UNIVERSITY OF URBINO CARLO BO

Department of Biomolecular Sciences

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Ph.D. course in **Biomolecular and Health Sciences**

XXXVII CYCLE

**Feasibility and effectiveness of innovative, technology-delivered, home-based resistance exercise testing and training interventions in older adults**

**S.S.D. MEDF-01/A**

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## ABSTRACT

The World Health Organization (WHO) projects a significant increase in the global elderly population by 2050, with a corresponding rise in life expectancy. This demographic shift poses challenges for public health and finances, primarily due to the age-related physical and cognitive decline. Even in the normal aging trajectory, the decline in muscle strength and power affects older individuals' abilities to perform daily activities, increasing the risk of functional limitations, loss of independence, and adverse health outcomes such as falls and mortality.

Resistance training (RT) has been identified as an effective intervention to counteract the age-related decline in muscle function. RT is typically conducted in specialized facilities (i.e., gym) and with specialized equipment (i.e., isotonic machines, etc.). While recognized as the gold standard, this approach has a high cost and could be difficult to afford for the elderly who may live in retirement. In this special population, other barriers to engaging in an active lifestyle exist, for instance, lack of public transport availability, lack of information about the importance of healthy behavior, and social isolation. Therefore, all these barriers contribute to reinforcing older individuals' sedentary behaviors, which in turn exacerbate the negative effect of aging on muscle function.

Home-based exercise programs have emerged as a viable alternative, demonstrating positive effects on physical function. However, these programs often lack proper monitoring of exercise execution. Technological advancements, including Wearable inertial measurement units (IMUs), offer potential solutions for effective home-based RT by enabling exercise prescription and monitoring.

Despite their promising features, these devices' safety, feasibility, and efficacy for sarcopenia prevention require further investigation.

Moreover, affordable and practical tools for assessing and monitoring physical function in older adults are essential. IMUs present an opportunity for objective and digitalized movement assessment, potentially leading to the early detection of functional decline. This approach could bridge the gap between exercise prescription and the need for accurate monitoring in home-based settings, ultimately supporting the health and independence of the aging population.

This thesis aims to investigate the opportunity to deliver directly in older individuals' home environments both physical exercise training programs and physical assessment for health

screening and monitoring of aging trajectories. In the introduction, an overview of the main issues related to older individuals' sedentary behaviors, as well as state-of-the-art effective active lifestyle initiatives, is provided. Then, the results of four original research articles are presented in chapters one, two, three, and four. Finally, in the general discussion section, the main findings of this research are presented.

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URL: https://www.sciencedirect.com/science/article/pii/S0167494324002358

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[Chapter 2]

Bochicchio, G.; Ferrari, L.; Bottari, A.; Lucertini, F.; Scarton, A.; Pogliaghi, S.; **“Temporal, Kinematic and Kinetic Variables Derived from a Wearable 3D Inertial Sensor to Estimate Muscle Power during the 5 Sit to Stand Test in Older Individuals: A Validation Study”**, (2023) Sensors, 23(10), art. no. 4802.
URL: https://www.mdpi.com/1424-8220/23/10/4802

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Bochicchio, G.; Ferrari, L.; Bottari, A.; Lucertini, F.; Cavedon, V.; Milanese, C.; Pogliaghi, S.; **“Loaded 5 Sit-to-Stand Test to Determine the Force–Velocity Relationship in Older Adults: A Validation Study”**, (2023) Applied Sciences (Switzerland), 13(13), art. no. 7837.
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URL: https://www.mdpi.com/2079-6374/14/11/529

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# INTRODUCTION

The World Health Organization (World Health Organization, 2019) stated that, in 2050, the number of older individuals aged 65 years and older will double compared to 2019, and life expectancy is expected to increase as well. A person who reaches age 65 years in 2045-2050 can expect to live, on average, an additional 19 years (World Health Organization, 2019). This represents a concern for public health as well as for public finances (J. Chen, Zhao, Zhou, Ou, & Yao, 2023; Lopreite & Mauro, 2017). Indeed, even a normal aging trajectory is characterized by a decline in almost all aspects of the older individual life, producing a decay in fitness, cognitive, metabolic, and muscle function (Chodzko-Zajko et al., 2009a).

Muscle function is defined as the ability of a muscle to express maximal strength, power, and physical performance (Cruz-Jentoft et al., 2019). Muscle strength is defined as the ability to exert a force on an external resistance and after the 65$^{th}$ year, a loss in muscle strength of 1.5% per year is expected (Anton, Spirduso, & Tanaka, 2004; Muollo et al., 2021) (Figure 1). Muscle power is defined as the ability to exert muscle strength in the shortest time possible; it declines at a faster rate (3-4% per year) compared to muscle strength (Figure 1) (Anton et al., 2004; Cruz-Jentoft et al., 2019; Fragala et al., 2019; Keller & Engelhardt, 2013) and seems to be the strongest predictor of functional limitations in aging (Alcazar et al., 2018; Regterschot, Morat, Folkersma, & Zijlstra, 2015), making it an ideal sentinel index for detecting individuals at higher risks and monitoring aging trajectory (McLeod, Breen, Hamilton, & Philp, 2016).

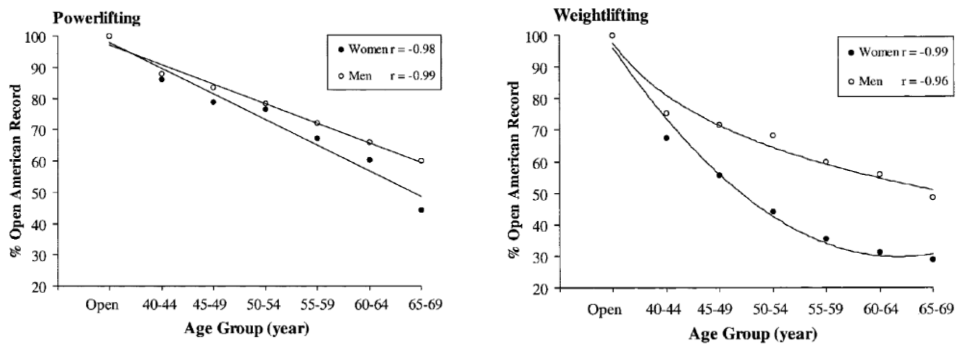


*Figure 1. Age-related decline in strength (powerlifting) and Power (Weightlifting) expressed as a % of Open category record, in American athletes of both sexes (Anton et al., 2004)*

Physical performance is defined as the ability to perform daily activities (such as walking, climbing a stair, standing and sitting from a chair, maintaining balance, etc.) and, therefore, is a crucial index of living independence and mobility (Cruz-Jentoft et al., 2019).

Physical activity has been shown to affect the aging trajectory positively (Figure 3). In particular, resistance training is a sub-type of exercise that has been shown to counteract the decline in muscle strength and power. Indeed, international guidelines for physical activity in healthy older adults (Chodzko-Zajko et al., 2009b; Fragala et al., 2019) recommend performing resistance training with a minimum frequency of 2 sessions per week on non-consecutive days. Resistance training is typically conducted in specialized facilities (e.g., gyms) and with specialized equipment (e.g., free weights, isotonic machines, etc.). While recognized as the gold standard approach, resistance training has a high cost and could be difficult to afford for the elderly who may live in retirement (Burton et al., 2017).

Older individuals experience also other barriers than the high cost to engaging in an active lifestyle. Lack of public transport availability, lack of information about the importance of healthy behavior, and social isolation (Burton et al., 2017; World Health Organization, 2019) are all factors enhancing older individuals' sedentary behavior, which, in turn, exacerbates the age-related decline in physical and muscle function producing a vicious cycle leading to an increased risk of adverse health outcomes, including falls, hospitalization, institutionalization, and mortality (Cruz-Jentoft et al., 2019; Muollo et al., 2021) (Figure 2).

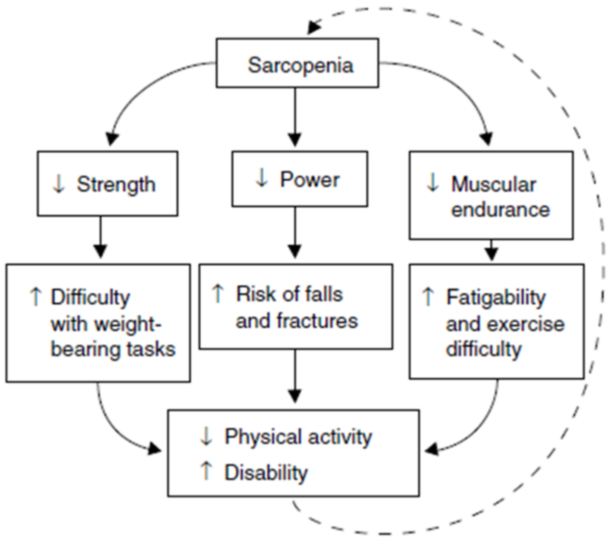


*Figure 2. A model of the functional consequences of age-related sarcopenia and the positive feedback loop by which the end result of reduced physical activity further exacerbates the progression of the disorder. ↓ indicates decrease; ↑ indicates increase* (Hunter, McCarthy, & Marcas M. Bamman, 2004).

This threat to independence, particularly late in life, represents a concern for healthcare operators and the scientific community, leading governments to search for and support effective, feasible, and sustainable interventions to counteract the age-related decline in muscle function, and encourage saving behavior and healthy lifestyles throughout life (World Health Organization, 2019).

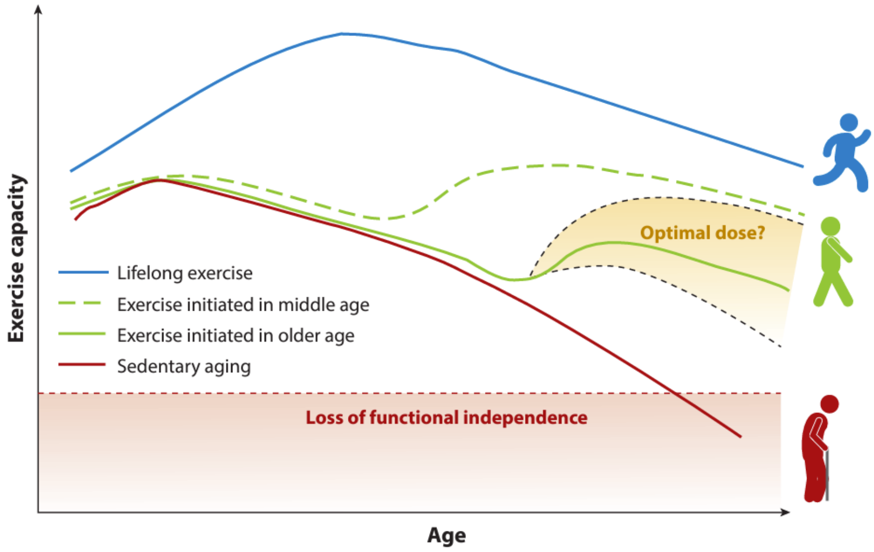


*Figure 3. Protective effect of physical exercise against the age-related loss of functional independence* (Carrick-Ranson, Howden, & Levine, 2021).

In this context, healthcare initiatives that promote an active lifestyle among older individuals focused their attention on home-based exercise interventions. These approaches have been shown to positively affect physical function (muscle function, walking, and sway ability) (Figure 4). Recent systematic reviews (Chaabene et al., 2021; Mañas et al., 2021; Song, Kim, & Kim, 2023) reported an overall modest to small effect of home-based exercise interventions on muscle strength and power (effect size 0.30-0.34 and 0.43-0.44, respectively), and balance and walking speed (effect size 0.28-0.32 and 0.28-0.34, respectively).

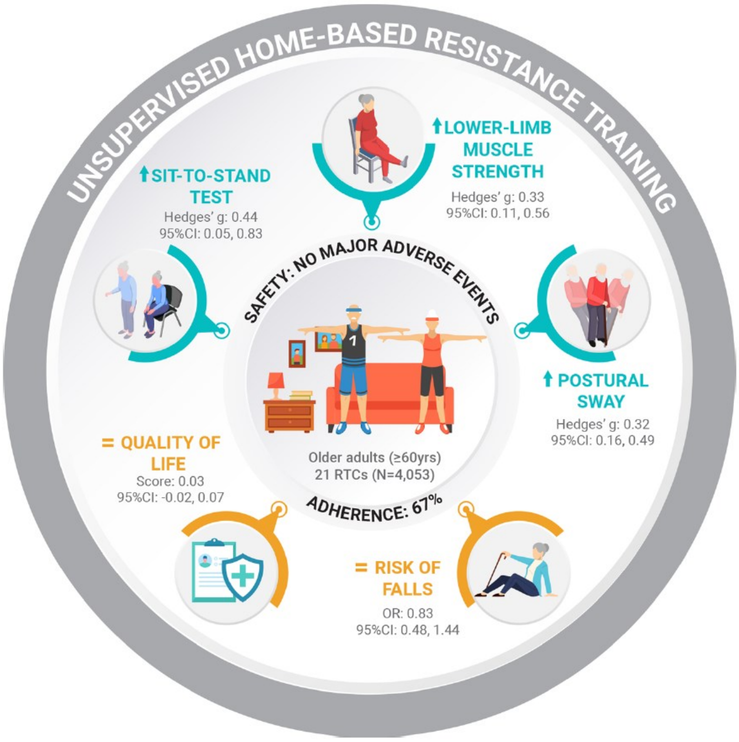


*Figure 4. Safety adherence and effectiveness of unsupervised home-based resistance training for community-dwelling older adults. A systematic review and meta-analysis of randomized controlled trials (Mañas et al., 2021)*

Different methods were used for the administration of exercise programs. Most of them required an operator (both in-person and live-streaming sessions), or the exercise program was conducted in an offline setting (i.e., with the use of video tutorials, pre-recorded training sessions, etc.) without any monitoring feedback about proper exercise execution and effort. Various technological solutions developed by the industry aim to fill the gap between exercise prescription and monitoring the execution of home-based training programs. Often, these devices combine wearable sensors and dedicated software that easily delivers the prescription of exercises from the trainer and monitors the correct and actual execution of the training session (Raquel Costa-Brito et al., 2024). While the above features appear extremely promising, the overall safety, feasibility, and efficacy of these devices in the specific context of sarcopenia prevention in older adults remain to be determined (Solis-Navarro et al., 2022).

From the already mentioned threat of the bad aging trajectory on older individuals' health status, it becomes crucial to provide clinicians with accessible tools for assessing and monitoring muscle and physical function. In health and exercise sciences, the assessment of lower limb muscle function (i.e., lower limb muscle strength and power) in older adults can be obtained with specific equipment (isokinetic dynamometer, isotonic machines,

Nottingham power rig) that requires standardized and relatively unnatural muscle actions (Glenn, Gray, & Binns, 2017). The laboratory gold standard instruments for measuring these variables during natural movements are motion capture systems and force plates. However, this equipment is expensive and requires specialized personnel and time-consuming procedures to collect and analyze data. On the contrary, the ideal approach for testing and monitoring, in clinical practice, requires affordable costs, relatively simple and time-efficient procedures, and movements that mimic daily activities' muscle function (Glenn et al., 2017).
Sway ability is a crucial component of physical function. It is measurable through static sway, defined as the ability to maintain the center of pressure (CoP) within the limits of the base of support while standing still (Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996). Typically, clinical tests are based on the visual assessment of the patient by using a scalar score (Berg, Wood-Dauphinee, Williams, & Maki, 1992; Guralnik et al., 1994). Despite being flexible and easy to use, clinical tests are sensible only to visible and gross balance deficits, excluding them as a tool for the early identification and monitoring of an increased risk of falls and/or the detection of subtle balance deficits (Michalska et al., 2021). To overcome these limitations of clinical tests, balance can be quantified objectively through posturography by using optoelectronic systems or force plates (Baker, Gough, & Gordon, 2021; Michalska et al., 2021). However, the already mentioned issues linked with the required equipment (high cost, needed for specialized personnel, not very transportable devices) limit the use of the gold standard equipment solely to laboratory settings.

Body-worn sensors (Figure 5) could provide the opportunity to perform objective and digitalized movement measures and possibly accurate muscle function estimates. Among body-worn sensors, wearable inertial measurement units (IMU) can record kinematic and kinetic information during various human movements (Pollind & Soangra, 2020; Rodríguez-Martín, Samà, Pérez-López, & Català, 2012; Witchel et al., 2018). An IMU typically includes a triaxial accelerometer that measures the proper linear acceleration, a triaxial gyroscope that measures angular velocity, and a magnetometer that measures both the amplitude and direction of the local magnetic field. All the components measure their respective physical quantities to a common three-axe frame (Ghislieri, Gastaldi, Pastorelli, Tadano, & Agostini, 2019).

*Figure 5. A wearable inertial measurement unit (IMU) (Image from https://euleria.health/home/).*

The increasing number of older individuals and the related health issues that came along even with normal aging, require special attention from clinicians and health operators. New and different approaches to promoting an active lifestyle have been investigated, and among them, home-based interventions seem to better fit older individuals' necessities and characteristics. New technology seems to fill the gap between the exercise prescription and the need to monitor the correct and actual execution of the training session. While the above features appear extremely promising, the overall safety, feasibility, and efficacy of these devices in the specific context of muscle function decline prevention in older adults remain to be fully determined. In addition, delivering physical functional assessment in a home environment by using wearable IMU sensors could be crucial in the monitoring and early detection of functional decline not associated with a normal aging trajectory.

# PURPOSES AND RESEARCH QUESTIONS

This thesis has two main purposes.

The first part of this thesis investigated the feasibility and effectiveness of a 6-month, home-based resistance exercise delivered by a remote technological solution in healthy older adults (chapter 1).
The second part of this thesis explored the portability and feasibility of common physical tests instrumented with IMU technology. The first study (chapter 2) investigated the validity of IMU sensors for assessing the lower limb musculature's maximal strength and power during a sit-to-stand task. The second study (chapter 3) investigated the validity of the IMU sensor to correctly characterize the force-velocity profile of the lower limbs during a loaded 5 sit-to-stand test. Finally, the fourth study (chapter 4) evaluated the validity of the IMU sensor to assess static sway ability during closed and opened eyes trials.

**CHAPTER**

# 1

**Feasibility and effectiveness**

**of a 6-month, home-based, resistance exercise**

**delivered by a remote technological solution**

**in healthy older adults.**

**Feasibility and effectiveness of a 6-month, home-based, resistance exercise delivered by a remote technological solution in healthy older adults.**

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## ABSTRACT

**Background:** Aging is characterized by a physiological decline in physical function, muscle mass, strength, and power. Home-based resistance training interventions have gained increasing attention from scientists and healthcare system operators, but their efficacy is yet to be fully determined. **Aims**: to verify the safety, feasibility, and efficacy of a home-based resistance training program delivered by innovative technological solution in healthy older adults. **Methods**: 73 participants (36 females) were randomly allocated to either a control (C) or an intervention (I) group consisting of a 6-months home-based resistance training program delivered through an innovative technological solution, which included a wearable inertial sensor and a dedicated tablet. The safety and feasibility of the intervention were assessed by recording training-related adverse events and training adherence. Body composition, standing static balance, 10-meter walking, and loaded 5 sit-to-stand tests were monitored to quantify efficacy. **Results**: No adverse events were recorded. Adherence to the training program was relatively high (61% of participants performed the target 3 sessions) in the first trimester, significantly dropping during the second one. The intervention positively affected walking parameters ($p < 0.05$) and maximal force ($p = 0.009$) while no effect was recorded on body composition, balance, and muscle power. **Conclusions**: The home-based device-supported intervention was safe and feasible, positively affecting walking parameters and lower limbs' maximal force. This approach should be incentivized when barriers to participation in traditional resistance exercise programs are present.

## KEYWORDS

aging, physical function, sarcopenia prevention, strength training, teleexercise

## INTRODUCTION

Aging is characterized by a natural decline in overall fitness, including loss of muscle mass, strength, and power. This decline contributes to an increased risk of adverse health events, resulting in a progressively limited ability to carry out daily activities (Cruz-Jentoft & Sayer, 2019). For this reason, governments have been increasingly implementing interventions to fight physical inactivity among the elderly as part of health policies intended to maintain a high quality of life (de Oliveira, Muzolon, Antunes, & Do Nascimento Júnior, 2019; Ferreira, Scariot, & da Rosa, 2023).

Resistance training is commonly considered an effective exercise intervention to promote increased skeletal muscle mass, muscle strength, and improvements in physical performance parameters (e.g., walking speed) while reducing relative adipose tissue (Geng, Zhai, Wang, Wei, & Hou, 2023; Maruya et al., 2016; Nascimento et al., 2019).

Traditional resistance training is typically performed in specialized facilities (e.g., commercial gym), under supervision, and utilizing special equipment (e.g., dynamic-constant external resistance machine). However, this setting may represent a disincentive to the older population due to accessibility and cost limitations, especially in low- and middle-income countries. Furthermore, older individuals experience intrinsic challenges such as reduced mobility, a high perception of difficulty, fear of injury, time constraints and lack of interest and/or knowledge. They also encounter environmental barriers including a lack of transportation or limited public transport options and distance from exercise facilities. These factors pose obstacles to maintaining an active lifestyle as well as to participating in exercise programs (Burton et al., 2017; Fyfe, Dalla Via, Jansons, Scott, & Daly, 2022; Thiebaud, Funk, & Abe, 2014).

Therefore, attention towards effective and accessible strategies (e.g., bodyweight exercise or small equipment) for resistance training administration for preserving strength and functional ability across the lifespan has been rising (Fyfe et al., 2022). In this context, home-based resistance training interventions have gained increasing attention from both scientists and operators of the health care system, but their actual efficacy is yet to be fully determined (Langeard et al., 2022). Recent systematic reviews (Chaabene et al., 2021; Mañas et al., 2021; Song et al., 2023) reported an overall modest to small effect of home-based exercise interventions on muscle strength and power (effect size 0.30-0.34; effect size 0.43-0.44 respectively), balance and walking speed (effect size 0.28-0.32; effect size, 0.34 respectively).

In a remote/offline setting, the lack of monitoring of exercise execution (i.e., safeness concern) and adherence to the training program with a proper progression of the overload could represent factors influencing efficacy (Chaabene et al., 2021; Lacroix, Hortobágyi, Beurskens, & Granacher, 2017). While most studies reported fair to good compliance (mean adherence: 67-70 %), this variable is typically based on participant-filled training diaries (Chaabene et al., 2021; Mañas et al., 2021) known to potentially overestimate adherence (Chaabene et al., 2021). Therefore, objective quantitative and qualitative indexes of compliance to the prescribed sessions and, within the session, to the single exercises, sets and repetitions, are needed to evaluate actual program efficacy.

Various technological solutions developed by the industry aim to fill the gap between exercise prescription and monitoring the execution of home-based training programs. Often, these devices combine wearable sensors and dedicated software that easily delivers the prescription of exercises from the trainer and monitors the correct and actual execution of the training session. While the above features appear extremely promising, the overall safety, feasibility, and efficacy of these devices in the specific context of sarcopenia prevention in older adults remain to be determined.

The aim of this study is to test the safety, feasibility, and effectiveness, on body composition, balance, gait and strength, of a 6-month home-based resistance training program administered through an innovative technological solution, in healthy older adults.

## METHODS

### *Procedure*

This 6-month randomized control trial was conducted at the University of Verona (University of Verona, Verona, Italy) in accordance with the CONSORT statement (Schulz, Altman, & Moher, 2011). All procedures used in the study were approved by the Ethics Committee for Human Research from the University of Verona (28/2023) and were conducted in conformity with the Declaration of Helsinki. All participants signed a written informed consent prior to participation. During the 6 months, assessments were conducted at the beginning (T0), midpoint (T3), and conclusion of the experimental protocol (T6). At each assessment window, the participants attended the laboratory twice, at the same time of the day, separated by at least a 72-hour recovery. Participants were instructed to refrain from strenuous activities during the 24 hours preceding each visit. During the first visit, anthropometric measures and body composition markers were collected, while the second visit included balance, gait and strength test for each participant.

### *Participants*

Participants were recruited by local advertisement. The inclusion criterion was age above 60 years, while a preliminary telephone interview and a successive medical screening allowed excluding of individuals with any orthopedic, mental, or neurological diseases that could interfere with the ability to perform a resistance training protocol. Participants who met the inclusion criteria were divided into either the intervention group or the control group, randomized and counterbalanced for sex and age. The intervention group underwent a 6-month home-based resistance training program with an innovative technological solution (see further for the description), while the control group was instructed to maintain their regular lifestyle.

### *Exercise intervention*

The home-based intervention group was asked to perform a minimum of 3 and a maximum of 5 training sessions per week, each lasting from 30 to 70 minutes each, for a total of 6 months. Each session contained 8 to 10 exercises targeting major muscle groups of the core and the upper and lower limbs (2 - 3 exercises for each body district) (Fragala et al., 2019). Minimal equipment (body weight, chair, ab mat, elastic bands, and bottles of water) was used. Training periodization followed the principles of overload and progression: the training load was modulated by progressively increasing the volume (set x reps) of the exercises, the number of exercises (from 8 to 12), the strength requirement (e.g., the resistance of the elastic bands, from bi-podalic to mono-podalic exercise), and the postural challenge connected with the exercises (e.g. standing vs sitting) (ACSM guidelines, 2009). An example of the training progression is provided in Table 1.

Table 1. Example of workload progression in a weekly training session

| **Week 1** | | | | **Week 12** | | | | **Week 24** | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Exercise | Sets | Reps | Int. | Exercise | Sets | Reps | Int. | Exercise | Sets | Reps | Int. |
| Wall Push-ups | 2 | 8-10 | bw | Chair Push-ups | 4 | 10-12 | bw | Kneeling Push-ups | 4 | 12-15 | bw |
| Seated rower | 2 | 8-10 | yellow e.b. | Standing rower | 4 | 10-12 | red e.b. | Monolateral rower | 4 | 12-15 | black e.b. |
| Wall body french press | 2 | 8-10 | bw | Chair body french press | 4 | 10-12 | bw | Kneeling body french press | 4 | 12-15 | bw |
| Bicep curls | 2 | 8-10 | yellow e.b. | Monolateral Bicep curls | 4 | 10-12 | red e.b. | Monolateral Bicep curls | 4 | 12-15 | black e.b. |
| Sit to Stand | 2 | 8-10 | bw | Squat | 4 | 10-12 | bw | Bulgarian Squat | 4 | 12-15 | bw |
| Inverse lunges | 2 | 8-10 | bw | Lunges | 4 | 10-12 | bw | Chair single squat | 4 | 12-15 | bw |
| Glutes bridge | 2 | 8-10 | yellow e.b. | Sliding glutes bridge | 4 | 10-12 | red e.b. | Monolateral glutes bridge | 4 | 12-15 | black e.b. |
| Kneeling plank | 2 | 8-10 | bw | Plank | 4 | 10-12 | bw | Side plank | 4 | 12-15 | bw |
| Bird dog | 2 | 8-10 | bw | Banded anti-rotation | 4 | 10-12 | yellow e.b. | Banded anti-rotation | 4 | 12-15 | red e.b. |

The table outlines training variables (exercises, sets, repetitions, and intensity) for one of the three weekly target training sessions during the 1st, 12th, and 24th week. Reps: repetitions; Int.: intensity; bw: body weight; e.b.: elastic band.

The exercise program was designed and administered remotely through an innovative technological solution (Kari® system, Euleria, Trento) which included a web-based prescription interface for the trainer and a Tablet and a wearable inertial sensor (IMU-sensor) for each participant. The prescription interface allowed trainers to select from a menu of predefined exercises and, for each exercise, to customize the number of sets and repetitions. On the contrary, speed of movement (slow and controlled), time under tension (2 seconds for both concentric and eccentric phases), and rest periods (passive) were standardized for all exercises. The tablet displayed video instructions for the correct placement of the IMU sensor on the body (i.e. on the body segment that performs the movement) and the execution of the exercises, while the IMU sensor provided real-time feedback to the participant on the quality of the execution (i.e., the IMU gyroscope derived data checked about angular range of motion of the body segment around the targeted joint and speed of movement) by visualizing a feedback signal (Figure 1). In addition, the system calibrated the individual and exercise-specific maximal range of motion from the first 3 repetitions of the first set, and then used this reference for monitoring the quality of the execution of the remaining repetitions.

Figure 1. The figure illustrates the technological solution consisting of a tablet (the dedicated software displays video instructions and real-time visual feedback) and a wearable inertial sensor (magnetically attached to an elastic band).

Finally, the technological solution allowed the recording of the duration of the training sessions, training frequency (number of training sessions per week), number of exercises, sets, and repetitions completed, and the overall quality of the movement (i.e., an index of the overlapping between the target and actual movement pattern, as determined by the IMU-sensor).

Each participant received a 30-minute, one-to-one tutorial on using the technological solution.

*Data Collection and Analysis*

*Dropouts, Adverse Events, and incidents*

Drop out was defined as cessation of participation in the study protocol before it was completed. If a participant requested to interrupt their involvement in the study, the reasons were investigated and recorded. Adverse events were closely monitored during both physical tests and the intervention phase. Participants were instructed to use the technological solution warning system to report any difficulty, pain and discomfort experienced during training sessions. In case of warning report, participants were contacted to determine its origin and aetiology. An adverse event was defined as an intervention-related incident (such as muscle or joint soreness/stiffness) requiring a modification of the exercise program for 1 or more sessions (Fyfe et al., 2022).

*Adherence to the training program*

Adherence to the training program was evaluated based on the completion of the target training frequency (number of training sessions completed per week). To be considered "completed", a training session required that more than half (>50%) of the exercises and sets prescribed were actually performed. Subsequently, the weekly training frequency was averaged over the first and the second trimester of the intervention, for each participant and the group mean for each trimester was calculated. The percentage of participants who achieved the target and recommended number of training sessions (i.e. 3 and 2) every week was calculated, for each trimester.

*Anthropometric and Body Composition Measures*

Participants were asked to be barefoot and wear only underwear during anthropometric and body composition assessments. Body mass was measured using an electronic scale (Tanita electronic scale BWB-800 MA, Tokyo, Japan) with an accuracy of 0.1 kg. Height was measured with precision to the nearest 0.005 m using a Harpenden stadiometer (Holtain Ltd., Crymych, Pembs, UK). A dual-energy X-ray absorptiometry (DXA) scan was employed to assess total body composition, performed on a QDR Explorer fan-beam densitometer (Hologic Inc, Horizon C DXA System, USA). It was administered and analyzed using Hologic Discovery version 12.6.1 (Holtain Ltd, UK) (Nana, Slater, Stewart, & Burke, 2015). The body composition variables of interest included the percentage of body fat mass (%FM) and the appendicular lean mass index (ALMI) as the ratio between appendicular lean mass and height squared for all the participants.

*Physical test*

During the second visit to the laboratory, participants performed the following battery of tests in random order: i) 30 seconds of static balance, ii) 10 meters of straight walking, and iii) a loaded 5 sit-to-stand test. Before initiating the testing battery, all participants engaged in a 5-minute warm-up on a cycle ergometer (Monark 814 E, Monark, Vargerb SE) set at 50 watts (60 rpm) and completed four active lower-limb mobility exercises and 5-6 repetitions of the sit-to-stand movement (Bochicchio, Ferrari, Bottari, Lucertini, Cavedon, et al., 2023).

**Static balance test.** The body center of pressure was recorded by a force plate (1000Hz, AMTI Inc., Watertown, MA, USA) positioned under the participant's feet during 30” of static balance in a semi-tandem position (with the toe of the rear foot in contact with the front midfoot) with open eyes. After showing the correct posture, a familiarization trial was performed. An operator stayed near to the participant to prevent any risk of fall. Time was stopped at 30 seconds or if the participants moved their feet or grasped the operator for support.

Raw data were collected and subsequently analyzed with a self-written MATLAB code. Briefly, the force signal was low pass filtered at 5 Hz using a fourth-order Butterworth filter. After that, the ellipse area (cm2), anterior-posterior mean distance (mean distance AP), and mediolateral mean distance (mean distance ML) were extracted following the standard procedure (Prieto et al., 1996).

**10-meter walking test**. Participants were instructed to walk along a 10-meter straight path at a self-selected speed. Following a countdown, participants started walking from a standing position at 0.30 meters from the starting line. The start and finish time, velocity at 10 meters, cadence (in Hz), step length (cm), and percentages of double support (%) were measured with a validated system consisting of photocells (Witty gate, Microgate, Bolzano, Italy) integrated with 10, 1-meter photoelectric cells bars and dedicated software (Optogait, Microgate, Bolzano, Italy). Each trial was repeated twice. The system software automatically extracted the spatial-temporal parameters of gait, and the best trial (i.e., the lower time value) was recorded for further analysis.

**Loaded 5 sit-to-stand**. Participants began the test from a seated position on a 0.49-m height box and performed the test according to the following specific instructions: stand up and sit down from the chair 5 times, as fast as possible, with the arms crossed over the chest, making sure that the torso and shanks are perpendicular to the ground at the start of each repetition. Participants completed 2 sets of the 5STS test under 4 different weight conditions: body weight (BW), +12.5% BW, +25% BW, and +32.5% BW (Bochicchio et al., 2023). The added weight was obtained with a 0-30 kg adjustable weighted vest (Weight Vest bv30, Lacertosus, Parma, IT). The weight conditions sequence was randomized and counterbalanced and repeated twice, with a 3-minute rest between trials and a 5-minute break between conditions.

Ground reaction forces were recorded with a force plate (1000Hz, AMTI Inc., Watertown, MA, USA) placed under the participants’ feet. In addition, a marker was fixed on the greater trochanter to assess the kinematic variables of the movement using a motion capture system comprising 8 infrared cameras (100 Hz, Vicon, Oxford, UK). The force plate and motion capture system were synchronized during the entire data collection process and key variables (i.e. vertical force and velocity) were directly computed from Vicon software. Then, a second-order low-pass Butterworth filter was applied to the vertical force (cut-off frequency: 7 Hz) and velocity (cut-off frequency: 20 Hz).

Mean concentric force and velocity were computed following the procedure described in Bochicchio et. Al (Bochicchio, Ferrari, Bottari, Lucertini, Cavedon, et al., 2023) for each weight condition, in each participant. This allowed us to develop individual force-velocity (F-v, linear equation) and power-velocity (P-v, second-order equation) relationships and to calculate the muscle function indexes such as maximal force (F0, intercept between linear regression equation and y-axis), velocity (V0, intercept between linear regression equation and x-axis) and power (Pmax, apex of the second-order equation).

MatLab (Version R2021B, MathWorks Inc, Natick, Massachusetts, USA) scripts were employed for GRFs and kinematic signal analyses. Then, the variables of interest were exported in a Microsoft Excel spreadsheet (Microsoft 365, Version 16.0.16501.20228, Microsoft Corporation, Washington, USA) alongside anthropometric measures for further calculations.

*Statistical analysis*

Descriptive statistics were calculated and reported as mean ± standard deviation. Shapiro-Wilk test was run to test the normality of data distribution. For within participants' analysis of mean training frequency between the first and second trimesters, a paired t-test was run for the intervention group. In addition, one-way repeated measures ANOVA was run to compare the training frequency between the first and all subsequent weeks of training.

Anthropometric, body composition and physical performance measures at baseline (T0) between groups were compared by an unpaired t-test (for parametric data) or a Mann-Whitney test (for non-parametric data).

A one-way repeated measures ANOVA was used to determine changes within each group from the T0 time point for all the variables.

To test the effect of intervention, percentage differences from the baseline ( $\frac{T3\ or\ T6-T0}{T0} \times 100$) were computed for each variable. Then, changes between groups and time were analyzed by 2-way repeated measures ANOVA (Groups and Time), and Bonferroni correction was used for post-hoc analysis. Cohen effect size (*d*) was calculated as a measure of the magnitude (absolute values) of the within and between-group differences. Effect sizes (ES) were rated as trivial (< 0.2), small (< 0.6), moderate (0.6 < 1.2), or large (> 1.2). The level of significance was set at 0.05. The SigmaPlot 12.0 software (SigmaStat, San

Jose, CA, USA) was used to conduct all the statistical analyses. With a power of 0.80 and an α level of 0.05, 20 participants for each group were required to determine the between- and within-effect of the home-based training protocol based on a mean effect size of 0.30 (Chaabene et al., 2021; Mañas et al., 2021; Song et al., 2023) (G*power, Kiel, Germany). Given the long duration of the study, we recruited more participants than required for each group to overcome possible dropouts.

RESULTS

75 older adults of both sexes met the inclusion criteria. During the study, 2 participants dropped out due to health-related issues unrelated to the intervention protocol. 73 older adults of both sexes (37 males and 36 females) were therefore included in the final analysis. Of these, 46 participants (23 males and 23 females) were allocated to the intervention group (means ± SD: age 67.1 ± 5.8 years; body mass 74.2 ± 15.5 kg; height 1.68 ± 0.09 m) while 27 participants (14 males and 13 females) were allocated to the control group (means ± SD: age 66.6 ± 6.2 years; body mass 72.1 ± 13.8 kg; height 1.67 ± 0.10 m). No intervention-related adverse events were recorded. In addition, during the physical tests, only 2 participants were unable to perform all 4 conditions of the loaded 5 sit-to-stand. Therefore, these data were discarded. Figure 2 display the flowchart of the participant's screening and participation.

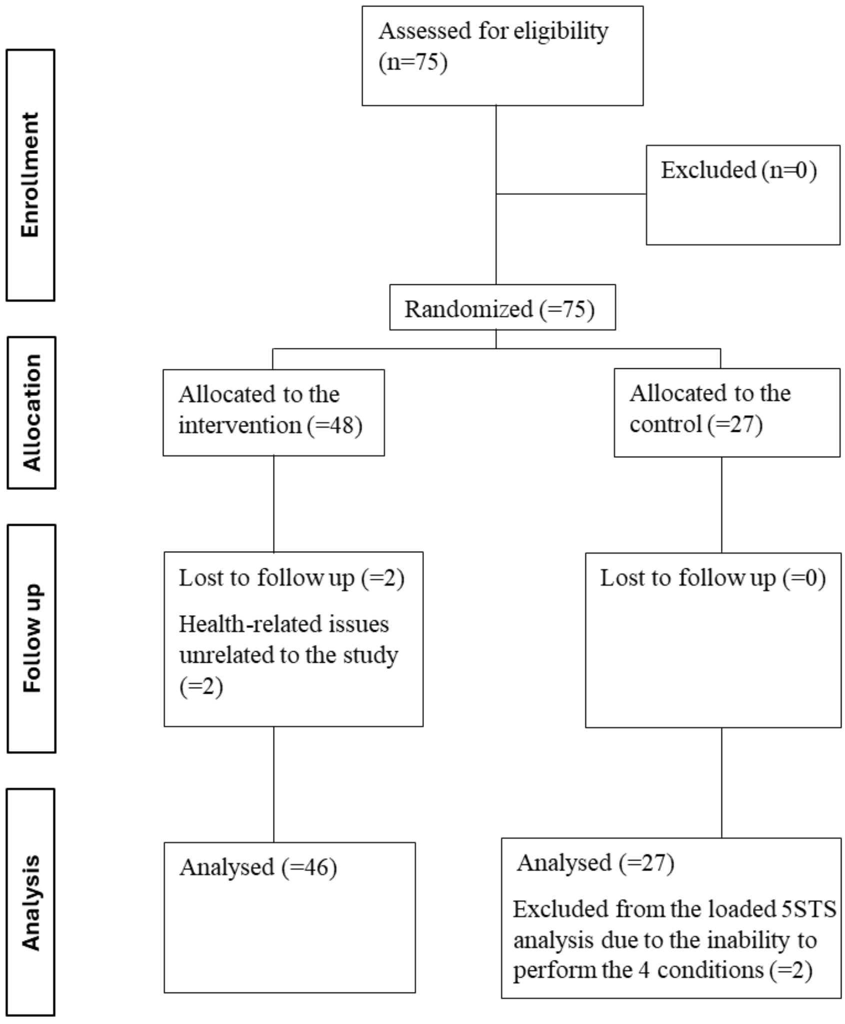

Figure 2. Participants flow diagram.

The mean training frequency of the first and second trimesters and weekly adherence to the training program are displayed in Figure 3. The mean training frequency during the second trimester was significantly lower than the first. In addition, 1-way RM ANOVA showed a significant drop in training frequency from the 16th to the last week of the intervention compared to the first week.

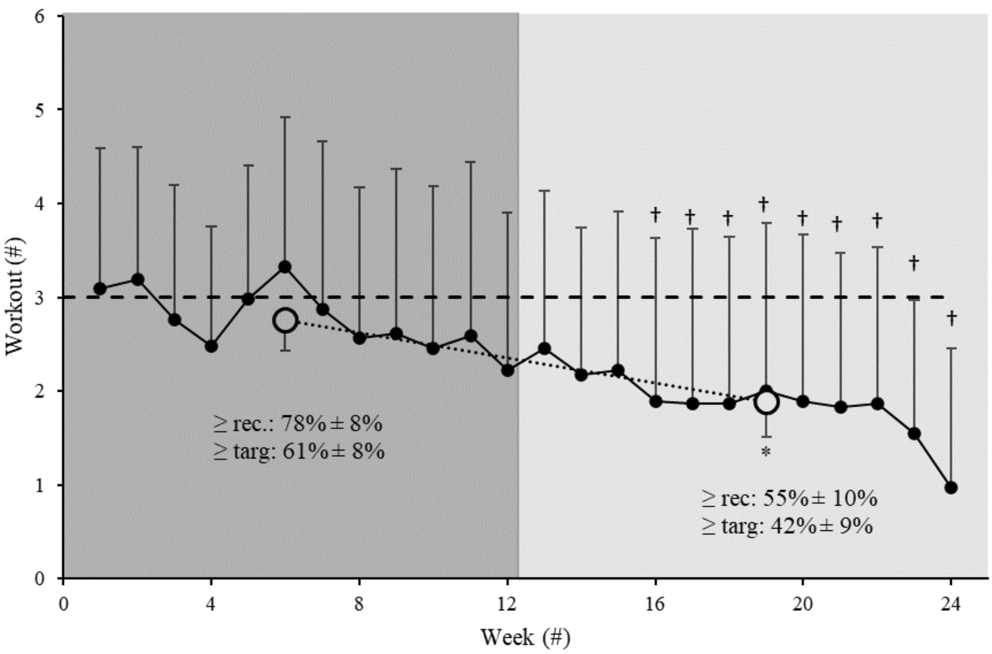


Figure 3. The graph displays the weekly adherence trend (●) to the training program along with the average number of weekly sessions performed (○) in the first (dark grey) and the second (light grey) trimester. In addition, the mean percentage of participants that completed the target (at least 3) and recommended (at least 2) training sessions per week are reported for each trimester. The bold dashed line represents the target workout frequency prescribed for the intervention program. * indicates significant differences between trimesters in mean training frequency (paired t-test on means). † indicates a significant difference from the training frequency of the first week (1way RM ANOVA). A significance level was set at $p < 0.05$.

Personal data, anthropometric, and physical measures of participants are reported in Table 2. No differences were found between the intervention and control groups at baseline.

Table 2. Individual characteristic and physical measures of participants at baseline (T0)

| Assessment | Variables | n | Intervention | n | Control | n | Total | *p-value* |
|---|---|---|---|---|---|---|---|---|
| Personal data | # (% females) | 46 | 50% | 27 | 48% | 73 | 49% | |
| | Age (yrs) | 46 | 67.1 ± 5.8 | 27 | 66.6 ± 6.2 | 73 | 66.9 ± 5.9 | 0.539 |
| Anthropometry and body composition | Height (m) | 46 | 1.68 ± 0.09 | 27 | 1.67 ± 0.1 | 73 | 1.68 ± 0.09 | 0.706 |
| | Body mass (kg) | 46 | 74.2 ± 15.6 | 27 | 72.1 ± 13.8 | 73 | 73.4 ± 14.8 | 0.558 |
| | ALMI ($kg{\cdot}m^{-2}$) | 44 | 7.2 ± 2.0 | 27 | 7.2 ± 1.2 | 71 | 7.2 ± 1.3 | 0.957 |
| | Body fat (%) | 44 | 31.7 ± 7.4 | 27 | 31.0 ± 8.0 | 71 | 31.0 ± 8.0 | 0.694 |
| Balance (Semi-tandem) | Ellipse area ($mm^2$) | 44 | 1499 ± 921 | 25 | 1380 ± 823 | 69 | 1456 ± 882 | 0.750 |
| | Mean distance AP (mm) | 44 | 9 ± 3 | 25 | 8 ± 3 | 69 | 8 ± 3 | 0.760 |
| | Mean distance ML (mm) | 44 | 6 ± 3 | 25 | 6 ± 3 | 69 | 6 ± 3 | 0.722 |
| Gait (10-meter walking) | Velocity ($m{\cdot}s^{-1}$) | 45 | 1.31 ± 0.28 | 27 | 1.38 ± 0.24 | 72 | 1.34 ± 0.22 | 0.183 |
| | Cadence ($step{\cdot}min^{-1}$) | 45 | 112 ± 19 | 27 | 115 ± 11 | 72 | 113 ± 11 | 0.174 |
| | Double support (%) | 45 | 27.1 ± 6.2 | 27 | 26.1 ± 3.6 | 72 | 26.7 ± 4.4 | 0.340 |
| | Step length (cm) | 45 | 70.1 ± 12.6 | 27 | 71.7 ± 9.1 | 72 | 70.7 ± 8.0 | 0.412 |
| Strength (Loaded 5 Sit-to-Stand) | F0 (N) | 44 | 1493 ± 499 | 25 | 1506 ± 671 | 69 | 1498 ± 461 | 0.616 |
| | V0 ($m{\cdot}s^{-1}$) | 44 | 0.86 ± 0.24 | 25 | 0.94 ± 0.34 | 69 | 0.89 ± 0.20 | 0.218 |
| | Pmax (W) | 44 | 321 ± 118 | 25 | 339 ± 138 | 69 | 327 ± 103 | 0.162 |

Mean ± SD of personal data, anthropometric, body composition, and physical measures are reported for intervention, control, and total group. ALMI, appendicular lean mass index; Mean distance AP, Antero-Posterior mean distance; Mean distance ML= Medio-Lateral mean distance; F0, maximum force; V0, maximum velocity; Pmax, maximum power. The p-values of the unpaired t-test (parametric data) or Mann-Whitney (non-parametric data) are displayed on the right side. The tests used for balance, gait and strength assessment are indicated in brackets.

One-way repeated measure ANOVA showed an effect of time in the intervention group for 10-meter walking speed (T3: $p<0.001$, ES= 0.53; and T6: $p<0.001$, ES= 0.59), cadence (T3: $p<0.001$, ES= 0.42; T6: $p<0.001$, ES= 0.51), step length (T3: $p<0.001$, ES= 0.43; T3: $p<0.001$, ES= 0.45), % double support (T3: $p=0.001$, ES= −0.23; T6: $p=0.001$, ES= −0.28), and F0 (T6: $p=0.009$, ES=0.26) compared to baseline. An effect of time was detected also for the control group at T3 for 10-meter walking speed (at T3: $p=0.044$, ES= 0.28) and step length (T3: $p=0.020$, ES= 0.25) compared to baseline.

Percentage changes in anthropometric and physical test parameters are displayed in Figure 4. The 2-way RM ANOVA showed only a significant interaction (TIME x GROUP) for F0. In particular, the intervention group showed greater delta change at T6 compared to T3 (+9.35% vs. +3.41%, $p=0.035$), which was also significantly different from the control group (+9.35% vs. +4.13%, $p=0.014$).

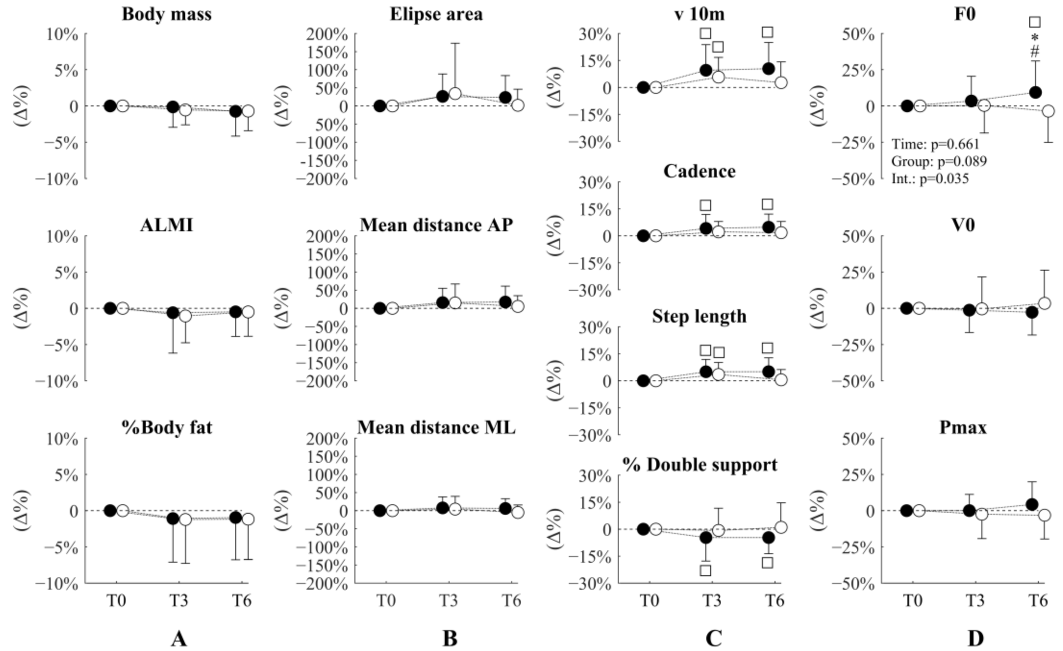


Figure 4. Black dots represent the Intervention group, while the white dots represent the control group. Comparison of mean percentage delta changes between trimesters are displayed for anthropometric characteristics (left side panel A), Walking (in the left-middle Panel B), Static sway (in the right-middle Panel C) and Loaded 5 STS test (right side Panel D) variables. □ indicates time-effect on the absolute values. The graphs displayed the main effects of Group, Time, and Interaction in a text box when significant differences were found ($p < 0.05$). For the post-hoc analysis, * indicates significant difference from T3-T0 ($p < 0.05$); while # indicates significant difference from control group ($p < 0.05$).

# DISCUSSION

This study tested the feasibility and effectiveness of a 6-month, self-administered, home-based resistance training program with an innovative device in healthy older adults of both sexes. Our results indicate that the exercise intervention delivered through the home-based device-supported is feasible in terms of safety and adherence. In fact, no adverse outcomes were recorded throughout the study, while the compliance was very good for the first 3 months, yet decreased markedly thereafter. Moreover, the training program had a marginal or no effect on body composition, balance, and muscle function indexes, except for the walking parameters and the maximal force of the lower limbs during the sit-to-stand task that displayed a modest yet significant increase.

Safety is one of the most challenging aspects of home-based training programs. Therefore, very easy (low coordinative challenge), low intensity, double support, and low postural challenge exercises are typically prioritized at the expense of potentially more effective movement tasks. Recent systematic reviews (Chaabene et al., 2021; Mañas et al., 2021; Song et al., 2023) showed that remote exercise interventions seem to be overall safe. Even if the training program used in our study contained several complex, standing, single-leg exercises, our data confirm that no training-related adverse events were registered during the whole protocol. We speculate that the very slow, controlled movements imposed by the device, through visual feedback on the articular range and movement speed, could have contributed to this positive result.

During the first trimester, we recorded an adherence to the training program (MTF: 2.8 ± 1.1, 61% of participants performed the recommended 3 sessions, 78% of participants performed at least 2 sessions per week) comparable to that described in the literature for programs that proposed 2-3 sessions per week (from 47% to 97% with a weighted average of 67%) (Mañas et al., 2021). However, a gradually decreasing adherence was observed (i.e., loss of 0.25 sessions per month) that brought the mean training frequency of the second trimester below the target value (3 sessions/week) and below the minimum recommended frequency for resistance training interventions (MTF: 1.9 ± 1.3 and 42% of participants performed the recommended 3 sessions while 55% of the participants performed 2 sessions per week). These observations suggest that long-term adherence to our program was possibly more difficult to maintain than in other studies (Chaabene et al., 2021). However, previous studies monitored adherence using training diaries filled out by the participants, an approach that is known to potentially overestimate this value (Chaabene et al., 2021). Moreover, many of the studies with the highest compliance included periodical direct contact with participants (i.e., via phone, internet, or personal visits) that may have contributed to maintaining a high adherence (Nilsson et al., 2020; Yamauchi et al., 2005). In our study, which was the first to measure objectively and automatically qualitative and quantitative adherence in real-time, this parameter should be free of overestimation.

*Effectiveness*

Our study was among the few (Lacroix et al., 2016; Maruya et al., 2016; Vitale et al., 2020) that measured anthropometric indexes of muscle mass and % body fat following home-based training programs. Resistance training is known to counteract the effect of aging on muscle mass (Chodzko-Zajko et al., 2009b; Geng et al., 2023; Zeng, Ling, Fang, & Lu, 2023). Muscle hypertrophy is mainly stimulated by metabolic stress and mechanical tension, activating intracellular pathways that induce muscle growth (Schoenfeld, 2010). Accordingly, our intervention was designed to progressively increase the volume (i.e., total amount of work) and the strength demands on a given muscle group (i.e., from dual to single limb) over time. However, in agreement with studies of similar duration and frequency (Lacroix et al., 2016; Maruya et al., 2016; Vitale et al., 2020), anthropometric indexes of muscle mass were unaffected by the training intervention (weight: $p= 0.169$; ALMI: $p= 0.404$).

Percent body fat is another health-related index that is threatened by aging and is potentially sensitivity to resistance training (Chodzko-Zajko et al., 2009b; Fragala et al., 2019). While traditional resistance training has been shown to positively affect the %BF in healthy older individuals ($ES= -0.53$, $p<0.001$) (N. Chen, He, Feng, Ainsworth, & Liu, 2021), our intervention group not displayed a significant reduction in %BF ($p=0.255$). Therefore, we can state that our intervention had no effect on body fat. These results appear more similar to those of other home-based studies that found small, non-significant effects on the fat mass (Maruya et al., 2016; Tsekoura et al., 2018).

In summary, the modest changes that were observed suggest that the overall training load delivered in our intervention may not have been sufficient to stimulate a gain in muscle mass and/or a reduction in body fat in our healthy older individuals.

The ability to control the body's center of pressure within the limits of the base of support is defined as balance (Yang et al., 2012). This ability decreases with aging and is one of the most critical factors associated with the augmented risk of falls in older individuals (Cyarto, Brown, Marshall, & Trost, 2008; Yang et al., 2012). Different forms of exercise training seem to positively affect balance (Cyarto et al., 2008) yet with somewhat inconsistent results. In fact, a recent systematic-review (Labata-lezaun, Rodr, Carlos, & Albert, 2023) that included in the analysis only studies whit interventions lasting 12 weeks, showed that traditional resistance training has no effect on the static balance ability in older individuals (ES= 1.99, p=0.19). On the other hand, previously described home-based resistance training protocols of comparable durations showed an overall small to modest yet significant effect (ES= 0.28-0.32) on balance (Chaabene et al., 2021; Lacroix et al., 2017; Mañas et al., 2021). In our study we did not find any improvement in all sway parameters (from p< 0.052 to p=0.504). Since the benefits of resistance training on balance are thought to be mediated by improved neuromuscular control (i.e., improved coordination between agonist and antagonist muscles, decreased variability of force, and more effective recruitment and synchronization of motor units) (Sousa, Silva, Lima-pardini, & Teixeira, 2013), we speculate that our program may have not been sufficiently intense or specific to effectively impact on this aspect. Therefore, whenever balance improvement is a desired outcome, either heavier intensity or else balance-specific exercise tasks should be incorporated in the training routine, the latter being easier and safer to administer in a home-based, unsupervised context.

Mobility can be broadly defined as the ability to move indoors and outdoors, with or without the use of some type of transportation (Webber, Porter, & Menec, 2010). In this context, mobility is a comprehensive term that includes different physical abilities and mental capacities. Among them, locomotion or walking ability is one of the most important. Indeed, during aging, a decrease in the mobility of older individuals occurs (Webber et al., 2010), and the reduced locomotion significantly impacts an individual's ability to engage in daily activities, making it a crucial factor influencing the lifestyles of older individuals (Song et al., 2023). A systematic review on home-based resistance training programs described no changes in gait parameters following training interventions (Mañas et al., 2021). In contrast to this, we found an improvement in all walking parameters (ES= −0.28-0.59; p≤ 0.001) that are comparable to the improvements described following traditional resistance training (i.e. gait speed ES= 0.42, p=0.008) (Lopez et al., 2018; Song et al., 2023). The larger increase observed in our study compared to other home-based resistance training studies could result from the concomitant lifting of COVID-19 restriction policies, allowing participants to be spontaneously more active. In fact, small improvements in 10-meter walking speed at T3 were observed even in the control group (ES= 0.28, p=0.043). Interestingly, the improvements in walking parameters were maintained for the entire 6 months only in the intervention group, indirectly confirming the long-term positive effect of our training intervention *per se*.

Muscle strength and power decrease with aging at a rate of 1.5% and 3–4% per year after 50, respectively (Skelton, Greig, Davies, & Young, 1994). Muscle power (i.e., the capacity to apply force quickly) seems to be more strongly associated with functional performance (i.e., ability to perform activities of daily living) than maximal strength (Gray & Paulson, 2014). Previous studies (Chaabene et al., 2021; Mañas et al., 2021; Song et al., 2023) showed an overall modest to small effect of home-based exercise interventions on muscle strength and power (ES= 0.30-0.34; ES= 0.43-0.44, respectively). Our results are partially in accordance with this evidence since we found an improvement for F0 (+9.35%, ES= 0.26) but not for maximal power (+4.25%, ES= 0.08) after 6 months of training in the intervention group. Although an increase in muscle power was not observed, a decrease was not observed either, highlighting that the training intervention may have mitigated the natural decline of muscle power that can be expected with aging. This view is corroborated by the observed significant decline in our control group (−4.88%, ES= −0.18). Slow muscle contractions have been shown to improve muscle strength in healthy older adults (Watanabe, Madarame, Ogasawara, Nakazato, & Ishii, 2014). In contrast, high-velocity muscle contractions are necessary and essential for improving peak muscle power (ACSM guidelines, 2009; Fragala et al., 2019). The slow and controlled execution imposed by the device could have provided an adequate stimulus to improve muscle strength, but it could have underexposed our participants to muscle power adaptations. Therefore, implementing a comprehensive resistance training intervention that includes power training may be a better approach to improving overall muscle function in healthy older adults, with the challenge of ensuring safety in performing high-velocity movement tasks in a home-based self-managed remote context.

In summary, our home resistance training delivered with a technological device achieved a negligible overall effect on body composition, balance, and muscle power, which are lower than previous studies conducted in a home environment, except for walking parameters and maximal strength, which were similarly and positively improved. Perhaps, in our fit and healthy population, the sole execution of bodyweight exercises or with the use of small equipment (i.e., an overall light relative exercise intensity) in conjunction with the slow speed of movement delivered by the device may have produced an overall sub-optimal intensity and limited the effectiveness of our home-based resistance training.

## CONCLUSION

In conclusion, our results indicate that the exercise protocol delivered with the technological solution is safe and associated with relatively good adherence for the first 3 months of intervention that decreased markedly thereafter. In addition, the 6-month of home-based resistance training positively affected walking parameters and the expression of the maximal force of the lower limbs during the sit-to-stand task while providing marginal or no effect on body composition, balance, and muscle power. While the extent of the long-term benefits of our home-based resistance training on muscle mass and function appear limited, exercise therapists and practitioners should consider this low-cost and accessible approach whenever barriers to an active lifestyle and participation in traditional resistance exercise programs are present.

# CHAPTER

# 2

## Temporal, kinematic, and kinetic variables derived from a wearable 3D inertial sensor to estimate muscle power during the 5 Sit to stand test in older individuals: a validation study.

**Temporal, kinematic and kinetic variables derived from a wearable 3D inertial sensor to estimate muscle power during the 5 Sit to stand test in older individuals: a validation study.**

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## ABSTRACT

The 5-Sit-to-stand test (5STS) is widely used to estimate lower limb muscle power (MP). An Inertial Measurement Unit (IMU) could be used to obtain objective, accurate and automatic measures of lower limb MP. In 62 older adults (30 F, 66 ± 6 years) we compared (paired t-test, Pearson's correlation coefficient, and Bland-Altman analysis) IMU-based estimates of total trial time (totT), mean concentric time (McT), velocity (McV), force (McF), and MP against laboratory equipment (Lab). While significantly different, Lab vs IMU measures of totT (8.97 ± 2.44 vs 8.86 ± 2.45 s, p=0.003), McV (0.35 ± 0.09 vs 0.27 ± 0.10 $m \cdot s^{-1}$, p<0.001), McF (673.13 ± 146.43 vs 653.41 ± 144.58 N, p<0.001) and MP (233.00 ± 70.83 vs 174.84 ± 71.16 W, p<0.001) had a very large to extremely large correlation (r=0.99, r=0.93, and r=0.97 r=0.76 and r=0.79, respectively for totT, McT, McF, McV and MP. Bland–Altman analysis showed a small, significant bias and good precision for all the variables but McT. A sensor-based 5STS evaluation appears to be a promising objective and digitalized measures of MP. This approach could offer a practical alternative to the gold standard methods used to measure MP.



KEYWORDS

IMU, sit-to-stand, validity, ageing, sensors, motion analysis, functional screening

## INTRODUCTION

Muscle function is defined by muscle strength, muscle power or performance in complex movements (e.g., walking speed) and is positively related to overall health, independence, and quality of life in aging [1]. Among the above indexes, the power of lower limbs is considered the stronger predictor of current and prospective muscle function [2]. Muscle power is lost at a rate of ~3.5% per year after 65th years [3], leading to a progressive loss in independence and mobility which in turn causes an inability to perform the activities of daily living (e.g., recovering balance, walking, sitting and standing from a chair) and further power loss [4]. This vicious cycle increases the risk of adverse health outcomes including falls, hospitalization, institutionalization, and mortality [3,5]. Therefore, it becomes crucial to provide clinicians with accessible tools for assessment and monitoring of the power of the lower limbs in ageing [5].

In health and exercise sciences, the assessment of lower limb muscle power in older adults can be obtained with specific equipment (isokinetic dynamometer, isotonic machines, Nottingham power rig) that requires standardized and relatively unnatural muscle actions [4]. In laboratory setting, Motion Capture system (MoCap) and force plate are considered the gold standard instruments to measure these variables during a variety of natural movements. However, this equipment is expensive, requires specialized personnel, as well as a time-consuming procedure for data collection and analysis. On the contrary, the ideal approach for testing and monitoring in clinical applications requires affordable costs, a relatively simple and time-efficient procedures, as well as the use movements that mimic muscle function in actual daily activities [4].

The 5 sit-to-stand test (5STS test) was developed as a time and cost-efficient field-test for the estimation of power of the lower limbs. Muscle power estimates with 5STS are highly correlated with indexes of functional fitness (e.g., longer time up and go, grip strength, dynamic balance, stair climbing, gait speed) [4,6], frailty [7], and health-related quality of life [8]. While this simplified approach has an undiscussed practical value for testing and periodical monitoring of muscle function, its accuracy and precision may be affected by the accuracy of the time measures (due to manual stopwatch measurement and recording) as well as by the assumptions on which it is based [9]. In this context, body-worn sensors could provide the opportunity to perform objective and digitalized measures of movement and possibly accurate estimates of muscle power. Among body-worn sensors, wearable inertial measurement units (IMU) can be used to record kinematic and kinetic information during a wide range of human movements. Several studies used this technology during the

5 STS test to discriminate between fallers and not fallers [10] or healthy vs unhealthy individuals during a single sit-to-stand task [11–17]. While these studies appear promising, a study comparing the validity of the measures of muscle power from the 5STS test based on a 3D inertial sensor (IMU) with a complete gold standard measurement setup (MoCap and force plate) is lacking.

Therefore, the purpose of this study was to assess (1) the accuracy and precision of the IMU-estimated time, velocity, and force and (2) verify the accuracy and precision of the estimates of lower limb muscle power compared to the gold standard laboratory instruments. Our hypothesis was that IMU could correctly measure duration, velocity, and force. In addition, we hypothesized that lower limb muscle power could be accurately estimated maintaining the ease of use of the field 5STS test but with an objective and automatically digitalized measures.

## METHODS

### *Study design*

For the purpose of the study, we used a validation design to examine (1) the accuracy and precision of the IMU-estimated time, velocity, and force and (2) verify the accuracy and precision of the estimates of lower limb muscle power compared to the gold standard laboratory instruments). All participants visited the laboratory once in which we took anthropometric measures and performed the 5STS test evaluation. IMU and laboratory instruments were used simultaneously for the measure of temporal, kinematic, and kinetic variables during the task.

### *Participants*

62 independently older adults were recruited by local advertisement (Table 1). The inclusion criterion was age above 60 years. Exclusion criteria were evaluated with a preliminary telephone interview and a successive medical screening to exclude individuals with any orthopedic, mental, or neurological disease that could have interfered with the ability to express lower limb maximal power or a Short Physical Performance Battery (SPPB) score below (9). All participants signed a written informed consent prior to participation. All procedures used in the study were approved by the Ethics Committee for Human Research from the University of Verona (28/2023) and conducted in conformity with the Declaration of Helsinki.

### *Data collection*

During the visit to the laboratory, participant’s anthropometric measures were collected prior to 5STS test. The anthropometric assessment was performed with participants barefoot and wearing only underwear. Body mass was taken to the nearest 0.1 kg with an electronic scale (Tanita electronic scale BWB-800 MA, Tokyo, Japan) and stature was measured to the nearest 0.01 m with a Harpenden stadiometer (Holtain Ltd., Crymych, Pembs, UK). Body Mass Index (BMI) was calculated as body mass · height-2 (kg·m-2).

Participants performed the 5STS test for lower limb muscle power determination. Immediately before the test, each participant performed a 10-min warm-up protocol, consisting of 5-min cycling on a cycle-ergometer at a fixed power and cadence (i.e., 50W at 60 rpm), 4 active mobility exercises for the upper and lower limbs, and 5-6 repetitions of the sit-to-stand movement, which was also considered as a familiarization to the 5STS test. The participant was sitting on a box (high 0.49 m) with the trunk and shanks positioned perpendicular to the ground and the arms crossed on their chest. The 5STS test consisted of 5 consecutive repetitions of the sit-to-stand movement, executed as fast as possible. The test was performed twice, with 3-min recovery between the two trials. The trial started after a countdown of three and ended when the participants touched the seat after the fifth repetition [9].

A wearable IMU containing a 3D inertial sensor in combination with a 3D gyroscope and 3D magnetometer (500Hz, Gyko, Microgate, Bolzano) was attached to the lateral face of the right thigh of each participant.

A MoCap was used with 8 infrared cameras (100Hz, Vicon, Oxford, UK) automatically synchronized with a force platform (1000Hz, AMTI Inc., Watertown, MA, USA) positioned under the participants' feet. To build a model of the thigh, three markers were placed respectively on the trochanter, mid-thigh (in correspondence with the device), and lateral epicondyle of the femur.

The IMU and laboratory instruments were recording simultaneously during the trials.

*Data Analysis:*

2.4.1 IMU data analysis

Raw data were collected by using the instrument's dedicated software and subsequently analysed with a self-written MATLAB code. The accelerometer and gyroscope were set to a full scale of 4g and 2000°·s-1, respectively. The calibration matrix provided by the company was used to convert the raw data into bits to acceleration in m·s-2 and angular velocity in rad·s-1. To reduce the integration error, the initial offset of the gyroscope data was removed. Data from the 3D gyroscope and accelerometer were filtered using a low-pass, second-order Butterworth filter with a 30Hz cut-off frequency. The cut-off frequency was chosen after a frequency analysis of the signal revealed that there was no significant information above it. The quaternions which described the orientation of the sensor over time, were computed using the Mahony filter. Euler angles according to the notation ZYX were extracted from the quaternions [18]. The angle around the principal axis of motion was considered for the subsequent analysis (see Figure 1).

The peaks of the Euler angle that corresponded to the maximum rotation of the sensor from the initial position were found. The beginning and the end of the sit-to-stand movement were detected by using a threshold corresponding to the 5% of the peaks previously individuated. The duration of all the phases was estimated from these data as follows:

- Sit to Stand transitions = from the threshold at 5% to the peak of the rotation.

- Stand to Sit transitions = from the peak to the subsequent passage on the threshold of 5%.

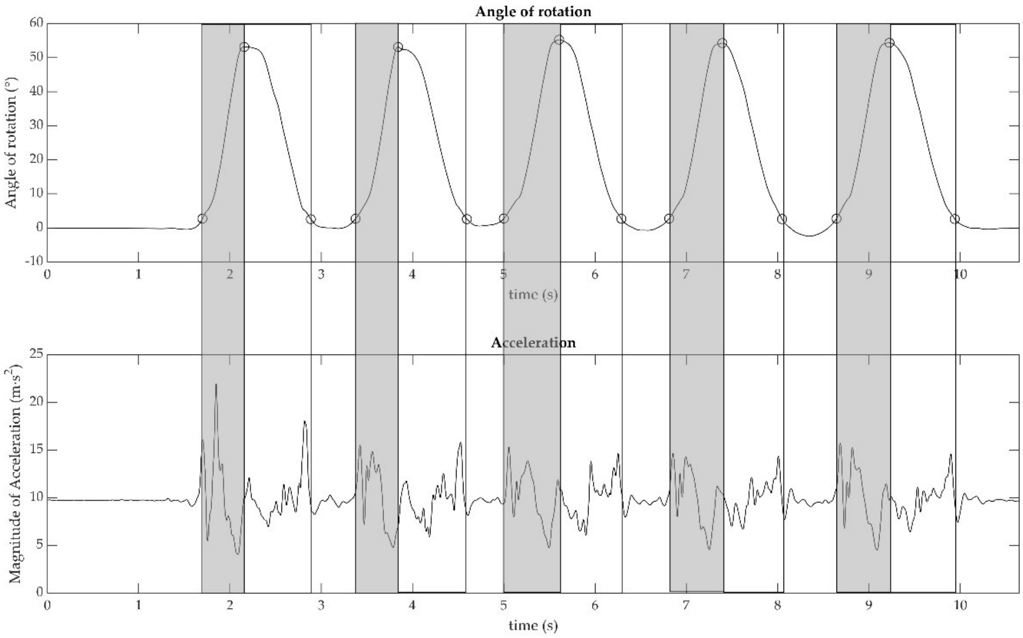



**Figure 1.** Angle of rotation around the main axis (expressed in degree °, upper graphic) and Magnitude of acceleration (m·s$^{-2}$, lower graphic) are plotted as a function of time during the 5 Sit to Stand test in a representative subject. The points (○) mark the events of each repetition (start, standing position, and end). Light grey sections highlight the concentric phases (i.e., raising phase) while white sections highlight the eccentric phases (i.e., sitting phase).

*Laboratory instruments data analysis*

Participants' kinetic and kinematic variables were measured using the force platform and the MoCap system (on the z-axis, perpendicular to the ground). Mid-thigh vertical velocity was automatically extrapolated by the system and used to make all the subsequent computations. Vertical velocity and force signals were low pass filtered at 7 and 15 Hz, respectively, using a second-order Butterworth filter. The identification of the concentric and eccentric phases, as well as the total trial duration, is depicted in Figure 2. Vertical velocity was used to recognize the concentric and eccentric phases, as well as the total trial duration. Positive and negative peaks of vertical velocity were identified and thresholds as 5% of peaks were calculated as follows: 1) the start and end of repetition were found when the vertical velocity reached the positive and negative threshold, respectively; 2) the end of the concentric phase was defined when vertical velocity crossing the zero after positive peak; 3) the start and the end of the total trial were defined as the first positive 5% and the last negative 5% respectively.

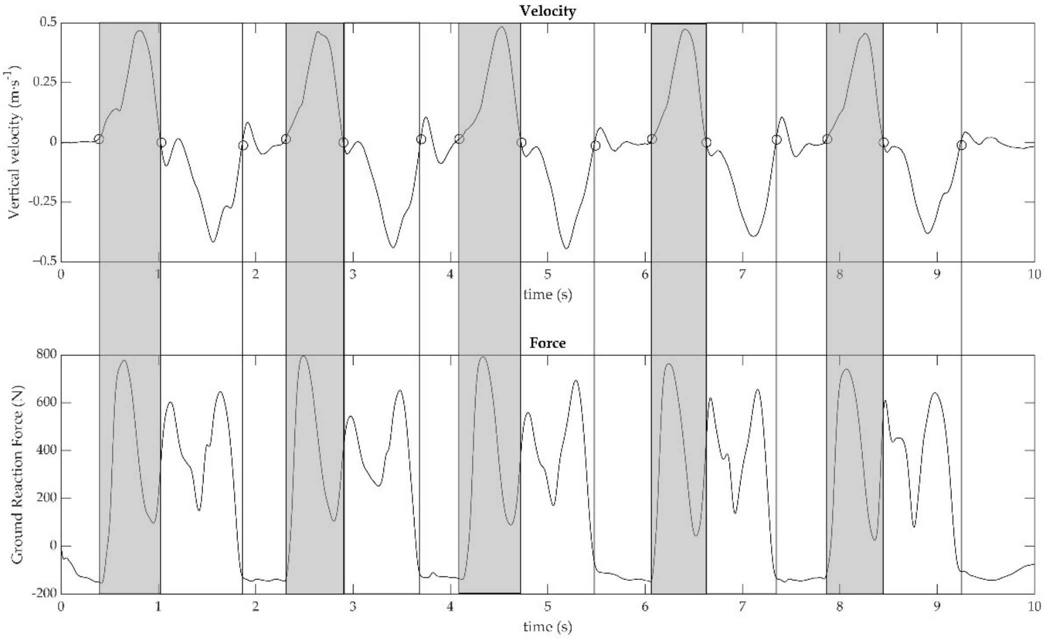



**Figure 2.** Vertical velocity (m· $s^{-1}$; above) measured by motion capture and vertical force (N; below) measured by force plate in the 5 Sit to Stand test are plotted in function on time (s) in a representative participant. The points (○) mark the events of each repetition (start, standing position, and end). Light grey sections highlight the concentric phases (i.e., raising phase) while white sections highlight the eccentric phases (i.e., sitting phase).

*Outcome measures*

2.5.1 IMU calculations

For the force and velocity estimation, the effect of gravity on the data from the IMU was compensated by rotating the raw acceleration data to align the z-axis with the direction of gravity. The magnitude of the acceleration was then calculated and after that, the following parameters were computed:

• Time: the total duration of the 5STS test (totT) was computed as the difference between the time coordinates of the first and last 5% thresholds. The duration of the single concentric phases was calculated as the difference between the time coordinates of the positive 5% threshold and the peak of the rotation. Thereafter, the duration of the five single concentric phases was averaged (McT).

- Velocity: data of the acceleration were segmented by using the instant of beginning and end of the movement previously computed (from the threshold at 5% to the peak of the rotation). These segments were integrated using the Simpson method to obtain the velocity of the movement. The average velocity for each sit-to-stand movement was then extracted (McV).

- Force: The mean concentric force (McF) was calculated by multiplying the average of the acceleration within each concentric phase with 90% of the body mass of the subject [19].

- Power: for each repetition, the lower limb muscle power (MP) was computed as the product between mean concentric velocity and mean concentric force.

***Laboratory instruments calculations***

- Time: total duration of the 5STS test was computed as the difference between the time coordinates of the first positive and last negative 5% thresholds. The duration of the single concentric phases was calculated as the difference between the time coordinates of the positive 5% threshold and vertical velocity crossing the zero after the positive peak. Thereafter, the duration of the five single concentric phases was averaged.

- Velocity: velocity was automatically computed from the Vicon. Mean concentric velocity was calculated as the average of velocity signal within the duration of each concentric phase (Figure 2, top panel). Thereafter, the velocity of the five single concentric phases was averaged.

- Force: mean concentric force was calculated as the average of the ground reaction force signal within each concentric phase (Figure 2, bottom panel). Thereafter, the force of the five single concentric phases was averaged.

- Power: for each repetition, lower limb muscle power was computed as the product between mean concentric velocity and mean concentric force.

*Statistical Analysis*

For the statistical analysis, only the fastest trial for each participant was used. All data were checked for normality using the Shapiro–Wilk test. An unpaired t-test was run to compare anthropometric variables between females and males. Paired t-test, Pearson correlation coefficient, and Bland-Altman analysis were run to test differences and absolute level of agreement between IMU estimates and laboratory measures of totT, McT, McV, McF, and MP. The correlation coefficient was interpreted according to the values of the r: trivial (<0.1); small (0.10–0.29); moderate (0.30–0.49); large (0.50–0.69); very large (0.70–0.89); extremely large (0.90–1.00) [20].

The Bland–Altman analysis was followed by a one-sided z-test on the bias. Bland–Altman analysis [21] was used to determine potential systematic bias, reporting mean bias, limits of agreement (LOA), and coefficient of determination ($R^2$) from regression analysis between differences and means of IMU and laboratory measures of power. Data are reported as mean ± SD. The level of significance was set at 0.05. The SigmaPlot 12.0 software (SigmaStat, San Jose, CA, USA) was used to conduct all the statistical analyses.

## RESULTS

All the participants were able to perform the whole procedure properly and it was well tolerated and no adverse events were recorded. In addition, no trials have been discarded.

Anagraphics, anthropometric measures and SPPB scores of the recruited subjects are reported in Table 1. Duration, velocity, force, and power variables of all the participants are reported in Table 2.

**Table 1**. Anthropometric characteristics of the participants.

| | Females | | | Males | | | Tot | | | *p-value* |
|---|---|---|---|---|---|---|---|---|---|---|
| # | 30 | | | 32 | | | 62 | | | |
| Age (yrs) | 65.3 | ± | 5.2 | 68.0 | ± | 6.3 | 66.7 | ± | 5.9 | 0.072 |
| Weight (kg) | 65.5 | ± | 11.4 | 81.8 | ± | 14.6 | 73.9 | ± | 15.4 | < 0.001 |
| Height (m) | 1.61 | ± | 0.06 | 1.74 | ± | 0.07 | 1.68 | ± | 0.09 | < 0.001 |
| BMI ($kg{\cdot}m^{-2}$) | 25.2 | ± | 4.1 | 27.1 | ± | 5.0 | 26.2 | ± | 4.6 | 0.112 |
| SPPB score | 11.0 | ± | 0.9 | 11.0 | ± | 0.7 | 11.0 | ± | 0.8 | 0.580 |

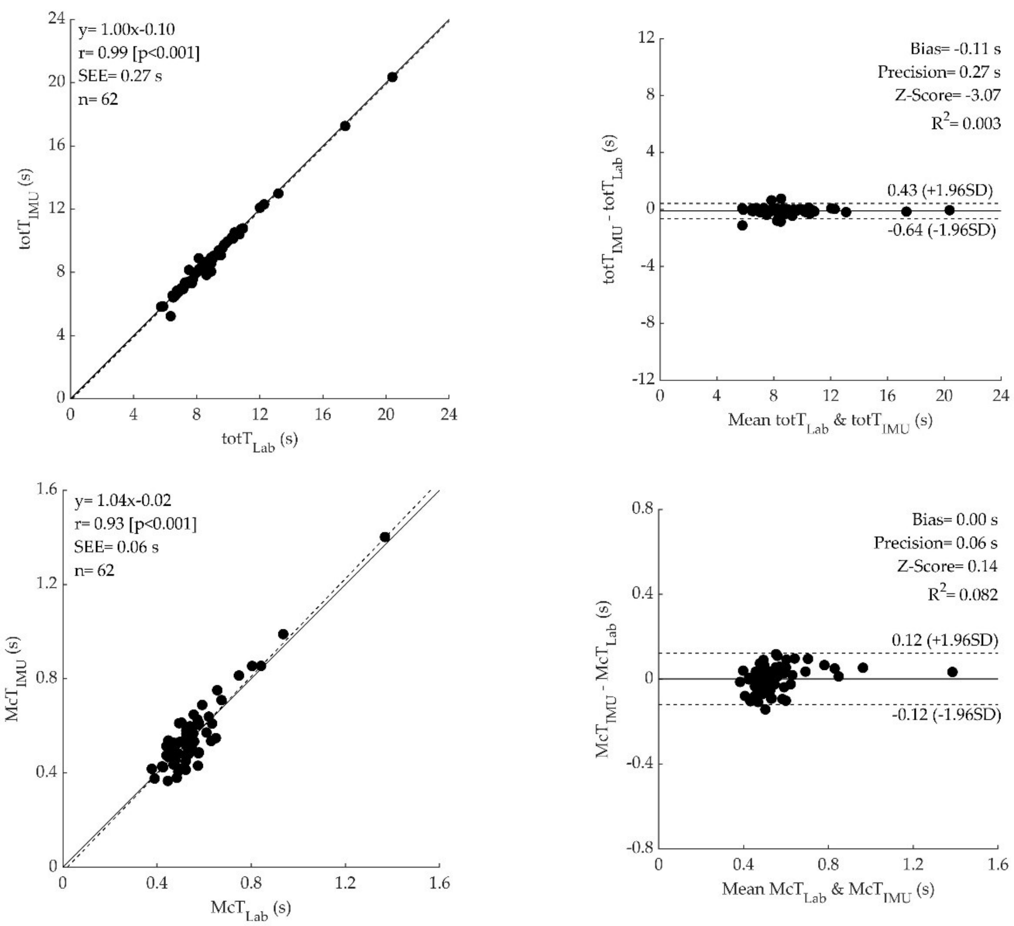
BMI, Body Mass Index; SPPB score, short physical performance battery score. Significant differences between sexes are indicated with the p-value <0.05 (unpaired t-test on means).

**Table 2**. Kinetic and kinematic variables of all participants measured with Lab and IMU methods.

| | | Lab | | | IMU | | | *p-value* |
|---|---|---|---|---|---|---|---|---|
| # | | 62 | | | 62 | | | |
| Duration | totT (s) | 8.97 | ± | 2.44 | 8.86 | ± | 2.45 | 0.003 |
| | McT (s) | 0.56 | ± | 0.15 | 0.56 | ± | 0.16 | 0.890 |
| Velocity | McV ($m{\cdot}s^{-1}$) | 0.35 | ± | 0.09 | 0.27 | ± | 0.10 | <0.001 |
| Force | McF (N) | 673.13 | ± | 146.43 | 653.41 | ± | 144.58 | <0.001 |
| Power | MP (W) | 233.00 | ± | 70.83 | 174.84 | ± | 71.16 | <0.001 |

Lab= laboratory method; IMU= 3D inertial sensor method; totT= total time; McT= mean concentric time; McV= mean concentric velocity; McF= mean concentric force; MP = mean concentric power. Significant differences are indicated with the p-value <0.05 (paired t-test on means).

Paired t-test showed a significant difference for all the variables ($p < 0.05$) except mean concentric time ($p = 0.890$). Pearson correlation coefficient (Figure 3, left panel) showed an extremely large correlation for the total time trial, mean concentric time, and mean concentric force (r=0.99, r=0.93, and r=0.97 respectively) and very large for mean concentric velocity (r=0.76), and mean concentric power (r=0.79). Bland–Altman analysis showed a significant bias (for all parameters but mean concentric time, that displayed a non-significant bias) and good precision for all the variables (Figure 3, right panel).



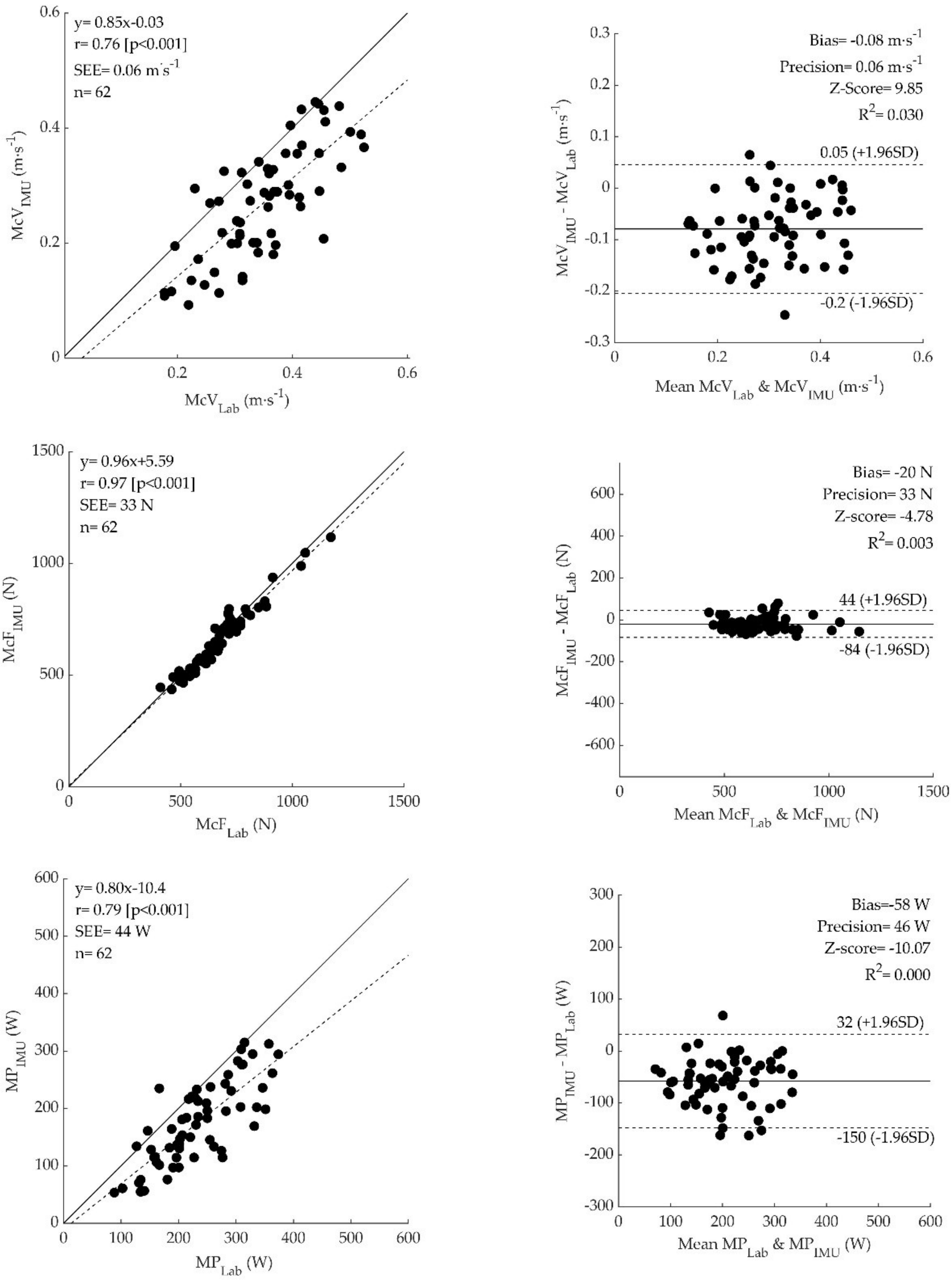



**Figure 3.** The left side of the figure shows the correlation plots between laboratory and IMU measures of 5 sit-to-stand total time (totT), mean concentric time (McT), mean concentric velocity (McV), mean concentric force (McF) and mean power (MP). Equation, Pearson's correlation coefficient (r), p-value, SEE and sample size are reported along with regression (dashed) and identity (solid) lines. The right side of the figure shows the Bland Altman analysis between laboratory and IMU of the same variables. Individual differences between lab and IMU measures are plotted as a function of the mean of the two. Bias, $R^2$, and Z-score are reported along with the LOA (dashed lines) and bias (solid lines).

# DISCUSSION

This is the first study that tested the accuracy of an IMU in estimating muscle power during the 5STS test in comparison with a fully objective, gold standard, and automated method. Our data indicate that a single IMU placed on the lateral face of the thigh provides estimates of kinetic and kinematic indexes of muscle action, as well as of muscle power, during the 5STS test which are highly correlated with gold standard laboratory measures. While a significant difference was recorded between measures, likely due to the sensor placement, IMU appears to offer a promising practical alternative to the gold standard methods used to measure MP.

The present results showed that an automated analysis of the instrumented 5STS using a 3D inertial sensor is feasible. Indeed, none of the 62 trials have been discarded due to signal problems and our algorithm was able to correctly identify each phase (i.e. concentric and eccentric phase) of the 5STS and successfully extract time, velocity, and force variables from the 3D inertial sensor signals of all trials.

## 4.1. Time

The value of Total time from the laboratory approach ($8.97 \pm 2.44$ s) and IMU ($8.86 \pm 2.45$ s) obtained in our study are comparable to those found in literature (from ~8 s to ~15 s, for not fallers and fallers of both sexes, 65-90 years old) [7,9,10,16,22,23]. In our study, we found a value of IMU-based Total time that differs from the time measured with the laboratory instruments ($p=0.003$). While statistically significant, the absolute difference is ~0.12 s (i.e. 1.2% of the average measure) which is less than the minimum detectable change for the 5STS test [24]. As such, this difference could be interpreted as not clinically or practically relevant. In addition, we found an almost perfect correlation ($r = 0.99$) between the total time of IMU and laboratory measures. Finally, the bias and the limits of agreement (bias = -0.11 s, z-score = -3.07, LOA= [0.43 -0.64 s]) were lower than the values found in other studies (bias = 0.48 s, LOA = 0.32 s) [16] and a null relationship was observed ($R2=0.003$) indicating that the difference between methods is similar across the entire range of time measures. Therefore, we can conclude that IMU is a valid approach to measure the total time of a 5STS test.

Instrumenting the 5STS test could be a cornerstone in the muscle function assessment because the instrumentation would provide more information than a simple chronometer used for collecting the time to complete the task. Indeed, a wearable device could discriminate between each repetition and within them, each standing or sitting phase, returning, for example, the time effect of the muscle fatigue or the variability within the task [25,26]. In our study, we found a lower value of mean concentric time for laboratory instruments (0.56 ± 0.15 s) compared to literature (range from ~0.8 to ~1.5 s) [27–29]. Conversely, greater mean concentric time values were found by the IMU (0.56 ± 0.16 s) compared to literature (0.41 ± 0.20 s) [30]. These discrepancies may be due to the different methods used for discriminating the standing phase, to the different types of population studied (healthy older subjects vs random community-dwelling, individuals affected by stroke) or to the different heights of the chair (in our case fixed at 49 cm vs adjusted according to participants anthropometry).

Within our study, IMU-based measures of mean concentric time were not different, very highly correlated ($r = 0.93$), without significant bias (bias = 0.00, z-score =0.14), and with small limits of agreement (L.O.A.= [0.12 s -0.12 s]) compared to the gold-standard laboratory approach. This would indicate that the method used for the data analysis of the IMU signal identifies the same “temporal windows” as the laboratory approach. Since we calculated the values of force and velocity within these temporal windows, we can conclude that the differences in these variables are not due to the difference in phase discrimination or instrument synchronization. Therefore, the IMU is a valid method to measure accurately and precisely the mean concentric time and these values can be used during clinical assessments.

4.2. Velocity

In our study, we found values of mean concentric velocity that are lower compared with the values of other studies for both laboratory equipment (0.35 ± 0.09 m·s-1 vs ~0.50 m·s-1) [17,23,31] and IMU (0.27 ± 0.10 m·s-1 vs ~0.65 m·s-1) [17,32]. This difference cannot be explained by age or low function indexes [33] since our population sample is younger compared to the references (66.7 ± 5.9 years vs 71 to 78 years) and characterized by high functionality as indicated by the average SPPB score (Table 1) [23,32]. Therefore, this discrepancy may lay in different methods used to calculate the velocity (Video, MoCap or force plate). Indeed, the most likely source of discrepancy may be that we calculated the mid-thigh velocity instead of the centre of mass velocity (i.e., trochanter level) or trunk velocity. Despite the angular velocity is the same during a rotation of a rigid segment independently of the distance from the fulcrum, the tangential velocity is directly influenced by the radius ($v = \omega$*radius). Since we calculated the velocity at the mid-thigh (nearer to the knee i.e., the fulcrum of the thigh rotation), we found lower values of linear velocity compared to the values calculated in literature. Indeed, if we consider a distance from the knee that is two-fold the one we used with our mid-thing placing (i.e., thigh length), the linear velocity could become roughly the value found in the literature (~0.55 m·s-1).

In our study, we found that the mean concentric velocity calculated by the IMU (0.27 ± 0.10 m·s-1) was lower ($p < 0.001$) than the values calculated by the laboratory equipment (0.35 ± 0.09 m·s-1). This discrepancy could lie in three methodological matters:

- The procedure of the phase's identifications. To discriminate each phase and repetition of the 5STS test from the IMU and the laboratory equipment, we considered the Euler angle and the marker's vertical velocity signal, respectively. The same method for phase identification was applied to both signals (see method section). Therefore, the differences observed in mean concentric velocity could lie in the fact that we applied the same methodological approach to two distinct types of signals.

- Another source of discrepancy may lie in the process used to compute the velocity from the IMU raw data. Indeed, the velocity was calculated by numerical integration of the acceleration where zero values were imposed as integration limits considering null the velocity at the beginning and the end of the standing motion. However, since we used the time boundaries from the Euler angle signal (as shown in Figure 1), the velocity in correspondence with these limits during the standing movement was not exactly zero. This could lead to an underestimation of mean concentric velocity.

- Amplitude of the sensor-based acceleration signal. Multiple studies in literature tried to estimate the ground reaction force by using accelerometer data. They discovered that, the closer the IMU is to the fulcrum, the smaller the amplitude of the acceleration signal measured [34]. Therefore, the mid-thigh IMU may have been subjected to lower absolute values of accelerations.

Overall, our results indicate that, while IMU slightly underestimates the velocity of the movement compared to gold standard methods, possibly due to methodological matters, it displays a very high correlation (r=0.76), significant and constant bias (bias= -0.08 m·s-1, z-score= -9.85, R2=0.030) with small limits of agreement (L.O.A. = [0.05 -0.20 m·s-1]). Then, the IMU could be considered overall a valid and precise instrument for the estimation and monitoring of mean concentric velocity.

4.3. Force

In this study, the mean concentric force values for both methods (~ 510-790 N) were similar to the values found in the literature (~550-700 N) [9,23,36–38]. Regarding the comparison between methods, we found that the force values were different ($p < 0.001$) but with almost perfect correlation ($r=0.97$), a very small significant bias (bias = -20 N; z-score = -4.78), and small limits of agreement (L.O.A. = [44 -84 N]). It is worth mentioning that the difference between methods corresponds to only ~2 kilograms and therefore could be considered not practically relevant. Previous studies found that during the sit-to-stand task, not all the body mass of the subject is accelerated, and different percentages of the body mass were found (90%, 87%, and 67% of body weight) [9,19,23]. The differences in these values could be due to different equipment, methods of phase discrimination, and assumptions used to identify the concentric phase. In this study, we considered that only 90% of body mass was accelerated and therefore, for the IMU, we calculated the mean force by multiplying this value by the mean concentric acceleration. Furthermore, most of the studies use more than one IMU sensor and develop models to describe the body as a linked chain of multiple elements in order to correctly estimate the force [35]. The use of one single sensor on the thigh could thus have determined a systematic underestimation of the absolute value of force.

### 4.4. Power

Peak values (~700-900 W) [27,39] or mean values (~300-600 W) [2,31] of the 5STS assessed with different instruments (wearable devices, force plate, motion capture, stopwatch) found in the literature are greater than our results (laboratory equipment = 233.00 ± 70.83 W; IMU = 174.84 ± 71.16 W). Since power was computed as the mechanical product between force and velocity, and considering the force was similar to the literature's values, power differences are attributable to our lower mean concentric velocity measured at the mid-thigh.

In addition, IMU power values were significantly lower than the values measured with laboratory equipment ($p < 0.001$). As above mentioned, power difference is attributable to the lower IMU mean concentric velocity measured at the mid-thigh since the force was similar to the laboratory's values. This discrepancy could be attributable to the placement chosen for the IMU on the participants which is less prone to greater accelerations. While the bias was significantly different, the limits of agreement were relatively small (bias = -58 W, z-score = -10.07, LOA = [32 -150 W]). In addition, a null relationship was observed ($R2 = 0.000$) indicating that the difference between instruments is similar across the entire range of power.

### 4.5. Limitations and future developments

The present study has some limitations. We placed the IMU on the lateral face of the mid-thigh, and this could have led to lower values of velocity and power compared to the literature. The most common sensors/markers placements vary from the front sternum or chest (27%) to the back trunk (57%) (i.e., L5-L3) or on the lower limb (44.5%) (such as the thigh, shank or ankle) [40]. We choose the mid-thigh among others because it is more accessible and facilitates the placing of the IMU even in the presence of a weight vest (in view of a possible future application of the method). The back trunk positioning was excluded because we wanted to isolate the sole leg muscle power by avoiding the possible noise from the oscillations of the trunk [13]. While the sensor placement does not jeopardise the comparison between methods, which is the focus of our study, it is plausible that the muscle power measured at the middle-thigh underestimates the lower limb's ability to express power. Therefore, a mathematical model for estimating the whole system power (i.e.: centre of mass or trochanter) by wearing the IMU on the lateral face of the mid-thigh could be considered for future studies.

Another limitation is that we used the same chair (chair height = 0.49 m) for all the participants and therefore we did not control for the individual articular angle/muscle length. Indeed, fixed-height chair leads to different articular angles/muscle lengths for people with different leg length; in turn, this may lead to a different expression of force over time [41]. While this is unlikely to affect the correspondence between methods of measurement, the standardisation of knee angle between subjects will facilitate the interpretation of data across individuals with different anthropometric characteristics (i.e., short vs long lower limb length).

## CONCLUSION

The 5STS test is a valuable and common test used to estimate lower limb muscle power thanks to its simplicity and low cost. However, in its field version (i.e., human eye and stopwatch), it is not independent of the operator's error. In our study, we find a not significant bias between measures of kinetic and kinematic parameters and muscle power measured with either a wearable IMU or a full, gold-standard laboratory approach. Therefore, a wearable sensor-based 5STS evaluation could allow a valid, low-cost alternative to the gold standard methods classically used to measure muscle power; moreover, it could be a more repeatable, objective, and immediately digitalized option compared to the stopwatch method.

# CHAPTER

# 3

## Loaded 5 Sit-to-Stand Test to determine the Force –Velocity Relationship in Older Adults: A Validation Study.

**Loaded 5 Sit-to-Stand Test to determine the Force–Velocity Relationship in Older Adults: A Validation Study**

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## ABSTRACT

Force–velocity profiling (F-v) evaluates muscle function through the identification of maximum force (F0), velocity (V0), power (Pmax) and optimal velocity (Vopt). The purpose of this study was to investigate the validity and feasibility of loaded 5 Sit-to-Stand (5STS) force-velocity profiling compared to the gold standard instruments (isokinetic Dynamometry, ISO) and asses the relationship between the 5STS-derived muscle function indexes with clinical markers of muscle mass and strength. Forty-six older adults (21 females: 63.8 ± 3.9 yrs) performed 5STS (four different weight conditions) and ISO tests (five different velocities). Paired t-tests, regression analyses, and Bland–Altman analysis were conducted. The results showed significant differences in F0, V0, and Vopt ($p < 0.001$) but no difference in Pmax ($p = 0.259$) between tests. Only F0 and Pmax were highly correlated between tests ($r = 0.71$, $r = 0.84$, respectively). Bland–Altman analysis showed a not significant bias and good precision ($p = 0.260$, 34 W) only for Pmax. Large to very large correlations ($r = 0.53$ to $0.81$) were found between F0 and Pmax and clinical markers of muscle mass and strength. In conclusion, loaded 5STS profiling could be a feasible, valid, and cost- and time-efficient alternative to ISO for the characterization of clinically relevant markers of muscle function in healthy older adults.



## KEYWORDS

force–velocity; sit-to-stand; aging, strength test; field test; validation; clinical evaluation; muscle power.

## INTRODUCTION

Muscle function (i.e., strength and power) is positively correlated with overall health, independence, and quality of life in aging [1]. The rate of decay of muscle function parallels the progressive loss in muscle mass from age 65 (1.4–2.5% per year for strength and ~3.5% per year for power) leading to adverse health outcomes and reduced ability to carry out functional tasks of daily living (e.g., recovering balance, walking, sitting and standing from a chair) [2–8]. This, in turn, triggers a vicious cycle of reduced mobility, frailty, and loss of independence that further amplifies the deterioration of health [9]. To stop this cycle, early identification of individuals at high risk of loss of muscle mass (sarcopenia) and muscle function (dynapenia) allows a timely and effective intervention for those most in need. Low-cost/accessible assessment and periodic monitoring of muscle function in older adults are instrumental to this aim.

Force–velocity (F-v) and power–velocity (P-v) profiling are typically used to characterize muscle function. Compared to methods that rely on maximal effort (such as one repetition maximum (1RM) or the estimation of 1RM with the repetition-based method), this indirect approach is less affected by motivation, less physically demanding, and reduces the risk of acute injury. Moreover, these relationships allow a full characterization of muscle function: maximum force (F0), velocity (V0), power (Pmax), and optimal velocity (velocity eliciting maximum power, Votp) [5,8]. In turn, these indexes inform us of the prevailing limitation in a given individual (e.g., a loss of maximal or specific strength would suggest a loss in muscle mass or contractile quality; a prevalent loss of velocity would indicate that the prevalent problem resides in the efficacy of neuromuscular activation) [5], and provide the bases for individualized exercise programs [10,11]. These characteristics make F-v and P-v profiling particularly valuable approaches for assessing maximal strength and power in older adults [8].

Typically, the F-v/P-v relationship is detected either by assessing velocity during a movement executed at different loads (isotonic evaluation) or by collecting force data during a movement executed at different velocities (isokinetic evaluation) [1,10]. The gold standard method for the determination of the lower-limb F-v relationship consists of a knee-extension test performed on an isokinetic dynamometer (ISO) [1]. However, this method requires expensive equipment and qualified personnel and is poorly correlated with everyday life activities, as it only examines single-joint movements [12].

In the search for more cost-effective and ecological approaches, alternative multi-joint movements (e.g., leg press, Nottingham power rig) with increasing loads have been used for “field” F-v profiling in older adults [10,13]. Still, this requires the use of relatively expensive and/or non-portable machines. Furthermore, the guidance offered by the machine during the movement nullifies the expression of coordination and balance ability.

The 5 Sit-to-Stand (5STS) test is a simple, low-cost testing approach that is widely used in clinical settings to indirectly assess lower limb muscle power [14]. It evaluates a multi-joint, everyday life movement and its results are strongly associated with markers of physical function (i.e., handgrip strength, walking speed, short physical performance battery score) in older adults [15]. These promising features have suggested the use of a modified version of the 5STS test for the determination of the F-v profiling in a group of predominantly female older adults [12]. While the study did not find a correspondence between indexes derived from 5STS test F-v profiling and isokinetic F-v profiling, we speculate that the use of non-gold standard instruments and protocol for the 5STS (i.e., video camera analysis, number of trials, and choice of load condition) may have affected the correspondence between measures.

Therefore, the purpose of the present study was to replicate the above study including both males and females using different F-v protocols and equipment for 5STS evaluation. Then, the first aim was to investigate the feasibility and validity of the loaded 5STS test compared to the gold standard for the assessment of F0, V0, Pmax, and Vopt through F-v profiling in older adults. The second aim was to investigate the relationship between F0 and Pmax, as measured with both loaded 5STS and the gold standard approach, with clinical markers of muscle mass and strength.

## METHODS

*Participants*

Forty-six elderly volunteers (25 males and 21 females; means ± SD: age 66.1 ± 5.7 years; body mass 74.1 ± 16.5 kg; height 1.68 ± 0.09 m) were recruited. Participants included in the present study were aged ≥ 60 and were free of cardiopulmonary, metabolic, musculoskeletal, and neurological diseases. Prior to study participation, all participants underwent a medical screening, signed a written informed consent and were briefed on the experimental procedures. The study was approved by the Ethics Board committee of the University of Verona and conducted in conformity with the Declaration of Helsinki.

*Study procedures*

A validity study was conducted to compare the 5STS and the ISO knee-extension test. All participants attended the laboratory twice at the same time of the day for 1 hour, separated by at least 72 hours of recovery to allow the muscle fatigue and soreness to return to baseline levels [16]. Participants were asked to avoid any strenuous activities in the 24 hours before the first visit and between the two visits and to avoid any type of exercise on the mornings of the experimental visits. During the first visit, the following clinical markers of muscle mass and strength were collected: anthropometric measures (height and weight), lean mass of whole-body level through dual-energy X-ray absorptiometry (DXA), and handgrip strength. Moreover, an ISO knee-extension strength test was performed at five different velocities. During the second visit, participants performed the 5STS test at four different weight conditions.

*Data collection*

*2.3.1. Clinical markers of muscle mass and strength*

Body mass was measured to the nearest 0.1 kg using an electronic scale (Tanita electronic scale BWB-800 MA, Tokyo, Japan). Stature was measured to the nearest 0.005 m using a Harpenden stadiometer (Holtain Ltd., Crymych, Pembs, UK).

Total body composition was assessed using a DXA scan on a QDR Explorer fan-beam densitometer (Hologic Inc, Horizon C DXA System, Bedford, MA, USA). Quality control of the DXA scan was performed daily with an encapsulated spine phantom (Hologic Inc, PDA/QDR-1, Bedford, MA, USA) to check for possible baseline drifts. Prior to scanning, participants were asked to empty their bladder, wear underwear only, and remove any metallic objects and reflective materials. The total-body DXA scan lasted about 7 minutes and was carried out and analyzed by the same trained technician (to ensure consistency) in accordance with “The Best Practice Protocol for the assessment of whole-body composition by DXA” [17].

Isometric grip strength test of the dominant hand was conducted by using the Jamar hand dynamometer (Model 5030 J1, Sammons Preston Rolyan, Bolingbrook, IL, USA). The protocol was designed following the proposal of the American Society of Hand Therapists [18]. Participants were positioned in a sitting position, with their shoulder adducted and neutrally rotated, elbow flexed to 90 degrees, forearm mid-prone, and wrist between 15 and 30° of dorsiflexion and 0–15° of ulnar deviation. The instrument had been placed in the hand with the handle placed in the second position. The operator was positioned in front of the subject to set the peak needle to zero. All subjects performed 3 trials squeezing as hard as possible. Each trial consisted of at least 3 seconds of isometric contraction, with 30 seconds of recovery between trials [18]. The operator read the grip strength measure and recorded the result to the nearest 1 kg. Before testing, familiarization with 2 trials of submaximal effort was conducted.

*2.3.2. Isokinetic Strength test*

An isokinetic dynamometer (CMSi Cybex Humac Norm Dynamometer, Lumex, Ronkonkoma, NY, USA) was used to assess participants’ maximum strength in ISO knee-extension movement. Calibration and correction of the force of gravity were carried out following the standard procedures of the instrument. Setting up of the seat and mechanical arm was carried out in accordance with the participants' anthropometric characteristics to allow the alignment between the center of rotation of the knee and the fulcrum of the dynamometer. Participants were asked to sit on the isokinetic dynamometer seat, keeping the trunk straight and the thighs parallel to the ground in order to for the hip joint to be at 90°. Subsequently, the trunk, hips, and dominant thigh were fixed with straps to the seat; the dominant ankle was fixed with straps to the mechanical arm. Once the participant was positioned correctly, the lever arm was measured with a tape between the fulcrum of the dynamometer and the point of application of the force to be used for data analysis. Finally, to keep the test safe, electronic and mechanical locks based on individual maximal knee extension and flexion were set to limit the range of motion of the machine.

Participants performed a warm-up immediately before the ISO strength test consisting of 2 sets of 5 consecutive knee extensions at moderate angular velocity (2.09 rad × $s^{-1}$). The isokinetic strength test consisted of performing 3 maximal isokinetic contractions at 1.05, 1.57, 2.62, 3.14, and 3.67 rad × $s^{-1}$ [5]. Each set was separated by a 3-minute recovery.

The order of the sets was randomized and counterbalanced. An additional trial was performed immediately before each set to familiarize the participants with the different velocities.

*2.3.3. STS test*

A 3D MoCap system (Vicon, Oxford, UK) consisting of 8 Vicon cameras was used to collect lower limb kinematics from a marker placed on the greater trochanter during the test. Ground reaction forces (GRF) were measured using a force platform (AMTI Inc., Watertown, MA, USA) placed in front of a box. Participants were instructed to sit on the box (height: 0.49 m), maintaining the trunk and shank perpendicular to the ground, arms crossed to the chest, and feet placed on the force plate (Figure 1). At the “start” command of the operator, participants had to stand (concentric phase) and sit (eccentric phase) from the box five times consecutively as fast as possible. The trial was considered valid if i) the stance of the feet was unchanged throughout the test and ii) the trunk reached the vertical position at the end of all concentric and eccentric phases. Otherwise, the test was repeated. To check for these requirements and to ensure safety, an operator was positioned close to the participant. Before commencing the test, all participants performed a 5-minute warm-up on a cycle ergometer (Monark 814 E, Monark, Vargerb SE) at 50 W (60 rpm) and four lower-limb active mobility exercises [19]. Following a familiarization session, participants performed two trials of 5STS tests under four different conditions: body weight (BW), + 12.5% BW, + 25% BW, and + 32.5% BW. The additional weight consisted of a weighted vest (Weight Vest bv30, Lacertosus, Parma, IT) worn immediately before the trial and secured to the participants’ abdomen with a belt strap. The order of the trials at the different weight conditions was randomized and counterbalanced. Participants were instructed to perform a single sit-to-stand movement before each testing condition in order to familiarize the participants with the different weights. To avoid fatigue accumulation, each condition was repeated twice, with a 3-minute recovery between trials and a 5-minute recovery between conditions.

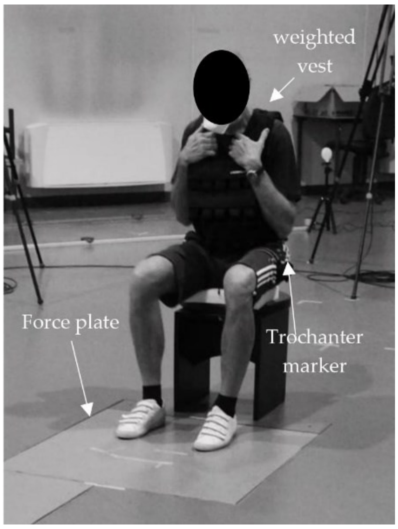


**Figure 1.** The picture illustrates the initial position of the participants during the test. The participant was seated on the box with the trunk and shank positioned perpendicular to the ground. The force plate, weighted vest, and marker on the trochanter have been highlighted in the figure.

*2.3.4. Adverse events*

Adverse events were closely monitored during the data collection and the following week. It was explained to the subjects that after the maximum strength assessments, they felt tired and could feel muscle pain associated with physical activity. An adverse event was defined as any episode evoking pain, discomfort, injury, or accident that occurred during the study. In case an adverse event occurred, its origin and etiology would be identified to classify it as study-related or not.

*Data analysis*

*2.4.1. Clinical markers of muscle mass and strength*

Participants' body mass index (BMI) was calculated as body mass × height$^{-2}$ (kg × m$^{-2}$) (see Table 1 for participants' characteristics).

The DXA scans were analyzed using Hologic Discovery version 12.6.1 (Holtain Ltd, UK). The technician localized the specific anatomical landmarks directly from the scans, to differentiate the standard regions of interest (arms (right and left), legs (right and left), and the trunk). The body composition variables of interest included whole-body lean mass (WBLM), dominant lean leg mass (d-LLM), and non-dominant lean leg mass (nd-LLM). To calculate lean leg mass (LLM), the lean mass of lower limbs was summed. Subsequently, LLM was divided by height squared to find the leg's skeletal muscle index (leg's SMI) [17]. Finally, the dominant lean leg mass was used to normalize the force and power data.

**Table 1.** Anthropometrics, body composition, and clinical markers of muscle mass and strength of the sample.

| | **Female** | **Male** | **Total** | ***p-value*** |
|---|---|---|---|---|
| # | 21 | 25 | 46 | |
| **Age (yrs)** | 63.8 ± 3.9 | 68.1 ± 6.3 | 66.1 ± 5.7 | 0.009* |
| **Weight (kg)** | 65.7 ± 12.1 | 81.2 ± 16.6 | 74.1 ± 16.5 | 0.001* |
| **Height (m)** | 1.61 ± 0.05 | 1.74 ± 0.06 | 1.68 ± 0.09 | <0.001* |
| **BMI (kg × m$^{-2}$)** | 25.3 ± 4.1 | 27.0 ± 5.9 | 26.3 ± 5.2 | 0.286 |
| **WBLM (kg)** | 39.4 ± 5.0 | 55.1 ± 8.9 | 48.4 ± 10.8 | <0.001* |
| **d-LLM (kg)** | 4.6 ± 0.9 | 6.3 ± 1.1 | 5.6 ± 1.3 | <0.001* |
| **nd-LLM (kg)** | 4.5 ± 0.6 | 6.2 ± 1.2 | 5.5 ± 1.3 | <0.001* |
| **LLM (kg)** | 9.1 ± 1.5 | 12.5 ± 2.3 | 11.0 ± 2.6 | <0.001* |
| **SMI (kg × m$^{-2}$)** | 4.9 ± 0.6 | 5.9 ± 1.0 | 5.4 ± 1.0 | <0.001* |
| **Handgrip strength (kg)** | 27.3 ± 5.6 | 41.2 ± 6.9 | 34.8 ± 9.4 | <0.001* |

Handgrip strength (kg) was calculated by averaging the measures of the three trials [18]. Mean ± SD values of anagraphic, anthropometric, and strength variables, in females, males, and in the total group. BMI, body mass index; WBLM, whole-body lean mass; d-LLM, dominant lean leg Mass; nd-LLM, non-dominant lean leg mass; LLM, lean leg mass; SMI, leg skeletal muscle index. The p-value of the unpaired t-test is displayed on the right side. * indicates a significant difference with females (F).

*2.4.2. Isokinetic Strength test*

Angular velocity (rad × s$^{-1}$) and torque (N × m) were obtained from the isokinetic knee-extension strength test (ISO) (Figure 2a). The sampling rate was set at 1000 Hz. The torque signal was filtered using a second-order low-pass Butterworth filter with a cut-off frequency of 15 Hz. Zero-crossings of the angular velocity signal were used to identify the beginning and the end of each extension phase (Figure 2a). The mean torque for each repetition was subsequently calculated and the three values obtained were averaged. In order to assess the F-v relationship, torque (N × m) and angular velocity (rad × s$^{-1}$) were converted into force (N) and linear velocity (m × $s^{-1}$). To do this, torque was divided by the length of individual lever arms and angular velocity was converted in rad × $s^{-1}$ and multiplied by the length of individual lever arms [20]. Since the force–velocity relationship of single-joint tasks is considered to be approximately linear [20], the individual F-v relationship was assessed by fitting a linear regression through the force and velocity data obtained from the five angular velocities tested (linear regression method) for each subject (Figure 2b). Maximum force (F0; force-intercept), maximum velocity (V0; velocity-intercept), and slope of the relationship (a = F0/V0) were detected. Finally, power and velocity values were fitted with a parabolic function (i.e., $y= ax^2+bx+c$) for the computation of maximum power as the apex of the parabola (i.e., $Pmax= -(b^2 - 4ac) \times 4a^{-1}$) and the corresponding optimal velocity ($Vopt= -b \times 2a^{-1}$). To fix the power–velocity relationship, the points of intersection of the parabola with the x-axis were added (at null velocity and V0, the power corresponds to zero) [21].

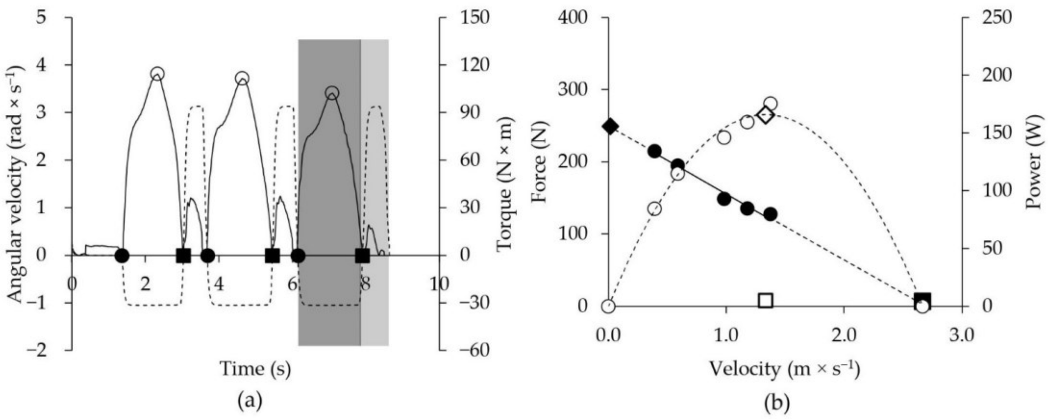


**Figure 2.** a) Graph of the isokinetic knee-extension strength test at 1.05 rad × $s^{-1}$ in a representative subject. Torque (N × m; solid line ——— ), and angular velocity (rad × $s^{-1}$; dotted line ····) are plotted in function on time (s). Knee extension (light grey area) and flexion (dark grey area) phases were identified when angular velocity was negative and positive, respectively. The start (●) and the end (■) of knee extension repetitions were reported where angular velocity crosses zero. Peaks of torque (○) were found for each knee extension. In b) force–velocity (● F-v) and power–velocity (○ P-v) relationship of the 5 Sit-to-Stand test in a representative subject are reported. Maximum force (♦ F0) corresponds to the intercept with the y-axis where velocity is null; maximum velocity (■ V0) corresponds to the intercept with the x-axis where force is null; maximum power (◊ Pmax) represents the apex of the power–velocity curve; optimal velocity (□ Vopt) is the velocity at maximum power.

*2.4.3. STS test*

Vertical force, position, and velocity were extrapolated from kinetic and kinematics data of the 5STS test in each of the four different weight conditions (Figure 3a). The sampling rate was set at 1000 Hz and 100 Hz, and a second-order low-pass Butterworth filter was used for both signals (frequency cut = 7 Hz and 20 Hz, respectively). Vertical velocity was used to recognize the repetitions and each concentric and eccentric phase. Then, positive and negative peaks of vertical velocity were identified, and thresholds were calculated as 5% of the peaks. The start and end of repetition were found when the vertical velocity reached the positive and negative threshold, respectively. Finally, the end of the concentric phase was defined when vertical velocity crossed zero after a positive peak (Figure 3a) [22]. For our purpose, only the concentric phases (i.e., the standing phases) were considered to calculate mean concentric vertical velocity and mean concentric vertical force [10], and the five values obtained from repetitions were averaged [23]. As a linear equation was expected from the F-v relationship of multi-joint movement [21], when a mean concentric vertical velocity of a repetition differed by more than 0.03 m × $s^{-1}$ from the estimated value (based on linear regression), the repetition was removed. If all five repetitions differed by more than 0.03 m × $s^{-1}$, the trial was deleted [10].

Then, it was possible to fit the individual force–velocity and power–velocity relationships for each subject (Figure 3b). All parameters of interest (F0, V0, a, Pmax, and Vopt) were identified with the same method used for isokinetic strength measurements (see above). Finally, in order to facilitate the direct comparison between ISO and 5STS tests, single-leg STS force and power were estimated by adding 23% and 20% of the total, respectively (to consider the bilateral deficit [8]), and then dividing by 2.

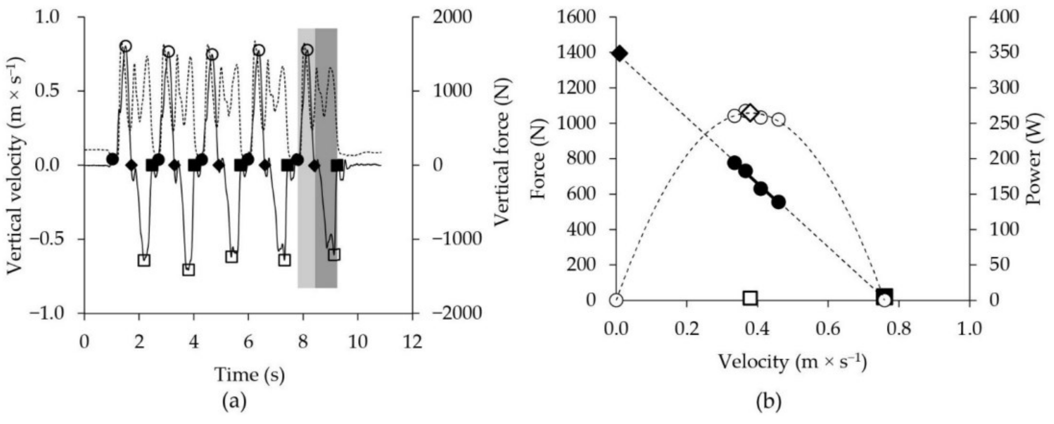


**Figure 3** a) Graph of the 5 Sit-to-Stand test in a representative subject is shown. Vertical velocity (m × s⁻1; solid line ——— ) and vertical force (N; dotted line ····) are plotted in function on time (s). The 5STS involves 5 repetitions, each of which is divided into concentric (light grey section) and eccentric phases (dark grey section). The start (●) and the end (■) of a full repetition (concentric-eccentric phase) are identified as the points at which the vertical velocity reaches 5% of positive (○) and negative (□) peak vertical velocity, respectively. The end (♦) of the concentric phase is identified as the time when the

vertical velocity crosses zero. In b) force–velocity (● F-v) and power–velocity (○ P-v) relationship of the 5 Sit-to-Stand test in a representative subject are reported. Maximum force (♦ F0) corresponds to the intercept with the y-axis where velocity is null; maximum velocity (■ V0) corresponds to intercept with the x-axis where force is null; maximum power (◊ Pmax) represents the apex of the power–velocity curve; optimal velocity (□ Vopt) is the velocity at maximum power.

All signal analyses of ISO and 5STS tests were performed by MatLab (Version R2021B, MathWorks Inc, Natick, Massachusetts, USA) scripts, and the results were exported to an Excel (Microsoft 365, Version 16.0.16501.20228, Microsoft Corporation, Washington, USA) spreadsheet with the anthropometric and body composition measures for subsequent calculations.

*Statistical analysis*

The mean and standard deviation were calculated for all the variables. Unpaired samples t-tests were run to determine the differences in age, anthropometrics, and clinical markers of muscle mass and strength between sexes.

An unpaired samples t-test was run to determine the differences in velocity, force, and power measured during 5STS and ISO evaluations in different weight and velocity conditions, respectively, between males and females. Moreover, for each individual, we calculated the coefficient of determination ($R^2$) of the force–velocity relationship for both the 5STS and ISO evaluation. The mean coefficient of determination ($R^2_{mean}$) of the subjects was taken as an index of feasibility for both tests and compared by two-way repeated measures analysis of variance (ANOVA) between tests (ISO vs. 5STS) and sexes (males vs. females).

Maximum force, velocity, power, and optimal velocity measured with 5STS and ISO evaluation were compared by paired samples t-test. Moreover, Pearson's correlation coefficient was used to interpret the strength of the relationship between ISO and 5STS test in maximum force, velocity, power, and optimal velocity. Finally, Bland–Altman analysis [24] was used to determine potential systematic bias, precision, and limits of agreement (LOA) between ISO and 5STS measures. Bland–Altman analysis was followed by a one-sided z-test on the bias to test its difference from zero.

Finally, a correlation between maximum strength and power and clinical markers of muscle mass and strength (WBLM, d-LLM, LLM, SMI, and handgrip strength) was calculated. The interpretation of correlation coefficient (r) was conducted according to the following values: trivial (<0.1); small (0.10–0.29); moderate (0.30–0.49); large (0.50–0.69); very large (0.70–0.89); extremely large (0.90–1.00) [25].

A significance level was set at $p < 0.05$. SigmaPlot 12.5 (SigmaStat, USA) was used for all the statistical analyses.

## RESULTS

Age, weight, and height were significantly higher in males (M) than in females (F) ($p < 0.05$), while body mass index (BMI) was not statistically different ($p = 0.286$). All lean muscle mass indexes and handgrip strength showed a significant difference between sexes (M>F, $p < 0.001$) (Table 1).

No adverse events were recorded during the study, either during the tests or in the hours or days following them.

Velocity, force, and power measured during 5STS and ISO evaluations in different weight and velocity conditions in males and females and in the total group is reported in Table 2.

$R^2_{mean}$ of the F-v relationship in ISO (0.97 ± 0.03; Table 2) and 5STS (0.97 ± 0.03; Table 3) showed no effect for sex ($p = 0.875$), method ($p = 0.581$) and their interaction ($p = 0.674$).

**Table 2.** Variables obtained in isokinetic strength test.

| Velocity condition (rad × $s^{-1}$) | Female | | Male | | Tot | | p-value | |
|---|---|---|---|---|---|---|---|---|
| | MF (N) | MP (W) | MF (N) | MP (W) | MF (N) | MP (W) | MF | MP |
| **1.05** | 222 ± 32 | 74 ± 12 | 310 ± 66 | 113 ± 21 | 270 ± 69 | 95 ± 26 | < 0.001* | < 0.001* |
| **1.57** | 201 ± 29 | 101 ± 15 | 272 ± 57 | 149 ± 27 | 240 ± 58 | 127 ± 33 | < 0.001* | < 0.001* |
| **2.62** | 160 ± 24 | 134 ± 21 | 225 ± 48 | 205 ± 39 | 196 ± 51 | 173 ± 48 | < 0.001* | < 0.001* |
| **3.14** | 147 ± 25 | 147 ± 25 | 200 ± 45 | 218 ± 43 | 176 ± 46 | 186 ± 51 | < 0.001* | < 0.001* |
| **3.67** | 135 ± 20 | 157 ± 23 | 182 ± 42 | 232 ± 47 | 160 ± 41 | 198 ± 53 | < 0.001* | < 0.001* |
| **F-v relationship** | | | | | | | | |
| **a** | 255 ± 37 | - | 354 ± 74 | - | 309 ± 78 | - | - | - |
| **b** | −107 ± 21 | - | −139 ± 36 | - | −124 ± 34 | - | - | - |
| **$R^2$** | 0.97 ± 0.04 | - | 0.98 ± 0.02 | - | 0.97 ± 0.03 | - | - | - |

Mean ± SD values of force and power variables during five different velocity conditions of isokinetic knee extension, in females, males, and in the total group. MF, mean force; MP, mean power; int., interaction; F-v, force–velocity relationship (y=ax+b); a, slope; b, y-intercept; R2, coefficient of determination. The p-values of the unpaired t-test are displayed on the right side. * indicates a significant difference with females (F).

**Table 3.** Variables obtained in the 5 Sit-to-Stand test.

| Weight condition | Female | | | Male | | | Tot | | | p-value | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | McV (m × s$^{-1}$) | McF (N) | MP (W) | McV (m × s$^{-1}$) | McF (N) | MP (W) | McV (m × s$^{-1}$) | McF (N) | MP (W) | McV | McF | MP |
| **BW** | 0.46 ± 0.05 | 517 ± 171 | 241 ± 70 | 0.55 ± 0.08 | 658 ± 135 | 356 ± 66 | 0.51 ± 0.08 | 595 ± 150 | 305 ± 89 | < 0.001* | < 0.001* | < 0.001* |
| **BW + 12.5%** | 0.42 ± 0.13 | 589 ± 188 | 247 ± 100 | 0.51 ± 0.13 | 745 ± 203 | 375 ± 110 | 0.47 ± 0.07 | 674 ± 159 | 317 ± 96 | 0.012* | 0.011* | < 0.001* |
| **BW + 25%** | 0.38 ± 0.09 | 633 ± 209 | 244 ± 78 | 0.46 ± 0.12 | 825 ± 158 | 377 ± 106 | 0.43 ± 0.07 | 739 ± 185 | 318 ± 95 | 0.014* | < 0.001* | < 0.001* |
| **BW + 37.5%** | 0.34 ± 0.04 | 713 ± 231 | 239 ± 47 | 0.42 ± 0.07 | 904 ± 174 | 378 ± 83 | 0.39 ± 0.07 | 819 ± 197 | 316 ± 98 | < 0.001* | < 0.001* | < 0.001* |
| **F-v relationship** | | | | | | | | | | **Main Effect** | | |
| | | | | | | | | | | **sex** | **test** | **int.** |
| **a** | - | 1277 ± 302 | - | - | 1859 ± 602 | - | - | 1593 ± 565 | - | - | - | - |
| **b** | - | −1669 ± 663 | - | - | −2263 ± 1143 | - | - | −1992 ± 991 | - | - | - | - |
| **R$^2$** | - | 0.97 ± 0.02 | - | - | 0.97 ± 0.03 | - | - | 0.97 ± 0.03 | - | 0.875 | 0.581 | 0.674 |

Mean ± SD values of velocity, force and power variables during four different weight conditions of 5STS, in females, males, and in the total group. BW, body weight; McV, mean concentric velocity; McF, mean concentric force; MP, mean power; int., interaction; F-v, force–velocity relationship (y=ax+b); a, slope; b, y-intercept; R2, coefficient of determination. The p-values of unpaired t-test (males vs. females) and 2-way ANOVA (sex x test) are displayed on the right side. * indicates a significant difference with females (F).

Comparison of means (t-test) of maximum force (F0), velocity (V0) and optimal velocity showed a significant difference between tests (ISO vs. 5STS). On the contrary, maximum power was not different between tests (Figure 4).

The correlation between parameters measured from ISO and 5STS test was significant and very large for both maximum force ($p < 0.001$, $r = 0.71$) and power ($p < 0.001$, $r = 0.84$). On the contrary, a not significant and small correlation was found between tests for maximum velocity and optimal velocity ($p > 0.05$, $r = 0.23$) (Figure 4).

Bland–Altman analysis showed a significant bias between ISO and 5STS measures of maximum force, velocity, and optimal velocity (bias = 650 N, $-1.60\ m \times s^{-1}$ and $-0.83\ m \times s^{-1}$, respectively; ($p < 0.001$)). On the contrary, Bland–Altman analysis confirmed a not significant bias (bias = 5.7 W ($p = 0.259$)) between measures of maximum power performed with the two tests (Figure 4).

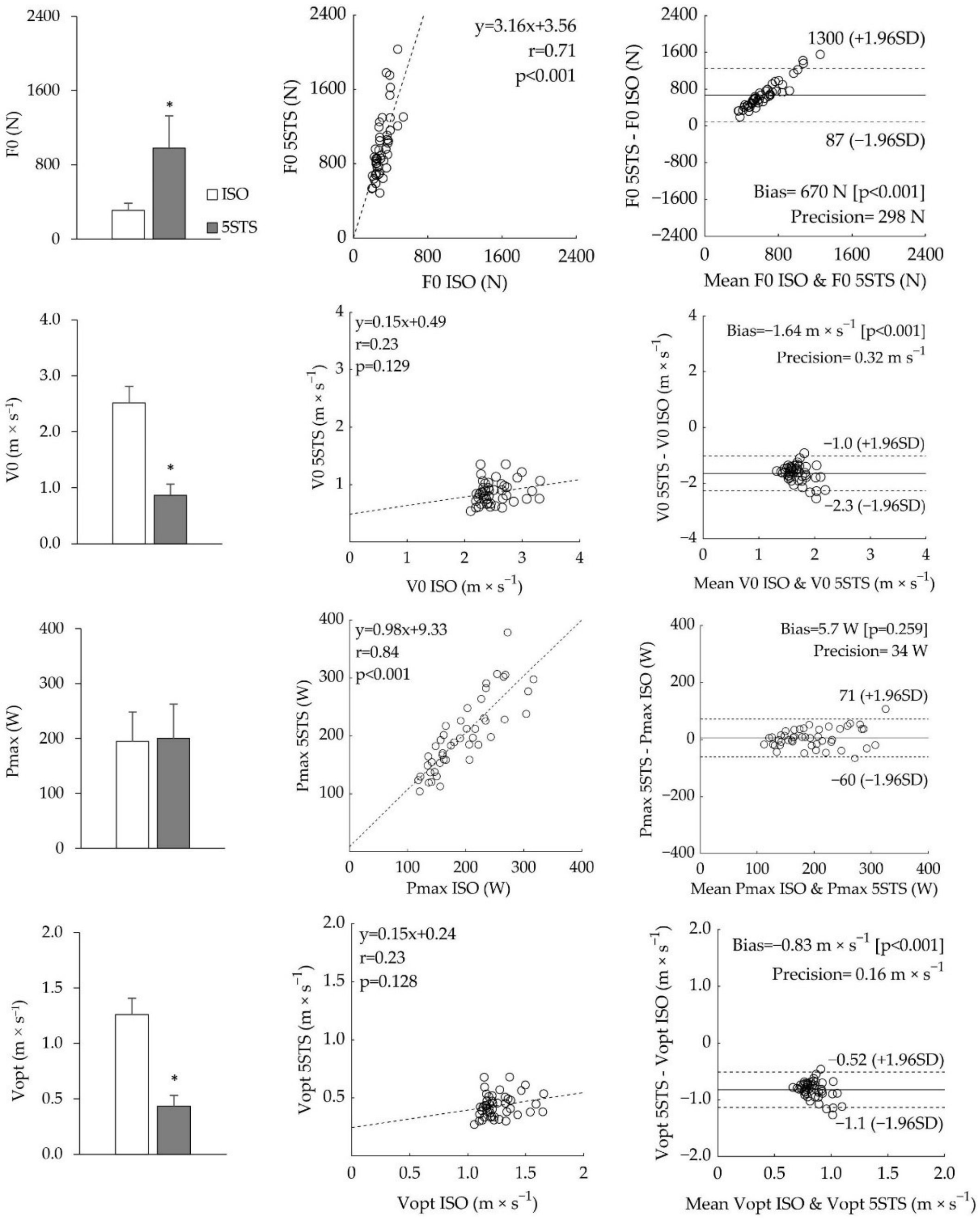


**Figure 4.** Comparison of means (left side), correlation graph (in the middle), and Bland–Altman analysis (right side) of maximal strength (F0), velocity (V0), Power (Pmax), and optimal velocity (Vopt), referred to the isokinetic strength test (ISO) and the 5 Sit-to-Stand test (5STS) are reported. In the bar graphs, * indicates a significant difference ($p < 0.05$) from ISO. In the correlation plots, the equation, Pearson's correlation coefficient (r), and p-value are reported along with regression lines (dashed line). In Bland–Altman plots, bias (solid lines), p-value, and precision are reported along with the limits of agreement (dashed lines).

A large to very large correlation was found between maximum force and power measured from ISO and 5STS tests and all clinical markers of muscle mass and strength (WBLM, d-LLM, LLM, SMI, and handgrip strength) (Table 4).

**Table 4.** Correlation coefficients between strength and power indexes and clinical markers of muscle mass and strength.

| | F0 (N) | | Pmax (W) | | Handgrip strength (kg) |
|---|---|---|---|---|---|
| | ISO | 5 STS | ISO | 5STS | |
| **WBLM (Kg)** | 0.75 | 0.73 | 0.84 | 0.81 | 0.74 |
| **d-LLM (Kg)** | 0.69 | 0.70 | 0.81 | 0.78 | 0.68 |
| **LLM (Kg)** | 0.71 | 0.71 | 0.82 | 0.79 | 0.68 |
| **SMI (kg × m$^{-2}$)** | 0.66 | 0.69 | 0.74 | 0.65 | 0.50 |
| **Handgrip strength (kg)** | 0.70 | 0.53 | 0.74 | 0.75 | 1.00 |

Pearson's correlation coefficients between the indicated variables as measured in the isokinetic strength test (ISO) and the 5 Sit-to-Stand test (5STS) are reported: F0, maximum force; Pmax, maximum power; WBLM, whole-body lean mass; d-LLM, dominant lean leg mass; LLM, lean legs mass; SMI, leg skeletal muscle index.

## DISCUSSION

The first aim of the present study was to investigate the feasibility and validity of the 5STS test compared to the ISO test to assess maximum force, velocity, and power through F-v and P-v profiling in older adults. The second aim was to investigate the relationship between maximum force and power with markers of muscle mass and strength that are typically used in a clinical setting. The main findings of the present study are as follows: i) 5STS profiling is a feasible and valid alternative to isokinetic testing for the characterization of muscle function in healthy older adults of both sexes; ii) while the absolute values of maximum force and maximal and optimal velocity are significantly different between the two tests, the maximum power values measured in 5STS and ISO are similar and highly correlated; iii) both maximal muscle strength and power are significantly and highly correlated with the most commonly used clinical markers of muscle mass and strength.

In our study, the feasibility of the 5STS test in terms of safety performing the protocol and building a proper F-v and P-v profile was verified. In fact, all participants completed all the trials for both ISO and 5STS tests without experiencing any adverse outcomes. Furthermore, the $R^2_{mean}$ of F-v profiling in 5STS (0.97 ± 0.03) was high and similar ($p = 0.581$) to the ISO (0.97 ± 0.03). Moreover, the values for 5STS are similar to the $R^2_{mean}$ presented in the literature for other lower limb multi-joint exercises performed by older adults (e.g., leg press, $R^2_{mean}$ from 0.95 to 1.00) of both sexes [10]. These results suggest that it is possible to use the 5STS test not only as a generic screening tool but, in its modified version which includes the use of overloads, for the full characterization of muscle function.

The values of F0 from ISO are typically presented in the literature in peak torque units [5,26,27]. However, as our study focused on the comparison of F0 between ISO and 5STS tests, we used mean rather than peak data and converted torque and angular velocity from ISO into force and linear velocity [20,28]. For these reasons, direct comparison with the literature data may be difficult. To the best of our knowledge, only Grbic et al. [20] presented F0 values for the isokinetic test in Newton units derived from both peak and mean values. The above study conducted in young females found F0 values (~350 N) similar to those observed in the present study (309 ± 78N). When our F0 values are converted into torque units (by multiplying force by the mean lever arm) the values found in our study (309 N × 0.34 m= 105 N × m) are lower than the literature (~155 N × m) in a comparable population [5,26,27]. This may be due to the fact that we used mean rather than peak values for F-v profiling.

To the best of our knowledge, only Piche et al. investigated F0 derived from STS movement [12]. The values for that study are much lower (62 ± 42 N vs. 980 ± 348 N) than our values. This difference could lie in the fact that Piche et al. calculated by methodological differences using different instruments (gravitational force of body weight vs. force plate) and populations (predominantly female older adults vs. both sexes). However, when considering other knee–hip extension movements (i.e., leg press, Nottingham power rig), our results are similar to those found in the literature (females: 751 N vs. 785 N, males: 1859 vs. 2145 N) [8,29].

The F0 between the 5STS and ISO tests showed a significant difference ($p < 0.001$) and bias (bias= 670 N, $p< 0.001$) with a high limit of agreement (L.O.A.= [87 to 1300 N]) and yet a high correlation between measures ($r = 0.71$). We speculate that the difference between the expression of lower limbs' maximum muscle strength between tests is due to the difference in muscle action (in terms of neuromuscular and biomechanical characteristics, e.g., muscle coordination of single vs. multi-joint movements and different contraction lengths) [8] required by the two movements (single-joint vs. multi-joint), and difference in the applied load (weight-bearing versus non-weight-bearing) between tests [13]. These discrepancies could have affected the slope of the force–velocity relationships and consequently the computation of F0. Moreover, our results differ from the only previous study that compared F0 values from these tests [8,29] and found no difference and a low correlation between tests. Again, this discrepancy could be related to methodological differences cited in the previous paragraph (i.e., methodological differences, indirect force estimation based on body mass) as well as to the high variability of measures in Piche et al. [12]. In summary, the measures of F0 as derived from 5STS, while highly correlated with ISO, are not an accurate and precise surrogate of the maximum isokinetic strength of the unilateral lower limb.

As previously mentioned regarding F0 measures, ISO maximum power (Pmax) data are typically computed based on peak rather than mean force data [5,26,27]. Therefore, it is not surprising that our Pmax values are lower than those reported in the literature in a comparable population [5,26,27]. Interestingly, the results derived from our ISO test and calculated from mean force (194 ± 54 W) (Figure 4) were similar to those computed using mean force values in young females [20].

The bilateral STS Pmax found in this study (333 ± 104 W) was closer to that found by other authors in bipedal multi-joint exercises (e.g., leg press) (~ 350 W) [2,11] in a comparable population. In comparison with the only other study that computed unilateral STS Pmax in young women (187 ± 147 W) [12], we found similar values with a considerably lower variability (200 ± 63 W).

The comparison of Pmax between the ISO and 5STS tests showed a non-significant difference and a high correlation ($p = 0.259$, $r = 0.84$). Bland–Altman analysis reported a non-significant and constant bias (bias = 5.7 W, $p = 0.259$) with small limits of agreement (L.O.A. = [−60 to 71 W]). Our correlation results between 5STS- and ISO-derived Pmax are in contrast with those found in the literature ($r = 0.31$) [12]. This discrepancy may lie in the factors discussed above for F0 and in the high variability (CV > 75%) of the 5STS-derived maximal power [12]. In summary, the F-v profiling of 5STS appears an accurate and precise alternative to ISO for the measurement of the maximum power of the lower limbs.

Maximum (V0, 7.5 ± 0.9 rad × $s^{-1}$) and optimal (Vopt, 3.8 ± 0.4 rad × $s^{-1}$) angular velocities for the isokinetic test are considered, and these values are similar to Piche et al. (7.1 ± 2.1 and 3.6 ± 0.8 rad × $s^{-1}$, respectively) who, like us, used a linear F-v profiling. Other authors, who used a hyperbolic or hybrid fitting of the F-v profile, found higher values (~ 10 and ~ 4.5 rad × $s^{-1}$, respectively) than ours [5,26]. This difference is likely due to the fitting model. Even though hyperbolic fitting is likely more appropriate for single-joint movement, we decided to use a homogeneous linear fitting for both tests and chose the linear model that is preferable for the multi-joint sit-to-stand action [21]. This is clearly an arbitrary decision. Interestingly, if we use a hyperbolic fitting to build the F-v relationship in ISO tests, our data align well with the latter authors (V0 ~10 and Vopt ~4.25 rad × $s^{-1}$).

Regarding V0 (0.9 ± 0.2 m × $s^{-1}$) and Vopt (0.4 ± 0.1 m × $s^{-1}$) results derived from the 5STS test, we found lower values from those found in the literature (6 ± 7 rad × $s^{-1}$ that approximatively correspond to 1.9 ± 2.2 m × $s^{-1}$)[12]. However, the V0 values are similar to those found for a more comparable leg press exercise in older adults (~0.9 m × $s^{-1}$) [10].

In order to compare movement velocity in the two tests, the angular velocity of the ISO test was transformed into linear velocity. Maximum and optimal velocity results were different and poorly correlated ($r = 0.23$, $p < 0.001$) between ISO and 5STS. Bland–Altman analysis reported a significant bias (V0, bias = −1.6 m × $s^{-1}$, $p < 0.001$; Vopt, bias = −0.8 m × $s^{-1}$, $p < 0.001$) with high limits of agreement (V0, L.O.A. = [−2.3 to −1.0 m × $s^{-1}$]; Vopt, L.O.A. = [−1.1 to −0.5 m × $s^{-1}$]). Our results confirm that velocity parameters extrapolated from the F-v relationship seem to have less concurrent validity and precision than other indexes [21]. Therefore, these parameters could have less relevance in medical screening and clinical assessment.

The second aim of this study was to investigate the relationship between maximum force and power with markers of muscle mass and strength. In our study, F0 and Pmax were highly correlated with all body and specific lean muscle mass for both tests (r from 0.65 to 0.82). These results are comparable to those found by Takai et al. [7] between STS power and leg muscle mass ($r = 0.80$).

It is well known that grip strength is a strong predictor of mortality, disability, complications, and length of stay [30]. Furthermore, this index represents the main reference value of the general strength for the diagnosis of sarcopenia [31]. The force values for the handgrip test (27.3 ± 5.6 kg for females; 41.2 ± 6.9 kg for males; 34.8 ± 9.4 kg tot) are consistent with those presented in the literature for healthy older adults of both genders (~26 kg for females; ~40 kg for males) [32,33]. Furthermore, Pmax was highly correlated with handgrip strength for both tests ($r = 0.75$), as in Glenn et al. ($r = 0.67$) [34].

Recently, it has been asserted that handgrip strength alone would be insufficient as a measure of overall muscle strength in clinical practice [35]. In fact, muscle power would appear to be a better indicator of loss of muscle function than muscle strength alone [36]. In addition, Winger et al. found that lower limb muscle power (from a jump test) was approximately 2-fold more correlated with all physical performance tests than handgrip strength [36,37]. Furthermore, a better association was found in this study between lower limb strength/power and appendicular lean mass indices with respect to handgrip strength.

This suggests that lower-limb strength tests could better reflect both the condition of physical function and appendicular lean mass in older adults.

The present study has some limitations. Due to logistic limitations, it was not possible to divide the subjects into two groups to randomize the order of the tests. We used loads computed as percentages of body weight to profile the F-v relationship independently of the percentages of body fat. This could lead to a different ratio between overload and the percentage of lean body mass. Usually, characterizing the F-v relationship requires two to four weight conditions that lead to differences of at least 0.5 m × $s^{-1}$ between the lightest and heaviest weight [38]. To use a pragmatic and easy approach, we used four loading conditions based on body weight that did not respect this velocity loss criterion. However, we were able to characterize the P-v relationship well because the weight conditions chosen were around the Pmax. Therefore, future studies may consider modifying the weight condition to better describe F0. Another possible limitation was that we used the same chair height for all subjects. In fact, different heights involve different vertical displacements and therefore different mechanical effort. Future research on 5STS F-v profiling could standardize the knee angle in the sitting position. Finally, although widely used in the clinical setting, there are more accurate instruments to assess muscle tissue than DXA. Other assessment methods may better describe the subjects' body composition (i.e., magnetic resonance imaging or computed tomography scan).

Future studies should investigate the feasibility and validity of the loaded 5STS performed using inexpensive and portable tools (i.e., 3D inertial sensor, linear transducer, phone app) to make the evaluation easier and more accessible. Furthermore, it would be useful to investigate a shorter protocol (i.e., two weight conditions, BW and W3) for a less time-consuming assessment.

## CONCLUSIONS

In conclusion, muscle profiling based on loaded 5STS test is a feasible, valid, and cost- and time-efficient alternative to isokinetic testing of the characterization of muscle power in healthy older adults of both sexes. In addition, maximal force and power derived from the F-v profile are significantly and highly correlated with the major clinical markers of muscle mass and strength. It is well-known that the decline of these variables is associated with adverse outcomes in aging (frailty, impaired physical function, and disability in daily living activities). Therefore, muscle profiling could be used as a monitoring tool for the early detection of individuals at higher risk of unhealthy aging and provide a valuable tool for the individualization of training interventions [3].

# CHAPTER

# 4

## Construct Validity of a Wearable Inertial Measurement Unit (IMU) in measuring Postural Sway and the effect of visual deprivation in healthy older adults.

**Construct Validity of a Wearable Inertial Measurement Unit (IMU) in measuring Postural Sway and the effect of visual deprivation in healthy older adults.**

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ABSTRACT

Abstract: Inertial Motor sensors (IMUs) are valid instruments for measuring postural sway but their ability to detect changes derived from visual deprivation in healthy older adults requires further inves-tigations. We examined the validity and relationship of IMU sensor-derived postural sway measures compared to force plates for different eye conditions in healthy older adults (32 females, 33 males). We compared the relationship of the center of mass and center of pressure (CoM and CoP)-derived total length, root means square (RMS) distance, mean velocity, and 95% confidence interval ellipse area (95% CI ellipse area). In addition, we examined the relationship of the IMU sensor in discriminating between open- (EO) and closed-eye (EC) conditions compared to the force plate. A significant effect of the instruments and eye conditions was found for almost all the variables. Overall, EO and EC variables within (force plate r, from 0.38 to 0.78; IMU sensor r, from 0.36 to 0.69) as well as between (r from 0.50 to 0.88) instruments were moderately to strongly correlated. The EC:EO ratios of RMS distance and 95% CI ellipse area were not different between instruments, while there were significant differences between total length ($p = 0.973$) and mean velocity ($p = 0.703$). The ratios' correlation coefficients between instruments ranged from moderate ($r = 0.65$) to strong ($r = 0.87$). The IMU sensor offers an affordable, valid alternative to a force plate for objective, postural sway assessment.

KEYWORDS

Aging, Postural control, Balance, Inertial sensors, Risk of falls.

## INTRODUCTION

Balance is a multidimensional concept referring to the ability of a person not to fall [1]. By the periodical monitoring of the age-related decline in balance, individuals with an increased risk of falls can be identified in a timely manner [2,3]. The early detection of the maintenance of balance is possible through the measure of postural control. The control of balance is associated with three different human activities: reaction to external disturbance (restoration), voluntary movement (achievement), and maintenance of a posture (maintenance) [1]. The latter is measurable through static sway, defined as the ability to maintain the center of pressure (CoP) within the limits of the base of support while standing still [4]. Static sway is one of the most studied balance components since it can be easily and safely estimated using clinical tests such as the Short Physical Performance battery test or the Berg balance scale [5,6]. Typically, these clinical tests are based on the visual assessment of the patient by using a scalar score [5,6]. Despite being flexible and easy to use, clinical tests are useful only for visible and gross balance deficits, excluding them as a tool for the early identification and monitoring of an increased risk of falls and/or the detection of subtle balance deficits [2].

To overcome these limitations of field tests, balance can be quantified objectively through posturography using optoelectronic systems or force plates [2,7]. Different useful variables can be extrapolated from the displacement of the CoP measured using the gold standard force plate method [4]. However, the well-known issues linked with the required equipment (high cost, needed for specialized personnel, not very trans-portable) limit the use of the gold standard equipment in laboratory settings only.

Therefore, alternative approaches have been sought to provide affordable yet objective measures that are also sensitive to change. In this context, instrumented clinical tests using inertial measurement units (IMUs) can provide a large amount of quantitative information about the individual’s sway. Typically, the IMU sensor is worn near the center of mass (CoM) at the lumbar spine level (on the L5 vertebra) and measures its tridimensional acceleration, velocity, and displacement [7,8]. Although these values could not be directly compared with the force plate-derived CoP variables [8,9], CoM and CoP are strictly related. Indeed, if the body moves like an inverted pendulum [10,11], a correlation close to 1 is expected between trunk acceleration and CoP displacement [9]. Therefore, a new wearable-based posturography method could be a valid alternative to the “classic” analysis of static sway [8]. Indeed, a recent systematic review [7] showed that IMU sensor-derived variables are moderate to highly correlated to CoP variables (r from 0.58 to 0.84 for mediolateral and anteroposterior sway) and are also able to distinguish between people of different ages (young vs. elderly) and health status (healthy individuals vs. Parkinson's disease, multiple sclerosis, and other neurological conditions).

The ability to maintain balance control could be challenged by using different surfaces, reducing the base of support (i.e., feet position), or by sense deprivation (i.e., open and closed eyes), all conditions that can produce a greater magnitude of sway variables. In particular, visual deprivation is known to affect balance control negatively [3], and the ratio between closed- (EC) and open-eye (EO) conditions is frequently used to assess the visual contribution to the static sway ability. While it is well known that both force plate-derived CoP variables and IMU sensor-derived CoM variables are sensitive to the changes between EO and EC conditions [3,12,13] in a wide spread of neurological disorders [14], less is known about the ability of IMU sensors to detect these differences in healthy older individuals. To the best of our knowledge, no studies di-rectly evaluated the relationship of the IMU sensor to detect changes in static sway variables due to visual deprivation compared to the gold standard instrument.

Therefore, the purpose of this study is twofold. In a group of healthy elderly individuals, we aimed to (i) evaluate the relationship between sway variables derived from an IMU sensor and derived from a force plate during static sway trials; (ii) evaluate the relationship between the variables derived from an IMU sensor and force plate in detecting CoM and CoP-related changes due to visual deprivation. We hypothesized that the IMU sensor and force plate-derived variables are strongly correlated and that the IMU sensor can detect changes in postural sway related to visual deprivation as the force plate.

## METHODS

### *Participants*

A total of 65 healthy older individuals (32 females, 33 males) were recruited by local advertisement. The inclusion criteria were an age above 60 years, while a preliminary telephone interview and a successive medical screening allowed the exclusion of individuals with any orthopedic, mental, or neurological diseases that could interfere with the postural control. All participants signed a written informed consent form before participating. All procedures used in the study were approved by the Ethics Committee for Human Research from the University of Verona (28/2023) and were conducted in conformity with the Declaration of Helsinki.

### *Data Collection*

Participants visited the laboratory one time. Personal (sex, age) and anthropometric (weight, height) data for each participant were first recorded. The anthropometric assessment was performed with participants barefoot and wearing only underwear. Body mass was taken to the nearest 0.1 kg with an electronic scale (Tanita electronic scale BWB-800 MA, Tokyo, JP), and stature was measured with a Harpenden stadiometer (Holtain Ltd., Crymych, Pembs., UK) to the nearest 0.5 cm. Body Mass Index (BMI) was calculated as body mass/height2 (kg/m2). Then, participants completed two (EO and EC conditions) 30 s standing balance tests with feet in a semi-tandem position (i.e., with the toe of the back foot in contact with the mid-front foot) [5]. The two tests were performed in randomized order.

For each test, all participants were instructed to stand upright with both feet on a single force plate (1000 Hz, AMTI Inc., Watertown, MA, USA), while, simultaneously, a single IMU sensor (500 Hz, GYKO, Microgate, Bolzano, Italy) was placed on the lumbar spine (i.e., L5 level) worn with the dedicated belt. According to the manufacturer's user instructions, the IMU sensor was oriented into the belt's pocket with the lead upward and outward, providing the x-axis to measure the anteroposterior displacement and the y-axis to measure the mediolateral displacement. The sensor's height from the force platform's surface was recorded and inserted in the dedicated software (GYKORe-Power, Microgate, Bolzano, Italy), which was in communication with the IMU sensor via Bluetooth®.

An operator showed the correct posture, which consisted of standing still with the feet in a semi-tandem position. There were no constraints on the arms' position. After that, a familiarization session lasting 2 min was performed before each test. The same dedicated operator remained close to the participant to prevent any risk of falls while another operator monitored the instruments. The force plate and IMU sensor were manually synchronized after a countdown of three, and the trial recording automatically stopped after 30 s. The trial was interrupted and repeated if the participants moved their feet or grasped the operator for support.

*Data Analysis*

Regarding the COP-derived variables, raw data from the force plate were collected and subsequently analyzed using a self-written MATLAB code. Briefly, the force signal was low pass filtered at 5 Hz using a fourth-order Butterworth filter. After that, the total length, root mean square (RMS) distance, mean velocity, and 95% confidence interval ellipse area (95% CI ellipse area), as well as the anteroposterior (AP) and mediolateral (ML) component for each metric, were extracted following standard procedures [4].

The calculation of the IMU sensor is based on the inverse pendulum model [10,11], which relates the controlled variable CoM with the controlling variable CoP, stating that the difference between these two physical quantities is proportional to the CoM horizontal acceleration, and this relation holds in both the sagittal and frontal plane (i.e., the AP and ML direction) [11]. The model is based on two assumptions: (i) all the subject's body mass is concentrated in one point (i.e., the CoM); (ii) the CoM is at the top of the inverse pendulum and is directly linked to the ankle joint by a rigid segment (the knee and hip joints are not considered). In a quiet state, the momentum applied on the CoM is counterbalanced by the active momentum applied by the ankle joint. Therefore, by using the accelerometer and gyroscope values and by knowing the height from the surface of the force plate (i.e., L5 to the surface of the force platform distance), it is possible to measure the horizontal displacement of the COM (in both directions).

The dedicated software automatically computed the CoM data from the IMU sensor, and the same CoP variables derived from the force plate were extracted. Finally, the ratios between variables derived from the EC and EO conditions were calculated [15].

*Statistical Analysis*

All data were calculated and reported as mean ± standard deviation. Shapiro–Wilk tests were run to test the normality of data distribution, and nonparametric tests were used when the assumption of normality was violated. A 2-way ANOVA was run to test differences between instruments and eye conditions. Pearson product-moment correlation (parametric data) or Spearman Rank correlation (nonparametric data) analyses were run to test the linear correspondences between eye conditions within the same instrument. Correlations were computed between instruments within the same eye condition to test the relationship between instruments in measuring the sway variables.

Finally, paired t-tests and correlations were also performed for the ratios of the most correlated variables between instruments. Correlation coefficients were interpreted as negligible (<0.1), weak ($0.10 \leq x < 0.40$), moderate ($0.40 \leq x < 0.70$), strong ($0.70 \leq x < 0.90$), or very strong (≥0.9) [16]. Statistical significance was set at $p < 0.05$. SigmaPlot 12.5 (SigmaStat, San Jose, CA, USA) was used for all the statistical analyses.

RESULTS

Participants' anthropometric characteristics are reported in Table 1.

Table 1. Anthropometric characteristics of the participants.

| | **Age (Years)** | **Weight (kg)** | **Height (m)** | **BMI (kg/m$^2$)** |
|---|---|---|---|---|
| Mean | 66.7 | 73.1 | 1.68 | 25.9 |
| Standard deviation | 5.83 | 15.3 | 9.58 | 4.70 |
| Minimum | 60 | 48.9 | 1.52 | 17.5 |
| Maximum | 81 | 124 | 1.88 | 40.7 |

BMI: body mass index.

Absolute values of the variables, along with their within and between instrument correlation coefficients, are reported in Table 2 for both eye conditions.

**Table 2:** Open and closed eye variables for both force plate and IMU sensor instruments, and correlation coefficients between instruments and between eye conditions within each instrument

| | EO condition | | | EC condition | | | EO-EC correlation | |
|---|---|---|---|---|---|---|---|---|
| | Force plate | MU sensor | | Force plate | IMU sensor | r | Force plate | IMU sensor |
| Total length | 796.8±247.4 | 45.7±104.5† | .62$^{***}$ | 1322.5±514.5$^{\#}$ | 637.5±243.8$^{\#}$† | 0.88$^{***}$ | 0.73$^{***}$ | 0.65$^{***}$ |
| AP | 440.2±156.1 | 41.1±65.58† | .50$^{***}$ | 736.6±314.3$^{\#}$ | 355.9±144.3$^{\#}$† | 0.82$^{***}$ | 0.68$^{***}$ | 0.67$^{***}$ |
| ML | 581.4±177.4 | 24.9±80.41† | .50$^{***}$ | )61.3±369.0$^{\#}$ | 456.1±178.5$^{\#}$† | 0.74$^{***}$ | 0.72$^{***}$ | 0.67$^{***}$ |
| RMS distance | 9.2±2.6 | .2±1.7† | .58$^{***}$ | 12.0±3.8$^{\#}$ | 6.5±2.7$^{\#}$† | 0.75$^{***}$ | 0.45$^{***}$ | 0.36$^{*}$ |
| AP | 6.1±2.3 | .0±1.7† | .39$^{**}$ | 7.2±2.6$^{\#}$ | 7.9±3.1$^{\#}$ | 0.55$^{**}$ | 0.43$^{**}$ | 0.43$^{**}$ |
| ML | 6.8±1.8 | .8±3.4 | .43$^{***}$ | ).5±3.1$^{\#}$ | 10.2±4.4$^{\#}$ | 0.74$^{***}$ | 0.38$^{**}$ | 0.43$^{**}$ |
| Mean velocity | 26.4±8.25 | 4.8±3.47† | .62$^{***}$ | 44.7±17.3$^{\#}$ | 21.5±8.2$^{\#}$† | 0.88$^{***}$ | 0.78$^{***}$ | 0.63$^{***}$ |
| AP | 14.6±5.20 | .05±2.18† | .51$^{***}$ | 24.9±10.9$^{\#}$ | 12.0±4.8$^{\#}$† | 0.82$^{***}$ | 0.74$^{***}$ | 0.64$^{***}$ |
| ML | 19.3±5.92 | 0.8±2.68† | .48$^{***}$ | 32.3±12.1$^{\#}$ | 15.4±6.0$^{\#}$† | 0.75$^{***}$ | 0.75$^{***}$ | 0.69$^{***}$ |
| 95% CI ellipse area | 876.0±594.0 | 8.5±48.9† | .65$^{***}$ | 1507.6±963.2$^{\#}$ | 144.5±101.8$^{\#}$† | 0.89$^{***}$ | 0.45$^{**}$ | 0.44$^{***}$ |

EO, open eye; EC, closed eye; IMU, inertial measurement unit; AP, anterior-posterior; ML, medio-lateral; RMS, root mean squared.

# indicates significant difference between eye conditions; † indicates significant difference between instruments; * indicates significant relation: *$p<0.05$; **$p<0.01$;***$p<0.001$

ANOVA showed a statistically significant main effect of the instrument and eye conditions for all the variables except for the RMS distances, which did not display a main effect of the instrument on the ML axis for the OE condition and on both AP and ML axes for the CE condition. Overall, the variables for open- and closed-eye conditions derived from the instruments were weak–moderately to strongly correlated (force plate r, from 0.38 to 0.78; IMU sensor r, from 0.36 to 0.69). Moderate to strong correlations (r from 0.50 to 0.88) were also found between instrument variables for both eye conditions, except for the RMS distance in the mediolateral direction for the open-eye condition ($r = 0.39$).

The comparison between means and correlation of the ratios derived from both instruments are displayed in Figure 1. The mean values of RMS distance and 95% CI ellipse were not different between instruments, while there was a significant difference between the total length and mean velocity variables ($p = 0.973$ and $p = 0.703$, respec-tively). Moderate to strong correlation coefficients were found for all the ratios (r from 0.65 to 0.87).

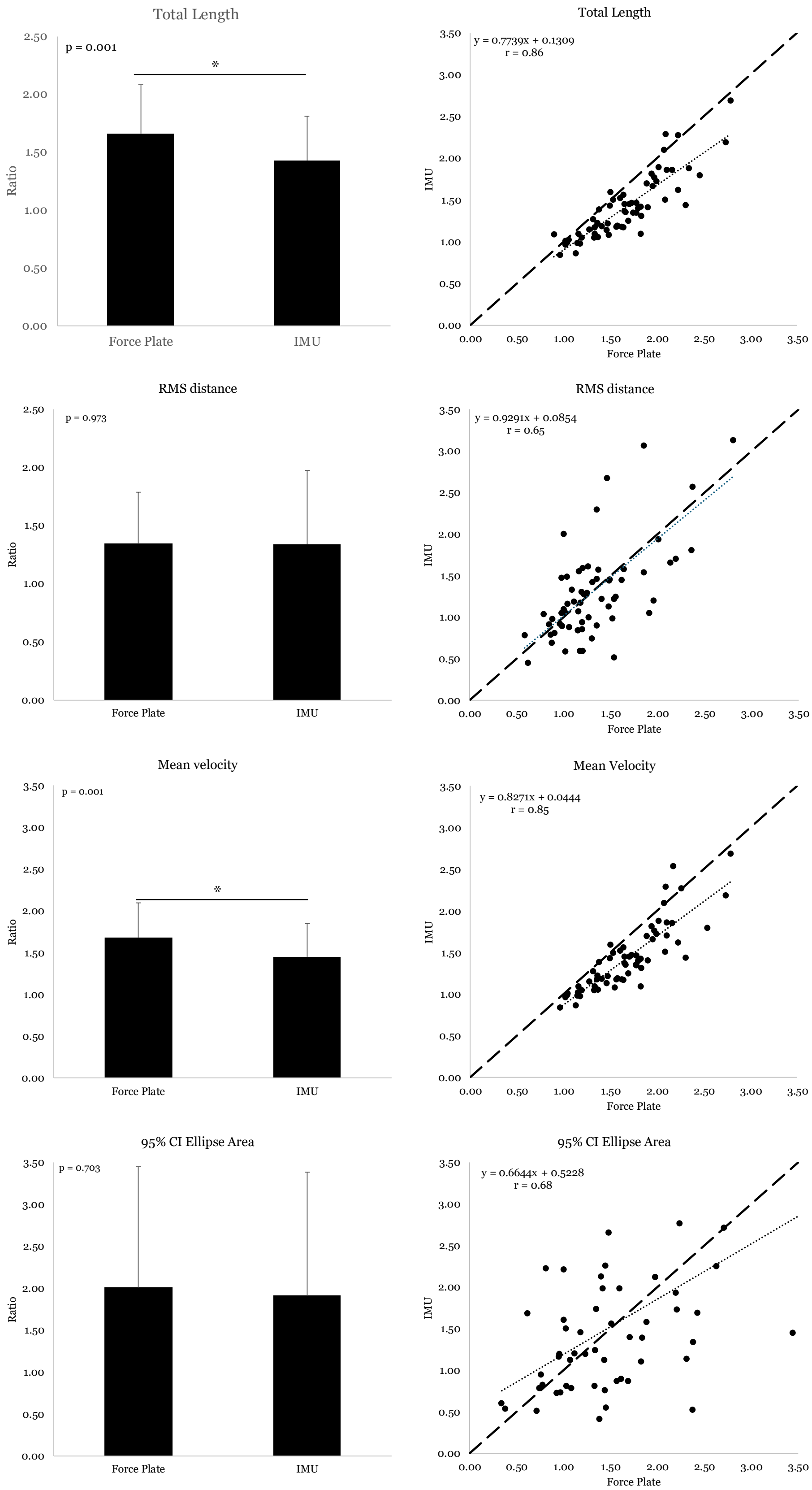


Figure 1: Comparison of mean between ratios along with correlation plots are displayed. * indicates statistical significance difference between means. Ratios were calculated by dividing the closed eye values by the open eyes ones.

## DISCUSSION

The first aim of this study was to assess the relationship between static sway variables derived from an IMU sensor compared to a force plate in healthy older adults. Our results indicate that IMU sensor-derived variables showed close to moderate to strong agreement (r ranged from 0.38 to 0.87) with the gold standard instrument. The second aim of this study was to assess the ability of the IMU sensor to discriminate between open- and closed-eye trials compared to the force plate. Our findings confirmed that the variables derived from the open-eye trials using the IMU sensor were significantly different from those derived from the closed-eye trials (except for RMS distance in the anteroposterior direction). In addition, the IMU sensor showed moderate to strong agreement with the gold standard instrument for the EC and EO ratio across all the considered variables.

A direct comparison between the absolute values measured in our study with those reported in the literature is difficult because of a wide range of populations, foot positions, eye conditions, trial duration, and variables that have been analyzed in studies investigating static sway. When considering a similar foot position (semi-tandem), eye condition (open and closed), and the instrument used (force plate and IMU sensor), our values are comprehensively consistent with the literature regarding a population of healthy adults of both sexes [3,17–21]. The need to obtain valid, repeatable, and objective measures of static sway while using an affordable approach in clinical assessments has increased the attention toward wearable IMU sensors. Despite the appeal of these solutions, their actual validity and overall capability to track changes over time and between conditions have not yet been fully explored [8]. Some challenges arise from the assumptions underlying the use of IMU sensors for tracking the CoM. IMU sensors derive spatial, velocity, and area variables from CoM accelerations, whereas force plates measure variables based on the ground reaction force at the CoP. The inverted pendulum model linking the CoM and CoP through a rigid segment anchored at the ankle joint implies that every CoP displacement produces a related and proportional acceleration of the CoM, resulting in a correlation close to 1 [9,11]. However, while correct from a physics standpoint, this model overlooks the possible role of knee and hip joints, which can play a role in maintaining postural balance during quiet standing [9,11].

In addition, since IMU sensors and force plates measure different physical quantities, direct comparisons are impossible. Therefore, correlation analysis is commonly used to test the relationship between IMU sensors and force plate measures [22,23]. In agreement with previous work, our study found moderate to strong correlations for sway length (total length and in both directions: 0.50 to 0.88; RMS distance and in both directions: 0.39 to 0.75), velocity (0.48 to 0.88) and area (0.65 and 0.89) variables [7,24] (Table 2). The relationship between the IMU sensor and the force plate confirms that IMU sensors can be a valid solution for measuring static sway in healthy older adults. The availability of population-specific normative data on sway ability is essential for examining individual data, facilitating interpretation in support of decision-making and individualized exercise prescription (i.e., level- and goal-specific interventions). In this context, our study offers a medium-sized database of sway ability in healthy older individuals of both sexes.

The ability to maintain balance control is challenged by sense deprivation (i.e., open and closed eyes), which can lead to a greater magnitude of sway variables. In accordance with the literature [3], all the variables measured in our study were significantly greater in the EC compared to the EO condition for both force plates (absolute changes in total length +66%, RMS distance +30%, mean velocity +69%, and 95% CI ellipse area +72%) and IMU sensor-derived data (absolute changes in total length +43%, RMS distance +25%, mean velocity +45%, and 95% CI ellipse area +63%) (Tabel 2). The above absolute changes align with what is expected in this population of healthy older adults [3]. Moreover, in agreement with two previous studies that compared wearable IMU sensors to gold-standard instruments, we confirmed a correlation between instruments within EC and EO conditions [24,25] (Table 2).

The relative change in the parameters obtained between EC and EO conditions is often used to assess proprioception-related neurologic disease by removing the visual and vestibular components contributing to balance maintenance [15]. In our study, we found ratios (expressed as % of EO condition) between variables derived from both instruments that are comparable to what was found in the literature for health- and age-matched individuals (from 134% to 201% and from 134% to 191% for the force plate and IMU sensor, respectively, vs. 147.4% ± 120.6%) [26,27] (Figure 1). In addition, our study was the first to directly focus on the ability of wearable IMU sensors to quantify the amplitude of the changes in sway parameters induced by the EC postural challenge, compared to gold standard instruments. The correlation coefficients (from 0.65 to 0.87, $p < 0.001$) showed that the ratios measured across all parameters were moderately to strongly related, highlighting the relationship between the variables derived from the instruments (Figure 1).

Tracking changes in postural sway over time is crucial when monitoring the healthy aging trajectory, and wearable sensors could be a valid, low-cost, and simple tool able to achieve this. However, this study did not investigate the IMU sensor's ability to track changes over time. Future studies should investigate wearable sensors' day-to-day reliability and sensitivity in detecting fine and subtle changes in postural sway. Moreover, practitioners and clinicians should be aware of the possible source of measurement error that arises from assessing postural sway under real-life conditions. Indeed, the moderate to strong correlations between IMU sensors and force plates in our study could be challenged outside the strictly controlled laboratory setting (i.e., clinical, home-based environments or self-administration). Among the sources of error that could arise from this type of setting, the strict control of the right posture during the sway assessment plays a key role. The biomechanical model (i.e., inverted pendulum model) on which the construct validity of the IMU sensor is based could be outflanked by postural sway strategies at the hip, knee, and ankle joints. For example, the pendulum's length is measured by taking the height of the IMU sensor from the ground. If, during the sway trial, a subject slightly bends the knees, the final sway could result in less absolute values. Similarly, if a subject relies on hip strategies for maintaining static sway, the postulation that the inverted pendulum model is based on a rigid segment fails, and the resultant sway variables will be affected.

## CONCLUSIONS

The current study examined the construct validity of an IMU sensor for postural sway assessment in healthy older individuals. Our results suggest that the IMU sensor is a valid alternative for postural sway assessment, as its CoM-derived measurements showed strong correlations with force plate-derived measurements. Moreover, this is the first study to investigate the ability of the IMU sensor to discriminate between trials with open and closed eyes in healthy older adults. The IMU sensor consistently distinguished and detected the expected differences in sway variables related to visual deprivation as the gold standard method. Therefore, the IMU sensor emerged as a valid option for evaluating sway and its perturbations induced by postural challenges in healthy older adults.

From a practical standpoint, the simple testing of the semi-tandem stance in both open- and closed-eye conditions, measured with an IMU device, could serve as a time-efficient yet valid tool for assessing sway ability in healthy individuals on a large scale. The relatively low cost and simplicity of use offer the opportunity to perform objective, digitalized, ample, and time-resolved measures in a clinical setting. This instrumented posturography could be used as a tool for the early detection and frequent monitoring of postural sway performance.

# GENERAL DISCUSSION

Home-based exercise programs have emerged as a viable alternative to the traditional Resistance Training (i.e. performed in specialized facilities with specialized equipment), demonstrating positive effects on physical function. However, these programs often lack proper monitoring of exercise execution (Chaabene et al., 2021; Lacroix et al., 2017). Various technological solutions developed by the industry (i.e. Inertial Measurement Units) aim to fill the gap between exercise prescription and monitoring the execution of home-based training programs. Often, these devices combine wearable sensors and dedicated software that easily delivers the prescription of exercises from the trainer and monitors the correct and actual execution of the training session. While the above features appear extremely promising, the overall safety, feasibility, and efficacy of these devices in the specific context of sarcopenia prevention in older adults remain to be determined.

The first part of this thesis investigated the safety, feasibility, and effectiveness, on body composition, balance, gait, and strength, of a 6-month home-based resistance training program administered through an innovative technological solution, in healthy older adults. We found that the exercise intervention delivered through the home-based device was feasible in terms of safety and adherence. In fact, no adverse outcomes were recorded throughout the study. At the same time, the compliance was very good for the first 3 months (Mean Training Frequency: 2.8 ± 1.1, 61% of participants performed the recommended 3 sessions, 78% of participants performed at least 2 sessions per week), yet decreased markedly thereafter (Mean Training Frequency: 1.9 ± 1.3 and 42% of participants performed the recommended 3 sessions while 55% of the participants performed 2 sessions per week). In our study, which was the first to measure objectively and automatically qualitative and quantitative adherence in real-time, this parameter should be free of overestimation compared to previous studies which recorded adherence using training diaries filled out by the participants, an approach that is known to potentially overestimate this value (Chaabene et al., 2021). Regarding the effectiveness, the training program had a marginal or no effect on body composition, balance, and muscle function indexes, except for the walking parameters and the maximal force of the lower limbs during the sit-to-stand task, which displayed a modest yet significant increase. We speculate that our program may not have been sufficiently intense or specific to effectively stimulate a gain in muscle mass and/or a reduction in body fat in our healthy older individuals. Moreover, the benefits of resistance training on balance are thought to be mediated by improved neuromuscular control, which our intervention seemed not to have sufficiently stimulated. We found an improvement in all walking parameters that are comparable to the improvements described following traditional resistance training. The slow and controlled execution imposed by the device could have provided an adequate stimulus to improve muscle strength, but it could have underexposed our participants to muscle power adaptations. Taking all together, implementing a comprehensive resistance training intervention that includes power and balance training may be a better approach to improving overall muscle function in healthy older adults, with the challenge of ensuring safety in performing high-velocity movement tasks in a home-based self-managed remote context. While the extent of the long-term benefits of our home-based resistance training on muscle mass and function appear limited, exercise therapists and practitioners should consider this low-cost and accessible approach whenever barriers to an active lifestyle and participation in traditional resistance exercise programs are present.

The second part of this thesis explores the portability and feasibility of common physical tests instrumented with IMU technology.

The first study investigated the validity of IMU sensors for assessing the lower limb musculature's maximal strength and power abilities during a sit-to-stand task.

This was the first study that tested the accuracy of an IMU in estimating muscle power during the 5STS test in comparison with a fully objective, gold standard, and automated method. Our data indicate that a single IMU placed on the lateral face of the thigh provides estimates of kinetic and kinematic indexes of muscle action, as well as of muscle power, during the 5STS test which are highly correlated with gold standard laboratory measures.

Within our study, IMU-based measures of mean concentric time were not different, very highly correlated ($r = 0.93$), without significant bias (bias = 0.00, z-score =0.14), and with small limits of agreement (L.O.A.= [0.12 s -0.12 s]) compared to the gold-standard laboratory approach. This would indicate that the method used for the data analysis of the IMU signal identifies the same “temporal windows” as the laboratory approach.

Moreover, while IMU slightly underestimates the velocity of the movement compared to gold standard methods, possibly due to methodological matters (i.e.: amplitude of the sensor-based acceleration signal on the mid-thigh), it displays a very high correlation ($r=0.76$), significant and constant bias (bias= -0.08 m·s-1, z-score= -9.85, R2=0.030) with small limits of agreement (L.O.A. = [0.05 -0.20 m·s-1]). Then, the IMU could be considered overall a valid and precise instrument for estimating and monitoring mean concentric velocity.

In addition, we found that the force values between instruments were different ($p < 0.001$) but with almost perfect correlation ($r=0.97$), a very small significant bias (bias = -20 N; z-score = -4.78), and small limits of agreement (L.O.A. = [44 -84 N]).

IMU power values were significantly lower than the values measured with laboratory equipment ($p < 0.001$). As above mentioned, power difference is attributable to the lower IMU mean concentric velocity measured at the mid-thigh since the force was similar to the laboratory's values. This discrepancy could be attributable to the placement chosen for the IMU on the participants which is less prone to greater accelerations. While the bias was significantly different, the limits of agreement were relatively small (bias = -58 W, z-score = -10.07, LOA = [32 -150 W]). In addition, a null relationship was observed ($R^2 = 0.000$) indicating that the difference between instruments is similar across the entire range of power.

While a significant difference was recorded between measures, likely due to the sensor placement, wearable sensor-based 5STS evaluation could allow a valid, low-cost alternative to the gold standard methods classically used to measure muscle power; moreover, it could be a more repeatable, objective, and immediately digitalized option compared to the stopwatch method.

The second study investigated the validity of the IMU sensor to correctly characterize the force-velocity profile of the lower limbs during a modified sit-to-stand test.
A first aim was to investigate the feasibility and validity of the loaded 5STS test compared to the gold standard (i.e.: Isokinetic Strength test, ISO) for the assessment of F0, V0, Pmax, and Vopt through F-v profiling in older adults. A second aim was to investigate the relationship between F0 and Pmax, as measured with both loaded 5STS and the gold standard approach, with clinical markers of muscle mass and strength. We have found that 5STS profiling is a feasible and valid alternative to isokinetic testing for characterizing muscle function in healthy older adults of both sexes. In fact, all participants completed all the trials for both ISO and 5STS tests without experiencing any adverse outcomes. Furthermore, the $R^2$ mean of F-v profiling in 5STS ($0.97 \pm 0.03$) was high and similar ($p = 0.581$) to the ISO ($0.97 \pm 0.03$). Moreover, while the absolute values of maximum force and maximal and optimal velocity significantly differ between the two tests, the maximum power values measured in 5STS and ISO are similar and highly correlated. We speculate that the difference between the expression of lower limbs' maximum muscle strength between tests is due to the difference in muscle action (in terms of neuromuscular and biomechanical characteristics, e.g., muscle coordination of single vs. multi-joint movements and different contraction lengths) required by the two movements (single-joint vs. multi-joint), and difference in the applied load (weight-bearing versus non-weight-bearing) between tests. The comparison of Pmax between the ISO and 5STS tests showed

a non-significant difference and a high correlation ($p = 0.259$, $r = 0.84$). Bland–Altman analysis reported a non-significant and constant bias (bias = 5.7 W, $p = 0.259$) with small limits of agreement (L.O.A. = [−60 to 71 W] ). Maximum and optimal velocity results differed and were poorly correlated ($r = 0.23$, $p < 0.001$) between ISO and 5STS. Bland–Altman analysis reported a significant bias (V0, bias = $-1.6$ m × $s^{-1}$, $p < 0.001$; Vopt, bias = $-0.8$ m × $s^{-1}$, $p < 0.001$) with high limits of agreement (V0, L.O.A. = [−2.3 to −1.0 m × $s^{-1}$]; Vopt, L.O.A. = [−1.1 to −0.5 m × $s^{-1}$]). Our results confirm that velocity parameters extrapolated from the F-v relationship seem to have less concurrent validity and precision than other indexes (Bochicchio, Ferrari, Bottari, Lucertini, Scarton, et al., 2023). Therefore, these parameters could have less relevance in medical screening and clinical assessment. Regarding the second purpose of this study, both maximal muscle strength and power were significantly and highly correlated (r from 0.65 to 0.82) with the most commonly used clinical markers of muscle mass and strength. It is well-known that the decline of these variables is associated with adverse outcomes in aging (frailty, impaired physical function, and disability in daily living activities) (Alcazar et al., 2018; Cruz-Jentoft et al., 2019). Therefore, muscle profiling could be used as a monitoring tool for the early detection of individuals at higher risk of unhealthy aging and provide a valuable tool for the individualization of training interventions

Finally, the third study evaluated the validity of the IMU sensor to asses static sway ability during closed and opened eyes trials. The purpose of this study was twofold. In a group of healthy elderly individuals, we aimed to evaluate: i) the coherence between sway variables derived from an IMU sensor and a force plate during static sway trials; ii) the coherence of the IMU sensor in detecting CoM-related changes between open ( EO) and closed eye (EC) conditions compared to the force plate. Some challenges arised from the assumptions underlying the use of IMU sensors for tracking the CoM. IMU sensors derive spatial, velocity, and area variables from CoM accelerations, whereas force plates measure variables based on the ground reaction force at the CoP. The inverted pendulum model linking the CoM and CoP through a rigid segment anchored at the ankle joint (i.e., not considering the possible role of knee and hip joints) implies that every CoP displacement produces a related and proportional acceleration of the CoM, resulting in a correlation close to 1 (Mancini et al., 2012; Mengarelli et al., 2019). Therefore, since IMU sensors and force plates measure different physical quantities, direct comparisons are not possible and correlation analysis is commonly used to test the coherence between IMU sensors and force plate measures (Ekvall Hansson & Tornberg, 2019; Heebner, Akins, Lephart, & Sell, 2015). Our results indicate that IMU sensor-derived variables showed close to moderate to strong agreement (r ranged from 0.38 to 0.87) with the gold standard instrument. The ability to maintain balance control is known to be challenged by sense deprivation (i.e., open and closed eyes), which can lead to a greater magnitude of sway variables (Roman-Liu, 2018). Our findings confirmed that the variables derived from open eyes trials using the IMU sensor were significantly different from those derived from closed eyes trials (with the only exception of Root Mean Square distance in the anteroposterior direction). In addition, the IMU sensor showed moderate to strong agreement with the gold standard instrument for the EC and EO ratio across all the considered variables. Therefore, the IMU sensor emerged as a valid option for evaluating sway and its perturbations induced by postural challenges, in healthy older adults. From a practical standpoint, the simple testing of semi-tandem stance in both open and closed eye conditions, measured with an IMU device, could serve as a time-efficient yet valid tool for assessing sway ability in healthy individuals on a large scale. This instrumented posturography could be used as a tool for the early detection and frequent monitoring of postural sway performance.

# CONCLUSION

The decline in muscle strength and power affects older individuals' abilities to perform daily activities, increasing the risk of functional limitations, adverse health outcomes (falls and mortality), and reduced independence (Alcazar et al., 2018; Cruz-Jentoft et al., 2019). Therefore, the early detection of deviations from the normal aging trajectory in older individuals becomes crucial in a world where the older population will double within 2050 with a concomitant increase in life expectancy (World Health Organization, 2019).

Efforts in reaching this special population to inform about and promote a healthy and active lifestyle are crucial to limit future pressures on the medical care system and finances (J. Chen et al., 2023; Lopreite & Mauro, 2017).

In this context, this investigation demonstrated that training interventions and assessments of physical function in a remote, technologically assisted in-home environment could be a valid, valuable, and viable alternative approach. Home-based resistance training seems to counteract the age-related decay in muscle strength and walking abilities while protecting against the decay in body composition, sway ability, and lower limb muscle power. Moreover, the instrumentations of common clinical tests could be a cornerstone to access affordable yet valid, reliable, objective, and quantitative measures for assessing physical function (muscle strength and power) and performance (sway ability).

Healthcare professionals and clinicians should consider a home-based approach when barriers to an active lifestyle are present.

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