain, ms	116.4 ± 74.4	162.4 ± 238.2	.79
Middle trapezius			
Time of contraction, ms	18.1 ± 3.5	18.2 ± 3.1	.85
Time of delay, ms	20.1 ± 2.7	20.1 ± 1.8	.96
Time of relaxation, ms	60.3 ± 44.7	77.1 ± 74.5	.98
Maximal radial displacement, mm	2.0 ± 0.8	1.9 ± 0.8	.81
Time of sustain, ms	111.2 ± 66.3	110.6 ± 84.7	.69
Lower trapezius			
Time of contraction, ms	41.2 ± 23.5	30.8 ± 12.7	.20
Time of delay, ms	22.8 ± 2.9	21.9 ± 1.5	.25
Time of relaxation, ms	101.6 ± 117.7	134.1 ± 228.7	.59
Maximal radial displacement, mm	4.6 ± 1.3	3.8 ± 1.7	.10
Time of sustain, ms	291.2 ± 192.9	318.4 ± 244.5	.83

[a] Indicates simple main effect ($P < .05$).

3.7.4 Discussion

Our study provides preliminary evidence that TMG could help detect skeletal muscle contractile changes after a fatiguing protocol in swimmers and provides an evaluation map of the most affected muscles. In particular, reduced D_m was reported in the overall assessed muscles, and the mLD and mPM were most affected. Although a reduction in T_c has been previously observed after a lower limb fatiguing protocol, the physiological mechanism underlying such alteration in the time needed to reach D_m remains unclear [119]. However, one might speculate that such reduced time could be associated with reduced D_m, as reported in this study [114]. Indeed, researchers have suggested that lower D_m, which is an indirect measure of muscle stiffness to electrical stimulation, could be observed after fatiguing tasks, and this might be due to the swelling response and increased intracellular water content after exercise-induced muscle fatigue, resulting in increased muscle stiffness [88], [114], [119], [120], [121]. More specifically, local fatigue has been reported to reduce Dm in skeletal muscles due to impaired propagation of the electrical stimulus along the sarcolemma, resulting from pH-driven alterations of the sodium and potassium gradient across the muscle membrane that influence excitation-contraction coupling [122]. Finally, Dm of the muscle might be further impaired by an accumulation of inorganic phosphate within muscle cells that results in reduced Ca^{2+} and subsequent excitation-contraction coupling or by direct accumulation of inorganic phosphate within muscle cells [114], [123]. In general, our findings seem to agree with those of some previous researchers suggesting that Dm could be reduced after exercise-induced muscle fatigue, whereas Tc presents conflicting results and requires additional research [116], [124]. The skeletal muscles acting on the shoulder joint during swimming that presented the largest changes postexercise were the mLD, mPM, and mLT, but only the values of the first 2 were statistically different. The mLD is primarily activated during the middle and late pull-through phases, whereas the mPM is mainly active during the early and middle pull through phases [125]. Electromyographic assessment of muscle fatigue during the 100-m front crawl showed decreased mean power frequency of these muscles [126]. The mLT plays an important role in scapula movement and positioning and dynamic scapula stability, and it might be important during swimming and impaired after a 3-minute maximal effort in swimmers [127], [127]. If the fatiguing exercise protocol in this study affected all tested muscles and, in particular, the mLD and mPM, given their important role in front crawl, the protocol's effect on the mUT and mLT compared with other muscles, if confirmed in other studies with larger samples, could inform about its contribution in swimming.

Shoulder strength was assessed during isometric tasks aimed at testing the effects of fatigue on flexion, extension, and external and internal rotation, providing values in line with those reported in previous literature [118]. Shoulder extension contributes to pulling the body over the upper extremity through the water in the front-crawl stroke [118]. By contrast, shoulder flexors might have a minor role in the front crawl, as no differences in flexion have been found between swimmers and healthy young adults [118]. In previous swimming research, Matthews et al reported that maximal strength during internal and external rotation was not affected by fatigue [90]. In our study, only extension was reduced after the fatiguing swimming protocol. Subscapularis muscles and the mLD are found to be more active during shoulder extension than flexion, whereas the supraspinatus, infraspinatus, deltoid, trapezius, and serratus anterior muscles are more highly activated during flexion than during extension [128]. In addition, the pectoralis muscles are part of the shoulder-stabilizing structures and are active in shoulder extension [129]. Regarding the trapezius, the mLT is mainly activated during flexion, although it might be active during extension [130]. Interestingly, the larger fatigue-induced TMG-related changes were found in muscles that might have a major role in shoulder extension, which showed an isometric strength reduction. Indeed, the correlation analysis suggested an association between the decrease in extension strength and decreased mLD Dm (ie, index of fatigue). Curiously, a longer time to contraction of the mLD and shorter time to contraction of the mUT were correlated with decreased extension strength, and a shorter time to contraction of the mUT was correlated with decreased flexion strength, confirming the above-mentioned conflicting results about this measure and fatigue. Taken together, our results are in line with those reported in the previous literature suggesting that TMG might help detect some hallmarks of muscle fatigue, and reduced Dm might reflect increased stiffness due to exercise-induced alterations of muscle contractile properties [120], [121]. Although such findings have been suggested in other studies, to the best of our knowledge, we are the first to report TMG alterations in shoulder muscles after a fatiguing swimming protocol [120], [121]. In addition, isometric strength was assessed during different shoulder movements, suggesting some potential associations between the muscles that presented with the most impaired TMG measures and reduced isometric strength. Despite such promising results, our study included a relatively small sample size, and other individual factors might have affected the outcomes; therefore, further studies are needed to confirm the proposed findings. Given the role of the tested muscles to stabilize and prevent pain and injuries, future research should be performed to evaluate the role of fatigue in the development of such conditions and to assess if "shoulder pain" could influence the reported outcomes in competitive swimmers. Exercise-induced muscle fatigue in swimming could be detected using TMG measures in the shoulder muscles and, in particular, as reduced Dm. The mLD and mPM were found to be the most affected muscles, although an effect of fatigue was reported overall for the tested muscles. According to the isometric strength assessment, only shoulder extension was reduced after the fatiguing exercise protocol. Our findings encourage the use of TMG as a noninvasive assessment tool to detect peripheral fatigue and also fatigue in shoulder muscles and could help athletic trainers and physiotherapists design training and rehabilitation protocols based on the most affected muscles and suggest specific exercises that might focus on the most affected muscles (eg, mPM and mLD), combining swimming and strength training [131].

3.8 Effects of swimming fatigue on neuromuscular parameters in young swimmers with unilateral shoulder pain [132]

3.8.1 Aim of the study

To the best of the authors' knowledge, no studies have previously investigated the effect of fatigue on shoulder muscles activation and isometric strength in comparing healthy swimmers with those characterized by unilateral shoulder pain. Therefore, this study aimed to evaluate the possible changes induced by a standardized swimming fatigue protocol on shoulder neuromuscular characteristics in terms of isometric strength and electromyographic activity, in swimmers with unilateral shoulder pain compared to healthy swimmers.

3.8.2 Materials and Methods

3.8.2.1 Participants

Across-sectional study was conducted on young swimmers of both sexes recruited from two local swimming clubs and who volunteered to participate in the study, between March and September 2023. To be included, swimmers competing at the national level (but not international), aged between 16 and 35 years, were asked to train in swimming for at least 3 years, not less than 4.5 hours per week in-water and 1 h dryland, front crawl had to be their main swimming style. Two swimming trainers (ACand LMa) assessed that similar training regimens were followed by all the participants. Participants were then categorized according to the presence (PAIN) or absence (controls) of symptoms related to unilateral shoulder pain. In particular, participants were suspected of suffering from unilateral shoulder pain if they subjectively reported manifesting it during or immediately after swimming, which had to be reported as being perceived in the neck-shoulder, and/or arm, for at least 3 months and intensity >4 (0-10). Furthermore, a group of physical therapists evaluated the subjects which were excluded if the presence of tissue pathology, high irritability or specific shoulder pathology such as primary impingement was found. Moreover, participants were excluded if they had a history of neck-shoulder surgery, previous fracture, bilateral shoulder pain, cervical radiculopathy, and recent (less than 12 months) physiotherapy or other treatment on the affected shoulder. For the physical assessment, the recommendations suggested by Tovin *et al*. [100] were considered, which suggest focusing on the impairments associated with the onset of symptoms such as hypermobility or instability of the glenohumeral joint, impaired posture, alteration of scapulohumeral rhythm, altered neuromuscular control or impaired rotator cuff strength. In particular, the physiotherapist evaluated the presence of altered shoulder range of motion or scapulohumeral rhythm, postural abnormalities and positive impingement tests. All the participants and their legal guardians read and signed informed consent and were instructed about the study procedures. The study was approved by the local university ethics committee (122/2022) and was conducted according to the Declaration of Helsinki principles.

3.8.2.2 Study protocol

During an initial session, demographics and anthropometrics were collected, and inclusion/exclusion criteria were confirmed. Additionally, all the participants were instructed about the training and evaluation protocol, including isometric strength assessment and both the “functional” and “strike” tasks. The training was repeated until the participant felt confident with the assessment techniques. The training/fatiguing protocol consisted of 30-minute front crawl swimming at different incremental intensities based on the rate of physical exertion (RPE) reported and heart rate. In particular, the protocol consisted of 4 min of warm-up exercise (zone 1; RPE 1-4; heart rate [HR] <70% maximum), 2.5 min of fundamental exercises, 4 min alternating 25 m only upper limbs/ lower limbs swimming, 13 min at different and progressing intensities (zone 2; RPE 5-7; HR=70-90% maximum), 5 min high-intensity swimming (zone 3; RPE 8-10; HR=90% maximum), 1.5 min sculling exercise (zone 2; RPE5-7; HR=70-90% maximum) and a final 100 m at maximum intensity swimming (zone 4; RPE 10) [133], [134]. 20 Between 7 and 14 days from the introductory visit, the experimental session was performed in a local swimming pool (25-m depth, water temperature between 26 °Cand 29 °C), in the morning, at least 2 hours after waking up. All experimental sessions were done on Tuesday to avoid training in the previous 48 hours, furthermore, participants were asked to refrain from smoking or alcohol consumption in the morning. After wearing their swimming suit, a 10-minute standardized dryland warm-up was performed, guided by the same swimming trainers, including arm swings in various positions. After the warm-up, neuromuscular assessment was performed before (pre-) and after the 30-minute fatiguing protocol (post-). The sEMG was recorded before the isometric strength assessment, first acquiring the maximum isometric voluntary contraction (MVIC) for each muscle considered and subsequently the muscular activations during the "functional" task and then during the "strike" task. Health controls were tested on their dominant or non-dominant shoulder according to the swimmers with unilateral shoulder pain, to match shoulder/arm dominance.

3.8.2.3 Surface electromyography

Muscle activation during the two different tasks has been recorded with Mini Wave sensors, a wireless sEMG system (Cometa srl, Milan, Italy). Data were stored in the manufacturer's software and then exported for offline analysis using MatLab (MathWorks, Natick, MA, USA). Ag/AgCl surface electrodes were applied to the skin, which was previously cleaned with alcohol tissue paper and eventually shaved. Electrodes were positioned aligned to the muscle fibers on the trunk and shoulder muscles of the painful side in the PAIN group and on the corresponding shoulder for the healthy controls selected based on previous literature: [135], [136] m.UT (upper trapezius), m.MT (middle trapezius), m.LT (lower trapezius), m.SA(serratus anterior), m.LD(latissimus dorsi), m.MD(middle deltoid), m.S (scalene), m.PM (pectoralis major), m.SCM (sternocleidomastoid). An experienced physiotherapist identified the muscles during manual muscle testing, according to guidelines and previous literature testing the reported muscles [136], [137]. Before any testing session (i.e., before and after the fatiguing protocol), sEMG was recorded during standardized 5 s maximum isometric voluntary contraction (MVIC), for each muscle. Three MVIC were performed, with 30 s of rest between each measurement, and the average of the three trials was used for normalization. The raw EMG signals were sampled at 2000 Hz, band-pass filtered (10-500 Hz) and after removing the ECG noise (according to the algorithm of Costa Junior et al.) [138], the RMS values were calculated following the characteristics of each task. The absolute RMS values were normalised with respect to the RMS values obtained during the MVIC for each muscle (%MVIC) [139].

3.8.2.4 Functional task and front-crawl strike task

After MVIC testing, the "functional" task and "strike" task, were performed on the painful shoulder in PAIN group, and corresponding in controls, as previously described to assess muscle activity in neck and shoulder pain [135], [136], [140]. Participants were asked to sit on an adjustable chair in front of a table in order to have their elbows aligned with the surface and both forearms resting on the table. On a paper positioned in the center of the table, circles with dimensions of 70 mm, which formed the corners of an equilateral triangle, were drawn with a distance of 23 cm between their centres. The participant was then asked to hold a pen on the painful upper limb, or corresponding in controls, and point it in the middle of the triangle to adjust the body distance to have a 30° arm flexion, as measured with a goniometer. Once the starting position was optimized, the participant was instructed to mark points with a pen within each of the three circles counterclockwise and in coordination with the metronome set at 88 beats/min. The sEMG signals were recorded during the functional task for a duration of 150 s, and the root-mean-square (RMS) values of EMG signals were calculated over the duration of each cycle of 3 consecutive positions. In each participant, the first and the last 10 cycles (about 20 s each) of the EMG recording were discarded and the RMS was averaged on the 54 cycles taken for the final analysis (similarly to Falla *et al.* and Nederhand *et al.*) [140], [141].

After the functional task, participants were asked to lie in the prone position on a physiotherapy table, in order to have the lateral edge of the chest aligned with the border of the table and a free range of movement of the tested shoulder. The head was rotated towards the tested side of the body during the whole protocol. Then, the participant was instructed to simulate a dry front crawl movement with the affected/corresponding upper limb in coordination with the metronome set at 60 beats/min to perform a rotation every 2 s. In particular, the assessed arm had to be over the head or along the body at each beat to perform one stroke every 2 beats. The sEMG was recorded during a simulated dry front crawl for 40 s, and the RMS values were calculated over the duration of the whole stroke. The first and last 2 strokes were discarded, and for each participant, the RMS was averaged on the 19 strokes considered.

Finally, in both tasks, the mean values (±1 standard deviation [SD]) of both controls and PAIN groups in pre-and post fatiguing sessions were separately calculated. The analysis was performed by using a proprietary program in MatLab (MathWorks). Considering the protocol used in this study, the standard error of the measurement was suggested to range from 6 to 24% in different muscles, in line with previous findings [142].

3.8.2.5 Shoulder strength

Isometric shoulder strength assessments were performed in the supine position on a physiotherapy table, with a digital handheld dynamometer (K-Pull, Kinvent, Montpellier, France), according to previous literature [134]. Shoulder flexion and extension were performed with the shoulder abducted at 140° in the scapular plane (30° anterior to the frontal plane) with the elbow extended and the forearm pronated. To assess internal rotation (IR) and external rotation (ER), the arm was positioned at 90° shoulder abduction with the forearm vertical and the elbow flexed to 90°. The IR:ER ratio has been also calculated as swimmers seem to present increased internal rotation compared to external rotation which could lead to muscle imbalance and reduction of glenohumeral stability [143]. An experienced sports physiotherapist performed all the isometric strength evaluations. A goniometer was used to check the correct positioning of the participants, and the dynamometer was aligned to be perpendicular to the forearm. These positions were chosen according to previously validated protocols to mimic the shoulder position during the front-crawl swim stroke [134]. Measurements were performed in a randomized order and two repetitions were performed for each strength test with a rest period of 15s between each repetition. Participants were trained to perform the assessment keeping the trunk from moving during testing and without any external stabilization. The swimmer was asked to gradually build up to a maximum force and maintain the effort, then relax when instructed after a total of five seconds. Verbal encouragement was provided during testing to produce maximum effort according to previous literature. Data were then normalized relative to body mass (%) according to previous literature [118]. Such protocol and shoulder positions have been previously suggested to provide excellent intra-rater reliability. Furthermore, the digital handheld dynamometer itself has good intra and inter-rater reliability, particularly for movements of the upper limb [144]. Considering the protocol used in this study, the standard error of the measurement ranged from 3 to 8% in different movements, in line with previous literature [145], [146].

3.8.2.6 Statistical analysis

The sample size was estimated based on the data from Sabzehparvar *et al*. [135] with an alpha probability of 0.05 and required power of 0.80, suggesting at least 7 participants had to be enrolled in each group. In this study we aimed to recruit 10 participants per group, with an a posteriori achieved power higher than 80% for the main findings. All statistical and primary analyses were performed with SPSS version 23 (IBM Inc., Armonk, NY, USA). Data were reported as means, SD, counts and proportions (%) as appropriate. Two-tailed testing was performed. Normality testing using the Shapiro-Wilk Test was performed for all datasets and non-normally distributed outcomes were log-transformed and further analysed. Analysis of variance (mixed-factors analysis of variance [ANOVA]) was performed: for isometric strength assessment, time (pre- and post-) and shoulder (affected and unaffected – corresponding for controls) were considered as within-group factors and group (pain and controls) was used as between-group factor. For the sEMG-derived parameters, ANOVA was applied considering time (pre- and post-) as a within-group factor, and group (pain and controls) as a between-group factor. In the event of a statistically significant interaction effect, *post-hoc* analysis was performed with Bonferroni's correction. In case of not significant interactions, significant time and group main effects were also reported. Greenhouse-Geisser correction was applied in case of lack of sphericity. The effect size for the ANOVA interaction was determined by partial eta squared (pη2) and was defined as small with values of <0.06, as medium with values between 0.06 and 0.14, and as large with values >0.14 [147]. In addition, the effect size of *post-hoc* differences were reported as Cohen's *d* (small = 0.2), medium = 0.5), and large = 0.8) [148]. Significance was set for P<0.05. The mean differences found in the *post-hoc* analysis were also considered with respect to the maximum standard error of the measure and were reported as not meaningful if lower than 15% for the isometric strength or 30% for the sEMG, in line with the recommended minimal detectable change [146].

3.8.3 Results

Ten swimmers with unilateral shoulder pain (23 y, range 17-27, 7 males 3 females) and ten healthy swimmers (22 years old, range 16-34, 6 males 4 females) were included in the study. Participants' demographics, anthropometrics and training habits are reported in Table 11. No significant differences were present between the two groups. Before the fatiguing protocol, the PAIN group reported a shoulder pain of 2 (range 0-4), whereas after the fatiguing protocol pain Numeric Rating Scale (NRS) was 6 (range 3-8). None of the controls reported pain in the shoulder before or after the fatiguing protocol.

Table 11. Demographics and training characteristics of the included sample. Data presented as means ± standard deviations. Independent samples t-test, p<0.05.

	Pain n=10	**Controls** n=10	**Significance**
Age [y]	23±3	22±5	0.523
Females [n (%)]	3 (30)	4 (40)	0.639
Body Mass [kg]	76.3±10.1	74.3±8.1	0.691
Body Height [m]	1.78±0.09	1.77±0.04	0.822
Years of swimming [y]	16±2	14±6	0.373
Training volume [h/wk]	13.4±2.9	13.0±0.5	0.824
History of shoulder pain [months]	18 (range 6-24)		

3.8.3.1 Isometric strength

A significant time × group × side interaction was found during internal rotation strength ($F_{1,18}$=5.704, P=0.028, $p\eta^2$=0.241) and IR:ER ($F_{1,18}$=6.743, P=0.018, $p\eta^2$=0.273) (Figure 23). In particular, in the PAIN group, the affected shoulder showed a significant internal rotation strength decrease from pre- to post- by 0.06 N/kg (95% CI: 0.018- 0.105, P=0.008, Cohen's *d*=0.483). Additionally, the PAIN group was characterised by lower internal rotation strength in the affected shoulder than in the unaffected shoulder pre- (-0.064 N/kg, 95% CI: -0.117 – -0.011, P=0.019, Cohen's *d*=0.556) and post- (-0.117 N/kg, 95% CI: -0.178 – -0.057, P=0.001, Cohen's *d*=0.996). Furthermore, in the affected shoulder, the PAIN group was characterised by lower IR strength than controls after the training by 0.127 N/kg (95% CI: 0.004 – 0.250, P=0.043, Cohen's *d*=0.988) (*Table 12*). No significant differences were found for the other movements. The effect size of the interaction effects and of the *post-hoc* analysis showed a large to moderate effect for all the significant parameters.

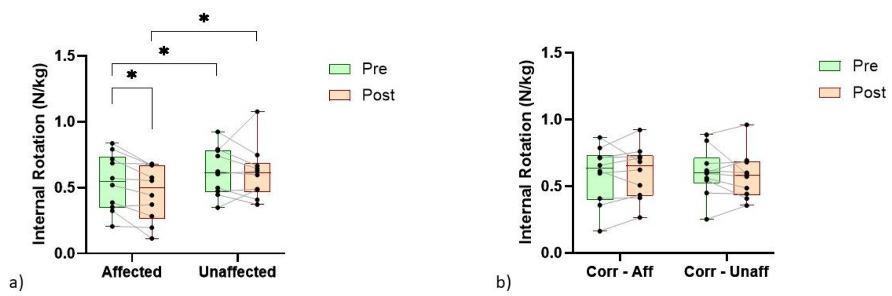


*Figure 23. Boxplots with individual data points representing isometric strength internal rotation of the affected or corresponding (Corr - Aff) and unaffected (Corr - Unaff) shoulder in (A) PAIN (N.=10) and (B) controls (N.=10), before (pre-; light gray [green in the online version]) and after (post-; dark gray [orange in the online version]) the swimming training protocol. ANOVA post-hoc analysis was performed for simple main factors. Corr: corresponding; Aff: affected; Unaff: unaffected; ANOVA: analysis of variance. *P<0.05; **P<0.01; ***P<0.001*

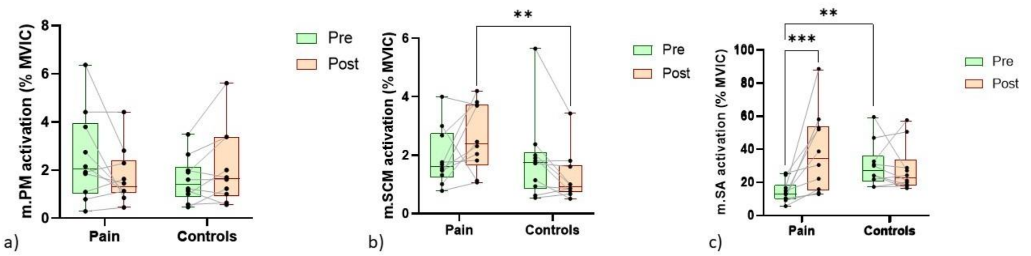


Figure 24. Boxplots with individual data points representing sEMG activation as percentage of maximum voluntary isometric contraction of the affected shoulder (PAIN) or corresponding (controls) during the functional task for (A) pectoralis major and (B) sternocleidomastoid, and during strike task for (B) serratus anterior in PAIN (N.=10) and controls (N.=10) before (pre-, green) and after (post-, orange) the swimming training protocol.
ANOVA post-hoc analysis was performed for simple main factors.
%MVIC: maximum voluntary isometric contraction; ANOVA: analysis of variance; m.PM: pectoralis major; m.SCM: sternocleidomastoid; m.SA: serratus anterior; sEMG: strength and surface electromyography.
**P<0.05; **P<0.01; ***P<0.001*

Table 12. Isometric strength outcomes in the included sample.
Data were presented as mean±SD. The effect size was reported as pη2. Mixed factors analysis of variance (within: time, within: side, between: group). SD: standard deviation; FL:EX: flexion:extension; IR:ER: internal rotation:external rotation; pre-: before fatigue training; post-: after fatigue training; Corr: corresponding.
*[a]Post-hoc analysis for significant difference between pre- and post-; [b]post-hoc analysis for significant difference between Pain and Control; [c]post-hoc analysis for significant difference between affected/corresponding affected and unaffected/corresponding unaffected shoulder, P<0.05; *statistically significant.*

	Pain (N.=10)		**Controls** (N.=10)		**Significance (pη2)**
	Affected	Unaffected	Corr. Affected	Corr. Unaffected	*Time x Side x Group*
Flexion (N./kg)					
Pre	0.277±0.095	0.300±0.112	0.315±0.108	0.300±0.107	p= 0.365 (0.046)
Post	0.255±0.105	0.298±0.112	0.302±0.097	0.281±0.091	
Extension					
Pre	0.267±0.104	0.301±0.118	0.331±0.088	0.347±0.090	p= 0.830 (0.003)
Post	0.251±0.091	0.280±0.087	0.321±0.300	0.321±0.091	
External rotation					
Pre	0.330±0.085	0.392±0.300	0.380±0.128	0.410±0.121	p= 0.541 (0.021)
Post	0.328±0.111	0.360±0.097	0.371±0.092	0.394±0.300	
Internal rotation					
Pre	0.385±0.131[a,c]	0.450±0.101[c]	0.435±0.145	0.448±0.110	p= 0.028 (0.241)*
Post	0.322±0.130[a,b,c]	0.441±0.108[c]	0.451±0.131[b]	0.432±0.111	
FL:EX					
Pre	1.08±0.29	1.02±0.26	0.97±0.27	0.87±0.25	p= 0.495 (0.026)
Post	1.01±0.28	1.09±0.35	0.96±0.19	0.89±0.23	
IR:ER					
Pre	1.21±0.50	1.12±0.23	1.15±0.22	1.12±0.19	p= 0.018 (0.273)*
Post	0.99±0.31b	1.26±0.26b	1.21±0.16	1.17±0.23	

3.8.3.2 Surface EMG

During the functional task, a significant time × group interaction was found for m.PM ($F_{1,18}$=5.783, P=0.027, $p\eta^2$=0.243) and m.SCM ($F_{1,18}$=7.482, P=0.014, $p\eta^2$=0.294) (Figure 24) (Table 13). Regarding m.PM, a tendency was found for a significantly decreased activation as an effect of fatigue only in the PAIN group by 0.8% MVIC(95% CI: 0.0-1.6, P=0.053, Cohen's *d*=0.440); in contrast, after fatigue, controls decreased the m.SCM activation by 0.6% MVIC (95% CI: -0.1-1.3, P=0.086, Cohen's *d*=0.485) whereas the PAIN group increased it by 0.7% MVIC (95% CI: 0.0-1.4, P=0.055, Cohen's *d*=0.633). In addition, the m.SCM was also characterised by higher activation after the fatiguing task in the PAIN group compared to controls by 1.4% MVIC(95% CI: 0.4-2.4, P=0.008, Cohen's *d*=1.225).

During the "strike" task, a significant time × group interaction was found for m.SA ($F_{1,18}$=9.514, P=0.006, $p\eta^2$=0.346), suggesting that the muscle activation increased after the fatiguing protocol only in the PAIN group by 24.2% MVIC (95% CI: 11.4-36.9, P=0.001, Cohen's *d*=1.356) (Figure 2). The effect size of the interaction effects and of the *post-hoc* analysis showed a large to moderate effect of shoulder pain on the fatigue response for muscle activation of m.PM, m.SCM and m.SA.

*Table 13. Surface EMG outcomes in the included sample. Data were presented as mean±SD. Muscle activation as a percentage of maximal isometric voluntary contraction [% MVIC] during the "functional" task and the "strike" task. Mixed factors analysis of variance (within: time, between: group). The effect size was reported as pη2. SD: standard deviation; m.UT: upper trapezius; m.MT: middle trapezius; m.LT: lower trapezius; m.SA: serratus anterior; m.LD: latissimus dorsi; m.MD: middle deltoid; m.S: scalene; m.PM: pectoralis major; m.SCM: sternocleidomastoid. [a]Post-hoc analysis for significant difference between pre- and post-; [b]post-hoc analysis for significant difference between Pain and Control, $P<0.05$; *statistically significant*

	Pain n=10		**Controls** n=10		**Significance**	
	Functional	Strike	Functional	Strike	*Time x Group Functional*	*Time x Group Strike*
m.UT [%]					*p=0.137 (0.118)*	*p=0.616 (0.014)*
Pre	14.0±10.6	28.6±9.5	13.1±13.8	23.6±10.0		
Post	22.5±15.8	27.4±4.9	11.2±5.1	25.4±4.2		
m.MT [%]					*p=0.810 (0.003)*	*p=0.353 (0.048)*
Pre	6.0±4.4	59.7±20.9	5.1±2.3	42.8±13.6		
Post	5.0±2.6	53.9±18.9	3.8±1.6	44.9±11.3		
m.LT [%]					*p=0.430 (0.035)*	*p=0.369 (0.045)*
Pre	5.1±1.6	37.3±10.5	5.2±2.0	36.6±10.0		
Post	6.6±4.4	36.6±15.9	5.4±1.5	41.9±10.1		
m.SA [%]					*p=0.500 (0.026)*	*p=0.006 (0.346)**
Pre	17.9±14.4	14.3±6.4[a,b]	22.5±15.7	30.4±13.4[b]		
Post	17.3±8.6	38.5±24.4[a]	18.0±8.5	28.1±14.2		
m.LD [%]					*p=0.629 (0.013)*	*p=0.516 (0.024)*
Pre	4.5±3.5	41.7±13.9	4.2±6.1	33.4±19.3		
Post	5.5±5.2	49.4±20.9	4.3±2.9	48.1±22.5		
m.MD [%]					*p=0.407 (0.039)*	*p=0.233 (0.078)*
Pre	8.5±7.8	47.6±10.3	8.4±10.4	54.7±12.3		
Post	7.2±3.5c	50.2±18.3	3.8±1.7	47.8±17.3		
m.S [%]					*p=0.117 (0.131)*	*p=0.676 (0.010)*
Pre	7.7±4.2	15.6±5.1[c]	7.6±5.5	10.2±5.1		
Post	7.0±4.1	20.6±17.1	12.9±10.0	12.5±7.4		
m.PM [%]					*p=0.027 (0.243)**	*p=0.977 (<0.001)*
Pre	2.5±1.9	3.9±3.1	1.6±0.9	3.3±1.9		
Post	1.8±1.2	3.3±2.0	2.1±1.6	2.7±1.4		
m.SCM [%]					*p=0.014 (0.294)**	*p=0.115 (0.132)*
Pre	1.9±1.0	3.7±1.6[a]	1.9±1.5	3.9±1.8		
Post	2.6±1.2[c]	8.1±4.6[a]	1.3±0.9	5.3±2.8		

3.8.4 Discussion

To the best of the authors' knowledge, this is the first study investigating the effects of fatigue on neuromuscular characteristics in swimmers with unilateral shoulder pain, compared with healthy controls. In particular, a functional upper-limb task was used, as previously suggested in the literature; [135], [136], [140], [141] additionally, a front crawl strike repeated movement performed out of the water was also performed.

The main findings from this study found that during the functional upper-limb task, fatigue induced a moderate reduction of m.PM activation in the PAIN group, as shown by the medium effect size, as well as a moderately increased activation of the m.SCM in the same group. For the controls, only m.SCM showed altered activation after fatigue, as a moderate decrease in activation. Considering this opposite behaviour of m.SCM, a large effect size suggested a higher activation after the fatiguing task in the PAIN group compared to controls. During the simulated "strike" task, the most affected muscle was the m.SA, which showed a large increase of its sEMG activity after the fatiguing task only in the PAIN group. Considering strength, internal rotation was mostly affected, as fatigue induced a moderate reduction of isometric strength in the PAIN group on the affected shoulder, leading to a largely significantly lower strength compared both to the unaffected shoulder and to the controls. None of these suggested differences was lower than the cut-off percentage for the standard error of measurement (sEMG from 32% to 169%, strength from 18% to 39%). The results of the present study are in line with previous research that found an augmented activation of the neck flexors muscles in swimmers with shoulder pain during a functional upper-limb task [140], [149]. Such increased activation of these muscles could suggest the presence of an altered pattern of activation of the superficial neck flexors in this population that has been previously hypothesized in other studies due to the adaptation pain model [150], [151]. In particular, in our study, m.SCM increased its activation compared to healthy controls after the fatigue protocol, in line with previous literature suggesting a hyperactivation of the SCM muscle during the same functional task in other neck-shoulder pain syndromes [140]. The SCM has the insertion on the clavicle, bone-related with the scapulae movement, so a hypothesis could be that altered kinematics of the scapulae could lead to an altered position of the clavicle and activation of the SCM [152], [153]. Furthermore, SCM is an accessory respiratory muscle and in particular, during the in-water fatigue protocol it could be particularly involved to maintain the correct respiratory pattern and head rotation [152], [153], [154]. In a previous study, an increased UT activation was reported during the same functional task among swimmers with a painful shoulder [135]. Even this muscle has its insertion on the clavicle, and it is involved in the upward rotation of the scapula, so even in this case, pain or altered kinematics could lead to an alteration in this muscle activity levels [79], [135], [155]. In our study, we found only a tendency to significance in an augmented activity of UT in the affected group after the fatigue protocol, which might be due to the limited sample size of the study. However, increased activation of UT in this population during this task is still debated as a similar study did not report altered activity of this muscle [136].

During the repeated strike task, SA was characterized by its sEMG activation being increased in the shoulder pain group after the fatigue protocol and it could be due to maintaining the compensation pattern during the swimming fatiguing protocol. A similar finding was also reported in previous literature during functional task [135]. SA, such as UT, has an important role in the upward rotation of the scapula during overhead upper limb movements [156]. Furthermore, this muscle ensures the posterior tilt and the external rotation of the scapula, accessory movements that allow the correct orientation of the glenoid fossa during the elevation of the upper limb resulting in functional contact between the humerus and the scapula, therefore promoting the correct kinematics pattern [157]. The altered stability of the glenohumeral system represents a risk factor for shoulder disorders; in fact, an alteration of the coordination of the shoulder muscles activity could lead to a translation of the humeral head by the centre of the glenoid cavity during upper limb movements [158]. Muscles such as SAand UT, have an important role in maintaining glenohumeral joint stability, as an alteration of the activity of these muscles could lead to dysfunctional scapula-thoracic kinematics resulting in the consequent alteration of the scapula-humeral kinematics [156], [157].

Consistently with the above-reported electromyographic alterations found in this study, decreased isometric strength during internal rotation of the shoulder after fatigue protocol was found in the affected arm of the swimmers with unilateral pain. Indeed, PM, which is the principal internal rotator of the shoulder, was characterised by reduced activation after fatigue in the same group. In addition, other previous studies have suggested the involvement of internal rotators of the shoulders in swimming-associated fatigue as a decreased range of motion after training was reported [159], [160], as well as reduced radial displacement, an indirect measure of muscle stiffness, evaluated with tensiomyography [161], [162]. Our study is in line with these previous findings, suggesting an increased difference in internal rotation of the affected side after the fatigue protocol.

A possible pathophysiological hypothesis of the reported neuromuscular alteration observed in this study could rely on the pain adaptation model [151]. In particular, this model suggests that pain could lead to a decrease in agonist muscle activity in favour of an increase of the antagonist muscles' activation, determining loss of strength and slower movement [151]. As such, despite the non-uniform findings regarding muscles activation changes in swimmers with shoulder pain, it might represent a physical manifestation of ischemia and pain, and at the same time it might lead to perpetuated pain and altered motor control [136]. The differences between some of the results of the present study with previous literature could be explained by the high heterogeneity of findings in the different studies [135], [136] which might depend also on the different clinical characteristics of the participants as previously suggested [135]. In particular, the participants in the present study reported pain principally during and after the training protocol with an increased reported pain after the protocol, whereas participants in previous research presented shoulder pain typically before exercise [135], [136].

3.8.5 Limitations of the study

Among the limitations of this study, the electrodes were not waterproof, so to provide corresponding positioning of the electrodes before and after the swimming training a waterproof marker was used to mark their position on the skin. To reduce the possible crosstalk from nearby muscles and contamination of signals because of muscles movement as previously reported [92], [118], a biomedical engineer with experience in electrophysiological signal analysis reviewed the traces. Another limitation of this study was represented by the presence of a restricted sample size of only 10 subjects per group, despite they were very similar in terms of age category and sex. Finally, participants in the unilateral shoulder pain group were included based on the subjective reporting of pain during the sport activity and a physiotherapy evaluation, as previously suggested [100]. However, despite following the above-mentioned guidelines for the evaluation should limit the heterogeneity of the clinical manifestation, some different clinical conditions could also underlie such pain perception; therefore, different pathophysiological mechanisms might be implicated. In addition to increasing the sample, future studies could investigate if sex differences might be present between and within the groups. Taken together, these findings suggest that shoulder pain in swimmers could be linked to altered neck flexors and shoulder internal rotator muscles, and these alterations are present after training when pain manifests. These findings are characterised by medium to large effect sizes, although the sample still is small and should be increased in future studies. Since the complexity of the clinical and pathophysiological characteristics of this condition, the results of the literature are conflicting and encourage future studies in this population. To limit the influence of a sampling based on poor inclusion criteria and therefore an alteration of the possible results obtained, in the present study we tried to implement a more rigid selection of the sample based on the indications of Tovin *et al* [100]. The results from the present study could support the development of new pathophysiological hypotheses and provide part of the groundwork for the development of novel and tailored prevention and treatment protocols. In particular, the results obtained in the present study could suggest the need to improve the perception of the sporting gesture in swimmers in order to work more with the rolling of the trunk, reducing the energy demand of the neck and serratus anterior muscles. Abetter activation pattern of muscles such as serratus anterior and pectoralis major could also reduce the effects of fatigue in terms of the effectiveness of the gesture.

Furthermore, regarding the strengths of this study, the consistency between the strength and electromyography findings of the shoulder internal rotators sustains a possible role of these muscles in swimming performance and injuries. Future studies are encouraged with similar protocols, including muscle activation analysis during swimming, in larger samples with a more accurate classification of the clinical condition.

3.9 Climbing

Over the past few decades, climbing has experienced a marked rise in popularity, particularly among younger populations, a trend further accelerated by its inclusion in the Olympic Games [163]. This surge in participation has been accompanied by a predictable increase in injury incidence. Epidemiological data indicate that 77.1% of climbing-related injuries affect the upper limbs, 17.7% involve the lower limbs, and 5.2% occur in the head, neck, and chest regions. Within the upper limbs, fingers (41.2%) and shoulders (20.2%) represent the most commonly affected anatomical sites [163], [164], [165].

From a biomechanical perspective, climbing places unique demands on the shoulder joint. It must accommodate an extensive range of motion to reach distant holds while simultaneously generating and transmitting substantial force, particularly during overhanging manoeuvres where the climber's body weight is largely supported by the upper limbs [166]. Any compromise in this balance, whether through pain, restricted mobility, or impaired force production, can severely limit climbing performance and increase injury risk. Repetitive, high-amplitude, and sustained upper-limb movements on vertical or overhanging terrain make climbers especially vulnerable to shoulder pathologies, including rotator cuff injuries, labral tears, and instability syndromes [163].

In this context, electromyography (EMG) has emerged as a valuable investigative tool in climbing research. Current studies have predominantly examined muscle activity levels and fatigue responses in the forearm and arm musculature, given their essential role in grip strength and endurance [167].

However, relatively few investigations have explored shoulder muscle activation in climbing-specific tasks, despite its pivotal role in stabilising the glenohumeral joint and ensuring efficient force transfer. Expanding EMG research to include the rotator cuff and scapular stabilisers could therefore provide deeper insight into both performance optimisation and injury prevention strategies.[167].

3.10 Shoulder muscle activation during standard climbing movement: an EMG analysis in individuals with shoulder pain

3.10.1 Aim of the study

This preliminary study investigates shoulder muscle activation with surface Electromyography (sEMG) during two standard climbing movements in individuals with shoulder pain, evaluates the impact of physiotherapy treatment on muscle activation patterns, and examines how the pathology influences muscle function. To the best of the authors' knowledge, this is the first study to focus on the muscular activity of the shoulder during climbing in painful conditions.

3.10.2 Material and Methods

3.10.2.1 Participants

A total of 8 athletes (3 females and 5 males with dominant right limbs and an average age of 29.9 ± 6.1 years) were recruited from a local climbing gym for participation in this study. The exclusion criteria encompassed conditions such as physical disabilities or sensory disturbances. Conversely, the inclusion criteria were designed to identify individuals aged between 18 and 45 years who experienced persistent shoulder joint pain and exhibited symptoms of algal disease for a minimum duration of four weeks. Additionally, participants were required to have a consistent training regimen, with an average weekly training volume of at least four hours, ensuring a suitable level of physical activity for the study's focus.

Before enrolment, all volunteers underwent a thorough evaluation conducted by a team of expert physiotherapists. This assessment was essential to ensure that participants met the study's inclusion criteria and to evaluate their overall condition and readiness for participation.

The purpose of the study, along with a detailed explanation of its potential risks and benefits, was communicated clearly to all participants to ensure informed decision-making. Ethical approval for the study was granted by the local university ethics committee, referenced under approval number 122/2022. According to ethical standards for research involving human subjects, all participants provided their written informed consent before being formally enrolled in the study.

3.10.2.2 Study protocol

We analysed the muscular activity of the most involved muscles in shoulder movement and stabilisation during climbing: medial deltoid, pectoralis major, latissimus dorsi, upper trapezius, lower trapezius, and anterior serratus. Before starting the task performance, the maximum voluntary contraction (MVC) was recorded, isolating each muscle with the overcoming isometric method [168]. Participants were then instructed to complete two distinct tasks involving climbing techniques. In the first task, they performed three repetitions per side of a frontal movement (Figure 25 A and B). Starting from a resting position, with both hands on the same hold, arms extended, legs bent, feet positioned on two holds, and the body aligned with the wall, the participants spread their legs and pulled with their arms to reach the next hold with one hand. In the second task, participants performed three repetitions per side of a flagging movement (Figure 25 C and D). Beginning in a resting position with both hands on the same hold, one leg bent with the foot on a hold, and the other leg extended along the same direction, participants extended the bent leg to grasp the next hold while swinging the other leg outward to the opposite side of the body.

At the beginning, the resting position is held for 5 seconds, followed by maintaining each subsequent position for 3 seconds. Finally, the resting position is held again for 5 seconds at the end of the task. With the assistance of a climbing instructor, the hand and foot holds were standardised to accommodate participants of varying heights, ensuring that everyone could perform the task effectively.

Each participant was assessed both before and after undergoing physiotherapy treatment of 10 sessions focused on active exercises based on the chaos-control continuum protocol [169]. This rehabilitative approach was chosen based on the specific sport practised by the athletes in the study, a sport that involves a certain level of chaos, characterised by dynamic movements, a high error rate, and significant energy expenditure. The protocol aims to progressively replicate this chaos through targeted exercises, transitioning from controlled conditions to increasingly complex and unpredictable scenarios. The treatment was divided into five progressive phases, each characterised by a decreasing level of control during sessions: high control, moderate control, control transitioning to chaos, moderate chaos, and high chaos. Exercises in each phase were tailored to the corresponding level of control and chaos. Initially, the focus was on recruiting the scapular muscles by teaching participants how to isolate their activation and perform exercises to improve shoulder mobility. As the treatment progressed, the exercises demanded greater muscular activation and enhanced awareness of scapular activation and positioning during various activities.

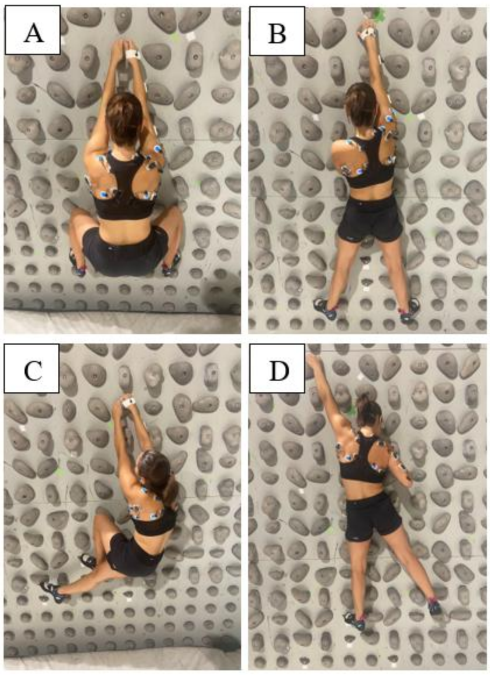


Figure 25. Visual representation of frontal (A start position and B final position) and flagging (C start position and D final position) movement.

3.10.2.3 Acquisition and analysis

Muscle activity was recorded using surface electromyography (sEMG) with a WaveX system (Cometa System, Milan, Italy) at a sampling frequency of 2000 Hz. The wireless sensors communicate with a receiver connected to dedicated software (Emg and Motion Tools, version 8.11.0.0) for data collection. Each sensor can operate in EMG mode, IMU mode, or both. Four sensors were configured as IMU and attached to the hand, forearm, upper arm, and chest using double-sided tape. Six sensors were set to EMG mode and placed on the following muscles: medial deltoid, pectoralis major, latissimus dorsi, upper trapezius, lower trapezius, and anterior serratus. The physiotherapist prepared the skin to optimise signal quality before placing the electrodes (Ambu BlueSensor M, 40.8 × 34 mm). Electrodes were secured over the muscle belly using adhesive film.

Using the signals from the IMU sensors, specific positions during task execution were identified: the initial resting position (1), arms bent in traction with leg extended (2), and arm fully extended while grasping the hold with the hand (3). For each interval and each muscle, the percentage of muscle contraction relative to the MVC was calculated and then averaged across the three repetitions for each task. We categorised the subjects into four groups based on the condition of their shoulders (right healthy RH, right painful RP, left healthy LH, and left painful LP) before (PRE) and after (POST) physiotherapy treatment. Comparisons were made both before and after physiotherapy treatment, as well as between the healthy and painful shoulders.

All the data were analysed through a proprietary program in MATLAB®. Statistical significance was evaluated using the Wilcoxon rank sum test for paired samples.

3.10.3 Results

The group comparisons revealed significant differences primarily in the lower trapezius, pectoralis major, and anterior serratus. Muscle contraction was analysed for each muscle across all intervals, with values compared between the groups in both tasks. Figure 26 and Figure 27 present boxplots showing the median, 25th, and 75th percentiles of the percentage of muscle contraction for groups presenting significant differences, respectively for the first and the second tasks.

For the first task, the lower trapezius (Figure 26 A and B) demonstrated a significant difference (p=0.029) in the RH group before treatment and after treatment in positions (1) and (2), respectively. Similarly, the pectoralis major (Figure 26 C and D) showed a significant difference (p=0.0286) in positions (2) and (3) between the LP and RH groups before treatment. Near-significant differences (p=0.057) were observed for the anterior serratus in position (1) before and after treatment in the LH group, as well as between the RP and LH groups after treatment. Additionally, the lower trapezius in position (3) showed near-significant differences (p=0.057) in the LP group between pre- and post-treatment.

For the second task, the lower trapezius (Figure 27 A) exhibited a significant difference (p=0.029) for the position (1) between the LP group before and after treatment. In position (3), the latissimus dorsi (Figure 27 B) showed a significant difference (p=0.0286) between the LP and RH groups before treatment. Furthermore, after treatment, the pectoralis major (Figure 27 C) displayed a significant difference (p=0.029) between the LP and RH groups. Near-significant differences (p=0.057) were also noted in the lower trapezius in position (3) before and after treatment in RH, as well as in LP groups (Figure 27 D).

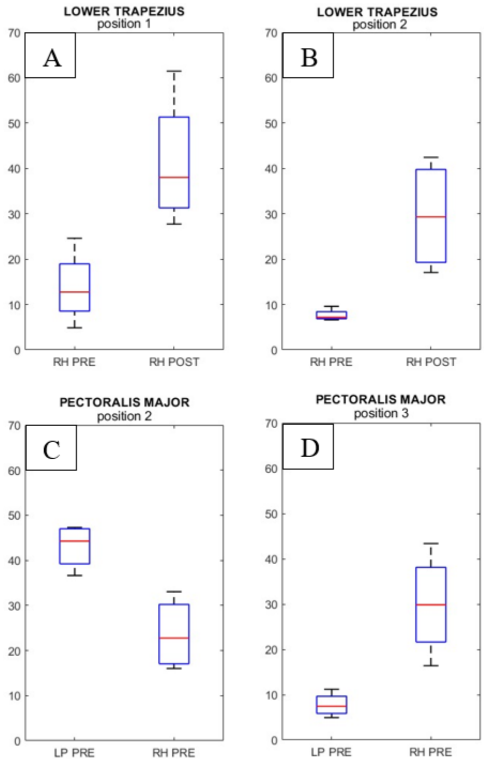


Figure 26. Box plots of the percentage muscle contraction values for groups with significant differences during task 1. The median values are highlighted in red with the box delimiting the 25th and 75th percentiles.

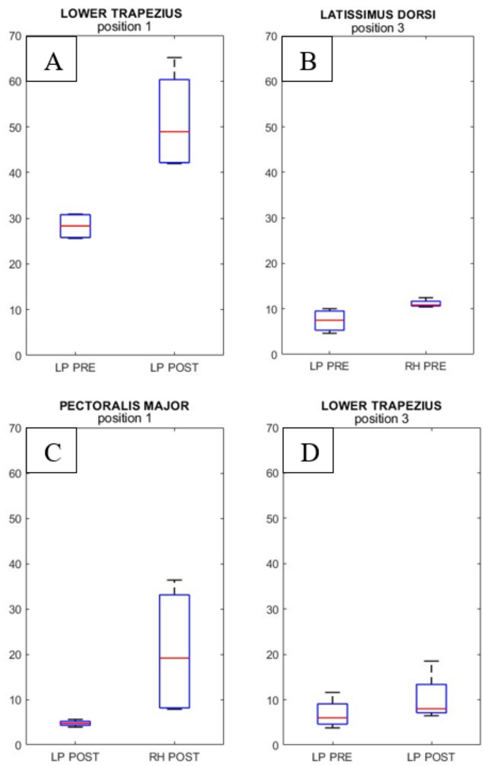


Figure 27. Box plots of the percentage muscle contraction values for groups with significant differences during task 2. The median values are highlighted in red with the box delimiting the 25th and 75th percentiles

3.10.4 Discussion

The physical effort required by climbing often leads to injuries, especially in the finger and shoulder joints. In this preliminary study, we aim to analyse shoulder muscle activation using sEMG during frontal and flagging movements. In response to pain, the body compensates through motor and postural adaptations to minimise strain on the affected shoulder while preserving movement efficiency [170], [171].

Based on the results for frontal movement in positions (1) and (2), the lower trapezius exhibits significantly higher contraction values for RH after physiotherapy treatment than before (Figure 26 A and B). Similarly, LP exhibits a comparable pattern, though the differences observed following treatment are not statistically significant. This may indicate an adaptive response to improve scapular control in the painful shoulder, which also influences the healthy side, enhancing symmetry in muscle recruitment. Comparing LP and RH before treatment (Figure 26 C and D), the pectoralis major has a different behaviour in position (2) and at the end in position (3). The pectoralis major is one of the main muscles used in the frontal position and is also fundamental for traction and shoulder stability. At the end of the movement, the contraction of the painful shoulder decreases, probably because it loses strength and control. The treatment made the recruitment of the pectoralis major more homogeneous between these two movement phases. In fact, even if not significant, the painful shoulder reduces initial activation and works better in the final; this may indicate an improvement in muscle coordination.

In the flagging movement, a significant increase in lower trapezius contraction is observed in position (1) when comparing LP before and after treatment (Figure 27 A). This muscle plays a crucial role in scapular upward rotation, which is essential for arm extension. The post-treatment increase suggests that active exercises enhanced scapular synergy. Figure 27 B illustrates the latissimus dorsi contraction trend in position (3), comparing LP and RH before treatment. Muscle contraction is higher in RH, including in positions (1) and (2), though not significantly, suggesting greater overall activation. At the end of the flagging movement, the latissimus dorsi contributes to shoulder stabilisation as the arm grasps the hold. These findings indicate that, on the painful side, strength generation and stability are less efficient, as the healthy side demonstrates greater activation. Finally, in position (1) (Figure 27 C), the pectoralis major, responsible for arm flexion and abduction, shows significantly higher contraction values on RH compared to LP after physiotherapy. While treatment helped create a more balanced activation pattern, the healthy side still exhibited greater muscle activation.

Additionally, these results suggest that the dominant side may influence muscle recruitment in individuals with shoulder pain [172]. A painful condition on the non-dominant side could lead to an altered activation pattern, likely due to both pain and a reduced capacity to generate strength. Notably, all participants in this study were right-handed, which may have contributed to the observed differences. This aspect needs further investigation to better understand how dominance affects muscle activation and compensation strategies in climbers with shoulder pain.

Although no similar studies in the literature have investigated muscle activity in painful shoulders of climbers during standard climbing movements, Baláš et al. [173] provide valuable insights by comparing naturally chosen and corrected shoulder positions, with a focus on scapular stabilizing muscles in typical static climbing postures. Notably, they found that climbers tend to naturally relax the shoulder during static positions, despite the recommended active centring of the glenohumeral joint to prevent overuse syndromes and muscular imbalances.

3.11 Water-polo

Water-polo is a sport that involves repetitive swimming actions with throwing movements. Differently from swimming, water polo players use an adapted upright swimming posture to allow transport of the ball. This adapted technique reduces the body roll typically observed in swimming, thereby increasing demands on the shoulder by promoting greater abduction and internal rotation during propulsion and throwing phases [174]. During throwing, players repeatedly elevate the arm into extreme positions of flexion, abduction, and external rotation. These movements, performed without a stable base of support in water, impose substantial loads on the shoulder complex and can contribute to overuse injuries and joint pathologies [175], [176]. The throwing motion itself can be divided into distinct phases: during acceleration, the scapula rotates upward, tilts posteriorly, and moves externally; whereas in deceleration, it rotates downward, tilts anteriorly, and moves internally [177], [178], [179], [180], [181].

In water-polo player activity, the scapula is the link between the body and the kinetic chain of the upper limb; in fact, any difference in scapula alignment represents a risk factor for shoulder injuries. Correct scapula control and symmetry are essential for efficient shoulder biomechanics. Thus, this mechanism can lead to functional alterations that could precede pathological changes and represent a risk factor for soft tissue injury and glenohumeral joint instability [177].

As in swimming, repeated exposure to sport-specific demands in water polo can result in adaptive structural and functional changes in the shoulder joint. While these adaptations may enhance performance, they can simultaneously compromise joint integrity and increase the risk of overuse injuries.

Given the high physical demands placed on the shoulder, understanding its biomechanical adaptations in water polo is essential. Such knowledge is critical for optimising performance, preventing injury, and developing evidence-based training and rehabilitation strategies tailored to the unique requirements of this sport [174], [182].

3.12 Surface electromyographic analysis of the overhead shot and front crawl in competitive water polo players

3.12.1 Aim of the study

The aim of this preliminary study was to investigate upper-limb muscle activation, measured via surface electromyography (sEMG), in water polo players during the front crawl and overhead shot, comparing athletes with shoulder pain to healthy controls.

3.12.2 Material and methods

3.12.2.1 Participants

19 volunteers (15 male, aged between 1 and 26 from local water polo teams were enrolled in this preliminary study.

All participants underwent a clinical evaluation conducted by a team of physiotherapists to confirm eligibility based on the inclusion criteria and to assess their suitability for participation. The inclusion criteria required participants to be between 14 and 26 years of age, engage in five training sessions per week lasting approximately two hours each, and have a minimum of 4 years of water polo experience. Exclusion criteria included the presence of specific shoulder pathologies, a history of shoulder or cervical surgery, fractures, bilateral shoulder pain, cervical radiculopathy, or physiotherapy treatment involving the shoulder within the previous 12 months. Fourteen participants were classified as healthy (Healthy group), as they reported no history of symptoms associated with unilateral shoulder pain. Five participants were classified as having shoulder pain (Pathological group) according to the criteria proposed by Tovin [100]. Specifically, these individuals experienced pain during or immediately after swimming for a period of at least three months, with an intensity score of 4 or higher on the Visual Analogue Scale (VAS). All participants underwent a clinical evaluation performed by a team of physiotherapists to verify eligibility based on the inclusion criteria and to assess their suitability for participation.

The study protocol was reviewed and approved by the local university ethics committee (reference number 122/2022). Prior to the data acquisition session, all participants were fully informed about the study's objectives, procedures, potential risks, and benefits, and provided written informed consent in accordance with ethical standards for human research.

3.12.2.2 Study protocol

Nine muscles were selected for analysis: the medial deltoid, latissimus dorsi, pectoralis major, lower trapezius, upper trapezius, serratus anterior, triceps brachii, biceps brachii, and external oblique. To normalise muscle activity relative to the MVC, the MVC values for each muscle were obtained using the standard overcoming isometric method [168]. Subsequently, participants performed two experimental tasks: a 25-meter front crawl swim at maximum intensity (Task 1), and six overhead shots (Task 2, three ipsilateral and three contralateral). Surface EMG signals were acquired unilaterally from the arm presenting shoulder pain, or from the dominant arm in healthy participants.

Participants with shoulder pain underwent a physiotherapy program consisting of ten sessions, each lasting one hour. The treatment aimed to reduce pain through manual therapy, restore both passive and active range of motion according to the affected movement, improve functional capacity and muscle strength, and enable pain-free execution of sport-specific gestures. Active exercises were progressively introduced with increasing complexity, varying in type of contraction, training volume, and resistance load. Upon completion of the physiotherapy program, participants in the pathological group were re-evaluated following the same protocol used in the initial assessment.

3.12.2.3 Acquisition and analysis

Muscle activity was recorded using a WaveX system (Cometa System, Milan, Italy) with a sampling frequency of 2000 Hz. Wireless sensors transmitted data to a receiver connected to dedicated software (EMG and Motion Tools, version 8.11.0.0) for acquisition and analysis. Nine sensors were configured for EMG recording and positioned over the medial deltoid, latissimus dorsi, pectoralis major, lower trapezius, upper trapezius, anterior serratus, triceps brachii, biceps brachii, and external oblique muscles. An additional four sensors were configured in IMU mode and placed on the chest, arm, forearm, and hand.

Prior to electrode placement, the physiotherapist prepared the skin to optimise signal quality. Disposable Ag/AgCl electrodes (Ambu BlueSensor M, 40.8 × 34 mm) were positioned over the muscle belly and secured with adhesive film.

IMU-mode sensors were employed to identify specific movement phases during task execution. For the front crawl, three phases were identified for each stroke: the pull, push, and recovery phases. Two distinct phases were defined for the overhead shot: the charging phase, characterised by arm elevation and external rotation, and the shot phase.

The data were processed and analysed using a proprietary program developed in MATLAB®. Statistical analyses were performed using the Wilcoxon test to evaluate differences between groups and conditions.

3.12.3 Results

The results are illustrated through box plots showing the distribution of the percentage muscle contraction for the relevant muscles across groups (Healthy and Pathological) and phases of the two tasks (Task 1: Push, Pull, and Recovery; Task 2: Charge and Shot). Each box plot displays the median, 25th, and 75th percentiles, providing a clear visualisation of central tendency and variability in muscle activation patterns between groups and across movement phases. Moreover, the whiskers extend to the most data points not considered outliers (plotted individually using a red '+').

Among the 19 participants enrolled in this study, two were excluded from the analysis of Task 1. Figure 28 presents the box plot illustrating the percentage of muscle contraction for the Healthy and Pathological groups before physiotherapy treatment across the three phases identified for each stroke. The anterior serratus and lower trapezius are the two muscles showing differences approaching statistical significance: the anterior serratus across all three phases (p=0.055 for push and pull, p=0.086 for recovery), and the lower trapezius during the push and pull phases (p=0.086). Although the differences did not reach statistical significance, this near-significant effect suggests that the muscle may play an important role in the movement patterns analysed and could represent a potential target for intervention in pathological participants.

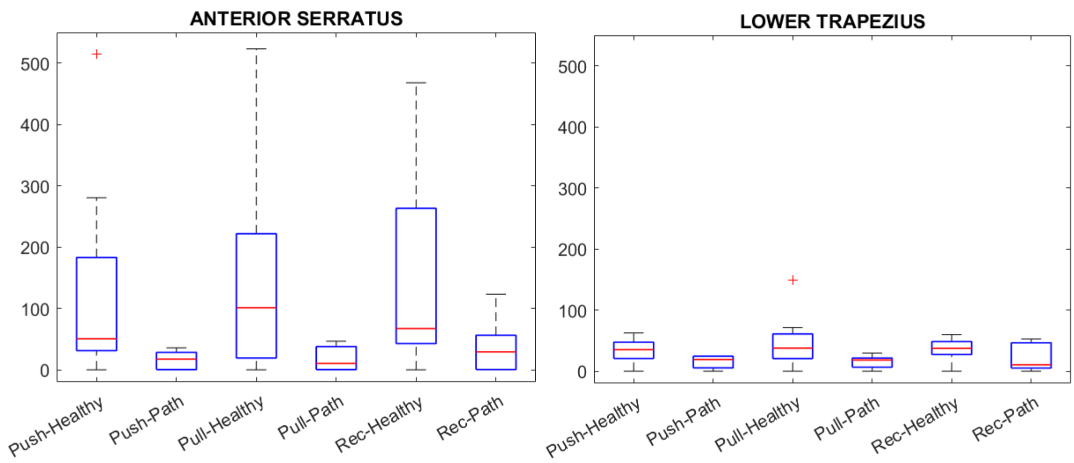


Figure 28. Box plots of the percentage of muscle contraction for each group and phase during Task 1. Healthy refers to the healthy group, and Path refers to the pathological group before physiotherapy treatment. Phases include Push, Pull, and Recovery, with differences approaching statistical significance highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

When comparing the Pathological group before (PRE) and after (POST) the physiotherapy treatment in Task 1 (Figure 29), several notable differences emerged across the three phases of the front crawl stroke. The upper trapezius, medial deltoid, pectoralis major, lower trapezius, and triceps exhibited differences approaching statistical significance (p=0.063), indicating a trend toward improved muscle activation following treatment.

Specifically, in the push phase, the pectoralis major, upper trapezius, and triceps showed increased activation; in the pull phase, changes were observed in the medial deltoid and upper trapezius; and in the recovery phase, both the lower trapezius and triceps demonstrated enhanced activity.

Overall, the mean muscle contraction of these muscles tended to increase after physiotherapy, suggesting that the intervention may have contributed to improved neuromuscular control and strength during the swimming stroke cycle.

For Task 2, all subjects were included in the analysis. In the comparison conducted before the physiotherapy treatment, the ipsilateral shots did not show any statistically significant differences, although a trend toward significance was observed for the external oblique during the Shot phase (p=0.087, Figure 30). In contrast, for the contralateral shots (Figure 31), the upper trapezius exhibited a difference approaching significance in the charge phase (p=0.056). During the shot phase, both the medial deltoid and external oblique showed significant differences (p=0.034 and p=0.010, respectively), while the latissimus dorsi displayed a trend toward significance (p=0.056). For these muscles, particularly the medial deltoid, external oblique, and latissimus dorsi, the pathological group generally demonstrated higher levels of muscle activation compared to the healthy group, suggesting altered or compensatory motor recruitment strategies during the overhead shot.

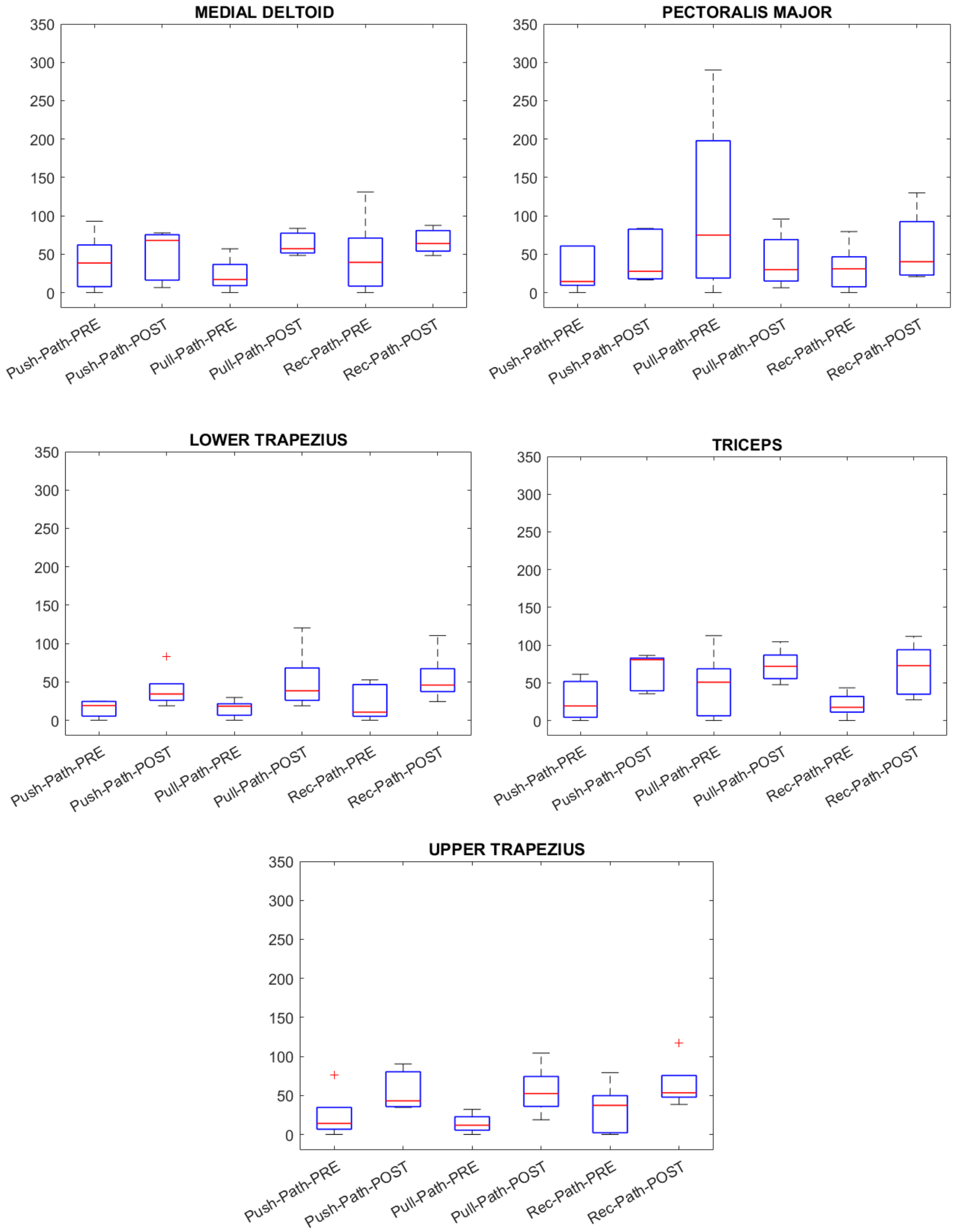


Figure 29. Box plots showing the percentage of muscle contraction for each group and phase during Task 1. Healthy refers to the healthy group, and Path denotes the pathological group; PRE indicates measurements before physiotherapy treatment, and POST indicates measurements after treatment. Phases include Push, Pull, and Recovery. Differences approaching statistical significance are highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

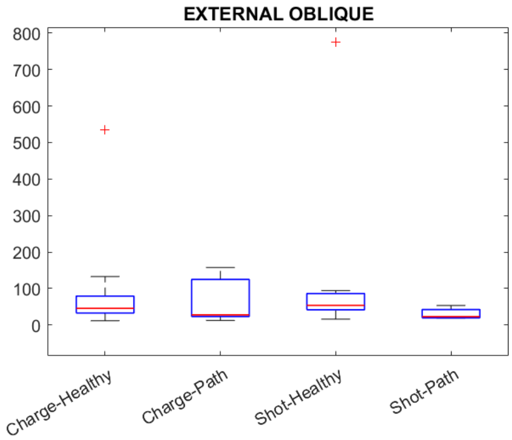


Figure 30. Box plots showing the percentage of muscle contraction for each group and phase during Task 2, ipsilateral shots. Healthy refers to the healthy group, and Path denotes the pathological group. Phases include Charge and Shot. Differences approaching statistical significance are highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

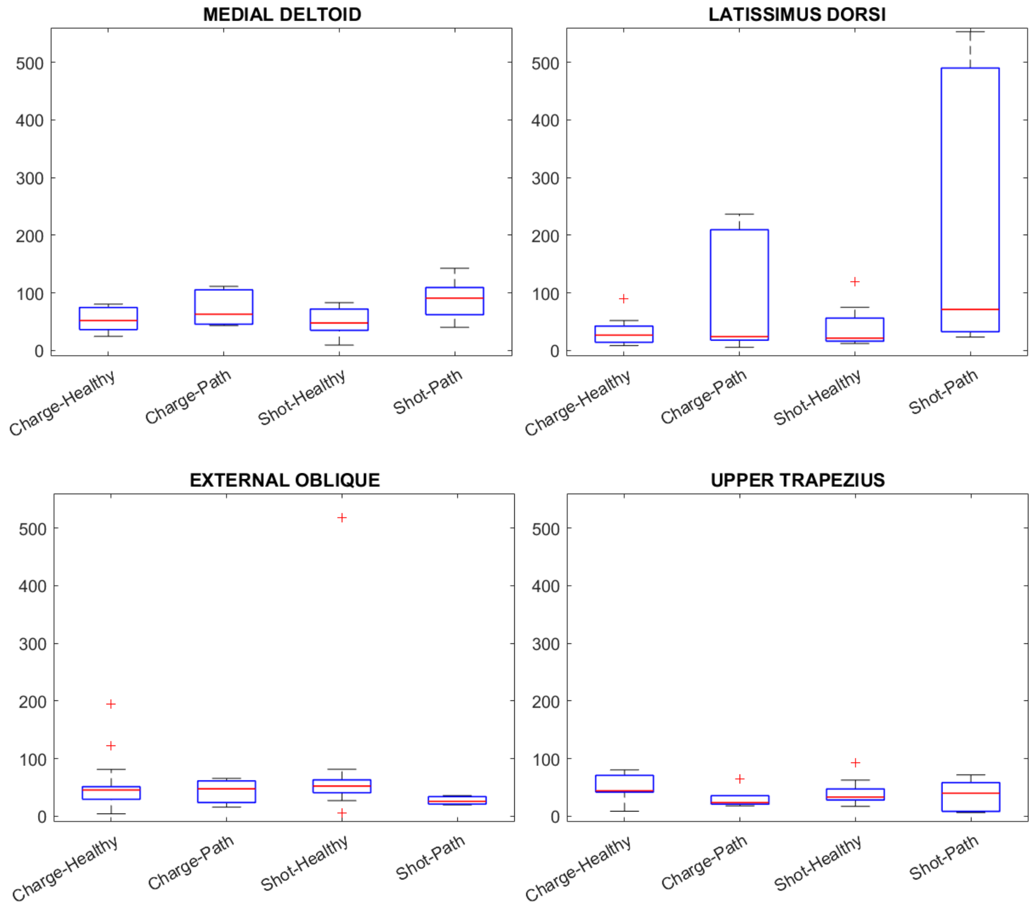


Figure 31. Box plots showing the percentage of muscle contraction for each group and phase during Task 2, contralateral shots. Healthy refers to the healthy group, and Path denotes the pathological group. Phases include Charge and Shot. Differences approaching statistical significance are highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

When comparing the pathological group before and after the physiotherapy treatment, both the pectoralis major and biceps brachii exhibited differences approaching statistical significance (p=0.062) during the shot phase. Specifically, in the ipsilateral shot (Figure 32), both the pectoralis major and biceps showed this trend, whereas in the contralateral shot (Figure 33), only the pectoralis major demonstrated a near-significant increase in activation after treatment. These findings suggest that the physiotherapy intervention may have contributed to improved muscle engagement and coordination, particularly in the primary muscles responsible for force generation during the overhead shot.

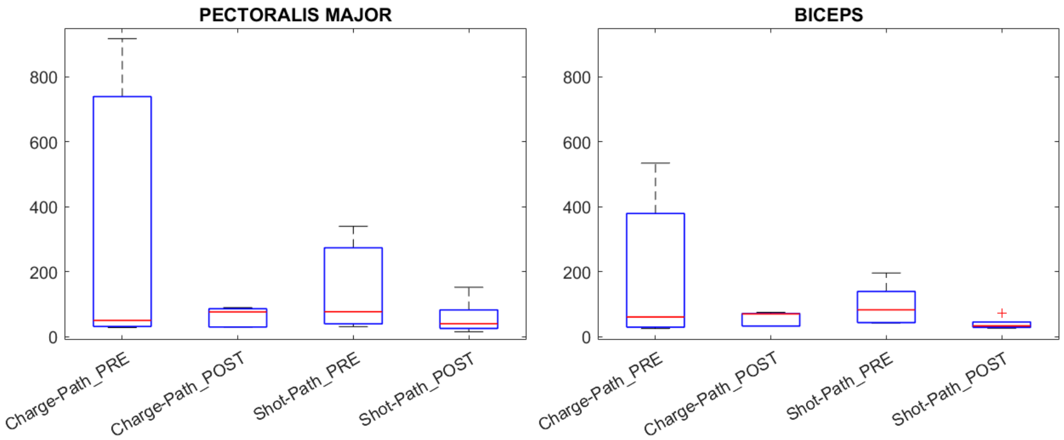


Figure 32. Box plots showing the percentage of muscle contraction for each group and phase during Task 2, ipsilateral shots. Healthy refers to the healthy group, and Path denotes the pathological group; PRE indicates measurements before physiotherapy treatment, and POST indicates measurements after treatment. Phases include Charge and Shot. Differences approaching statistical significance are highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

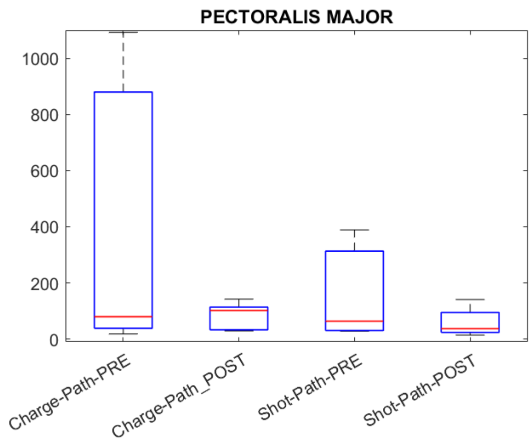


Figure 33. Box plots showing the percentage of muscle contraction for each group and phase during Task 2, contralateral shots. Healthy refers to the healthy group, and Path denotes the pathological group. Phases include Charge and Shot. Differences approaching statistical significance are highlighted. The median values are shown in red, and the boxes represent the 25th to 75th percentiles.

Due to the relatively small sample size and the considerable inter-individual variability observed among participants, as illustrated in the figures above, only a limited number of statistically significant differences were detected. This variability likely reflects individual differences in technique, muscle recruitment patterns, and response to fatigue or pain, which may have influenced the overall statistical power of the analysis.

3.12.4 Discussion

This preliminary study aimed to characterise upper-limb muscle activation in water polo players using surface electromyography (sEMG) during two sport-specific tasks: a 25-meter front crawl and six overhead shots. Participants were divided into two groups: a Healthy group and a Pathological group presenting with shoulder pain.

Like swimming, water polo is classified as an overhead sport, characterised by repetitive arm movements performed above the head that require substantial shoulder elevation and external rotation. These repetitive and high-intensity actions impose significant mechanical stress on the shoulder complex, making shoulder injuries the most common type of injury among water polo players [97]. Understanding the muscular patterns associated with these movements is therefore essential for identifying potential risk factors for injury and guiding targeted rehabilitation strategies.

The muscles analysed in this study included the medial deltoid, latissimus dorsi, pectoralis major, lower trapezius, upper trapezius, anterior serratus, triceps brachii, biceps brachii, and external oblique. To calculate the percentage of muscle activation, the maximum voluntary contraction (MVC) was recorded for each muscle using standardised isometric testing procedures. However, examination of the results suggests that the MVC may not have been consistently or accurately acquired for all muscles across participants. In some cases, suboptimal MVC execution or inconsistent stabilisation during testing may have led to underestimation of true maximal values, consequently inflating normalised activation percentages. This limitation should be taken into account when interpreting the results, as it may have introduced variability and affected the comparability of muscle activation levels between subjects and conditions.

According to the results of Task 1, the 25-meter front crawl, comparison between the Healthy and Pathological groups revealed that only two muscles showed differences approaching statistical significance: the anterior serratus (in pull, push, and recovery phases) and the lower trapezius (in pull and push phases), both exhibiting higher activation levels in the Healthy group.

During swimming, particularly in the front crawl, the anterior serratus plays a key role in scapular upward rotation and protraction, actions essential for maintaining an efficient arm trajectory and stable shoulder mechanics [183]. Fatigue or dysfunction of this muscle can compromise scapular control, reducing stroke efficiency and potentially increasing the risk of shoulder impingement. The lower trapezius, together with the serratus anterior, contributes to the coordinated scapulothoracic movement required for overhead propulsion. The reduced activation observed in the Pathological group may therefore reflect compensatory strategies or neuromuscular inhibition related to shoulder pain. Moreover, previous studies have reported the critical role of these muscles in maintaining scapular stability and managing shoulder function, in particular during overhead movements [184], [185]. Intervention targeting these muscles could be beneficial in pathological populations where neuromuscular control may be compromised [186].

Following physiotherapy, significant improvements in muscle activation patterns were observed in the pathological group, especially during the specific push and recovery phase, showing higher activation levels in the POST-treatment condition. This suggests that the physiotherapy intervention contributed to improving muscular strength and neuromuscular efficiency in the upper limb. Specifically, increased activation was observed in the anterior deltoid and upper trapezius during the Pull phase, the pectoralis major, triceps brachii, and upper trapezius during the Push phase, and the lower trapezius and triceps during the Recovery phase. This overall increase in muscle activity may reflect enhanced motor unit recruitment and better scapulohumeral coordination following rehabilitation. Physiotherapy likely improved not only the strength of key stabilising and propulsive muscles but also their timing and activation patterns, leading to a more efficient and balanced contribution during each stroke phase. The greater activation of the scapular stabilisers (upper and lower trapezius) suggests a restoration of proper scapular kinematics, which is often impaired in individuals with shoulder pathology. These results align with findings suggesting that targeted physiotherapy can positively affect scapular and shoulder mechanics in compromised populations [187].

In Task 2, the analysis of shooting indicates that, while ipsilateral shots display a near-significant difference in the external oblique, in contralateral shots pathological group presents a higher activation for the medial deltoid and latissimus dorsi in the Shot phase. This may indicate a compensatory movement or altered recruitment strategies prevalent in the pathological group, trying to generate force during the overhead shot.

Furthermore, comparing muscle activation before and after physiotherapy treatment revealed improved neuromuscular engagement in the primary muscles responsible for force generation during tasks like shooting. The activation of pectoralis major and biceps approached significance during the ipsilateral and contralateral shot, reinforcing the idea that therapeutic interventions can enable participants to achieve more effective movement patterns [177], [188], [189].

The near-significant differences and overall trends observed following physiotherapy intervention highlight important directions for future research and clinical attention, particularly regarding the anterior serratus and lower trapezius, which play a crucial role in maintaining optimal shoulder stability and scapular control in overhead athletes. High inter-individual variability in muscle activation is a well-documented phenomenon in both swimming and water polo, even among elite performers. Such variability can mask statistical significance and is likely influenced by individual technique, fatigue levels, training background, and compensatory movement patterns developed to mitigate shoulder discomfort or weakness [190].

Chapter 4. CONCLUSION

This thesis presents the studies and findings arising from the application of two state-of-the-art technologies, IMU and sEMG, along with their signal processing methodologies, to develop innovative and effective protocols for the analysis of human movement. The overarching goal is to facilitate the transition from traditional observational assessments to objective, data-driven approaches that enhance the accuracy of diagnosis, improve treatment planning, and optimise rehabilitation outcomes in both musculoskeletal and neurological disorders.

By integrating kinematic and electromyographic data, this work aims to provide a comprehensive understanding of motor function, enabling clinicians and researchers to better quantify movement quality, compensatory strategies, and neuromuscular coordination. Such integration not only supports personalised therapeutic interventions but also contributes to the development of digital health tools and evidence-based rehabilitation frameworks for clinical practice and sports medicine.

The first part of this thesis (Chapter 2) presents the development and application of a novel protocol designed to analyse and assess the tasks included in the Tinetti Fall Risk Assessment Scale (balance section) using IMU. One of the major limitations of traditional clinical scales lies in their subjectivity, as outcomes often depend on inter-rater variability and the level of clinician expertise required for proper administration and scoring. By integrating objective kinematic data from wearable sensors, this approach aims to reduce subjectivity and standardise fall-risk evaluation, thereby improving the reliability and reproducibility of clinical assessments.

In this work, we propose a movement acquisition protocol based on IMU, coupled with kinematic parameter extraction and quantitative analysis, to identify and describe different levels of fall risk within the elderly population. Starting with the application of the proposed protocol on a young, healthy group, we established a benchmark dataset to serve as a reference for comparison with the elderly population. From this initial study, several parameters were identified as effective descriptors of movement quality, such as angular peak and movement duration during chair rise and sitting tasks, and FD and spectral power during standing balance exercises. When these results were compared with data collected from an elderly cohort characterised by different fall-risk levels, significant differences emerged, particularly in the spectral power values during standing balance tasks, highlighting measurable distinctions in postural stability between age groups. Finally, in the third study, which focused exclusively on the elderly population, we investigated potential correlations between kinematic parameters and Tinetti scores. Although strong correlations were not established, notable trends were observed in step parameters, specifically median step duration and mean step amplitude, during the 360° turning task, suggesting a possible link between gait regularity and balance performance.

The findings indicate that not all tasks within the Tinetti protocol exhibit the same sensitivity in detecting differences in motor control. Among them, the 360° turning task proved to be the most responsive, effectively capturing variations in gait regularity and postural stability. The high inter-individual variability observed, particularly among elderly participants, likely influenced the expected correlations and highlights the heterogeneity of motor performance within this population. Despite these limitations, the present findings reinforce the value of integrating wearable sensor technologies into clinical fall-risk assessment protocols, thereby contributing to improved mobility evaluation, fall prevention, and the promotion of functional independence in the elderly.

The second part of this thesis (Chapter 3) focuses on three studies investigating the kinematic patterns and muscular activity of the shoulder joint across different overhead sports, swimming, climbing, and water polo. The shoulder, being the most mobile joint in the human body, sacrifices stability for mobility, making it particularly vulnerable to overuse injuries and dislocations.

In this context, the integrated use of wearable technologies such as IMU and sEMG offers a valuable opportunity to capture high-resolution data on joint angles, accelerations, and muscle activation patterns during sport-specific movements. For each sport, a dedicated experimental protocol was developed, comprising both a data acquisition phase and a physiotherapy intervention.

Swimming and water polo share similar movement patterns that demand high flexibility and repetitive overhead actions. These characteristics often promote joint hypermobility, potentially compromising the stability of the shoulder capsule and predisposing athletes to injury. Across both disciplines, clear differences were observed between healthy and pathological athletes, particularly in muscles primarily involved in the front crawl and overhead shot movements. Following physiotherapy treatment, improvements were noted in muscle activation balance and proprioceptive control, suggesting that rehabilitation plays a central role in restoring optimal shoulder function.

In contrast, climbing imposes unique biomechanical demands, requiring not only extensive ranges of motion to reach distant holds but also the ability to generate and transfer force effectively to support body weight, especially on overhanging routes. Similar to the aquatic disciplines, comparisons between pre- and post-physiotherapy findings revealed meaningful improvements in the activation of key stabilising and propulsive muscles critical for climbing performance.

These studies collectively demonstrate the potential of wearable sensor technologies, particularly IMU and sEMG, as reliable tools for quantitative analysis of shoulder biomechanics. Ultimately, integrating wearable systems into sports and rehabilitation settings could enhance injury prevention, support individualised physiotherapy strategies, and advance our understanding of shoulder mechanics in overhead athletes.

Despite their potential for enabling motion analysis in real-world settings, wearable technologies are subject to several limitations that should be considered. Measurement accuracy can be affected by sensor placement variability, soft tissue artefacts, and misalignment between sensor and anatomical reference frames, particularly during dynamic tasks [191], [192].
In inertial-based systems, signal drift and magnetic disturbances may further compromise the reliability of kinematic estimates over time [193]. Similarly, sEMG signals are sensitive to electrode positioning, skin-electrode impedance, cross-talk from adjacent muscles, and magnetic disturbances, which can influence signal quality and repeatability [194], [195]. When measurements are performed in aquatic environments, additional challenges arise, as water exposure may alter electrode adhesion and skin-electrode contact, and increase susceptibility to motion artefacts or signal attenuation, thereby affecting overall sensor performance [196], [197].

Beyond instrumentation-related limitations, the relatively small sample size represents a further constraint of this thesis, particularly in the studies aimed at characterising shoulder kinematics and muscle activity during selected overhead sports. The limited number of participants reduces statistical power and may restrict the representativeness and generalizability of the findings to the broader population, especially in the presence of high inter-subject variability [198].

Building on the findings of the present work, several directions can be identified for future research. The acquisition protocol for fall-risk assessment in the elderly population using IMU could be further improved by incorporating additional tasks within the gait section of the Tinetti test and narrowing the range of Tinetti scores under consideration, targeting participants within specific fall-risk intervals to enhance the sensitivity and interpretability of the results, and focusing on kinematic parameters, which have already shown promising results. Moreover, the development of predictive models based on kinematic parameters may provide substantial added value by classifying individuals into discrete fall risk categories or estimating a continuous probability of fall risk, capturing subtle motor impairments that may not be detectable through conventional clinical scores alone. Furthermore, predictive modelling may support early identification of high-risk individuals, facilitate longitudinal monitoring, and enhance clinical decision-making in both clinical and community settings.

The second part of this thesis focuses on the investigation of shoulder kinematic patterns and muscular activity across different overhead sports. The relatively small sample size represents a major limitation, restricting the statistical significance and interpretability of the results. Future research should therefore aim to expand participant cohorts, include healthy control groups, and further refine measurement protocols in order to confirm and extend these preliminary findings.

To mitigate the limitations associated with wearable instrumentation and to enhance measurement robustness, future experimental setups could integrate wearable sensors with optical motion capture systems [199], [200]. In addition, the experimental protocols included the execution of standardised movements, such as shoulder flexion, abduction, and internal rotation, which could be exploited in future studies to enable direct comparisons across different overhead sports as well as with non-overhead athletic populations.

Finally, during the data processing phase, the combined use of IMU and sEMG could enable a truly multimodal assessment that integrates kinematic information related to movement execution with neuromuscular activation patterns underlying motor control strategies. The integration of these complementary modalities may enhance sensitivity to subtle impairments and improve the robustness and interpretability of the measurements, supporting a more comprehensive characterisation of human movement [201], [202].

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