BioMed Research International

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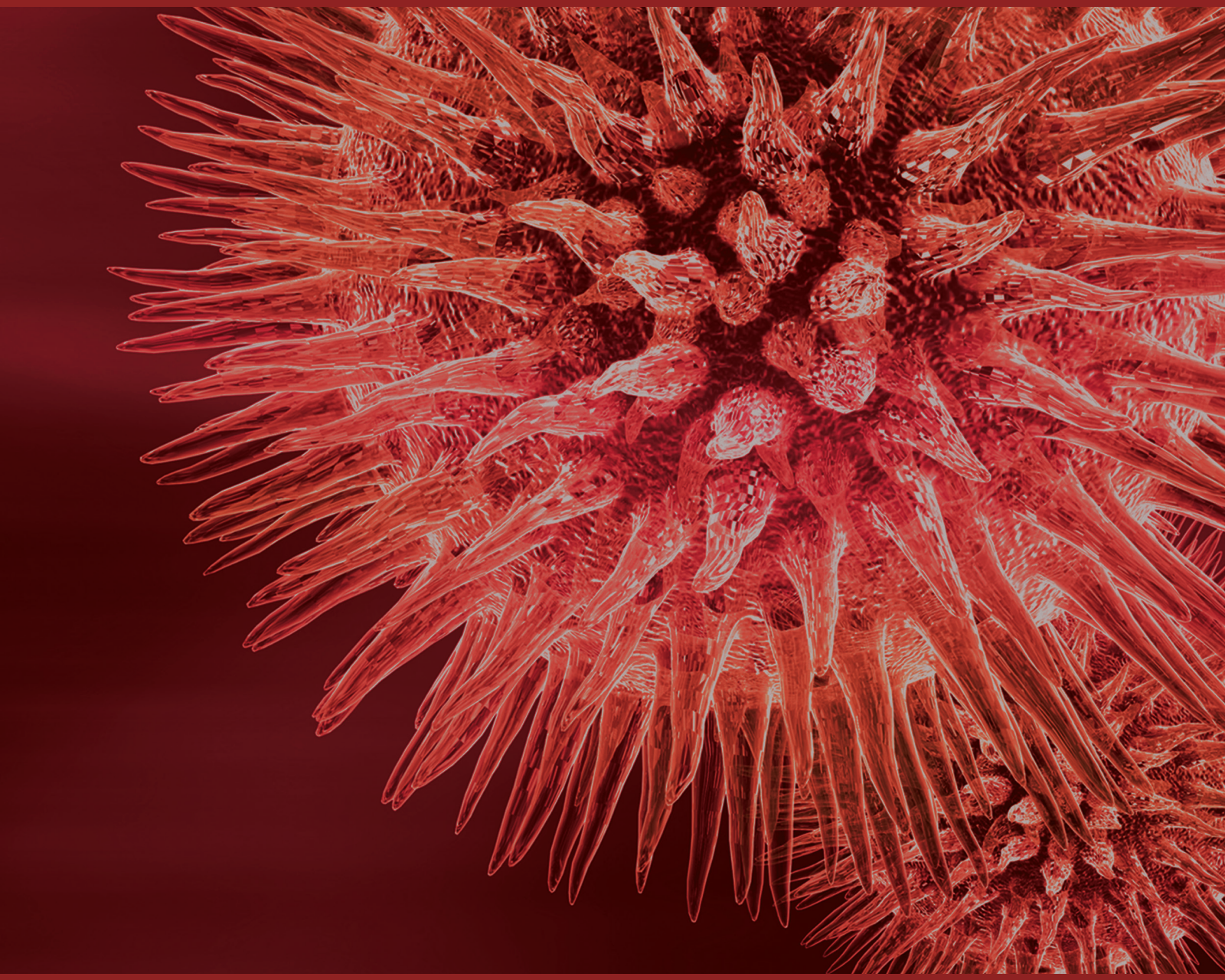
Guest Editors: Jullia Maria D'Andréa Greve, Angelica Castilho Alonso, Luiz Mochizuki, Paulo R. Lucareli, Chandramouli Krishnan, and Richard Baker



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Editorial

Motor Functional Evaluation from Physiology and Biomechanics to Clinical and Training Application

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This special issue was proposed to combine biology, biomechanics, and human locomotion and the use of this knowledge to improve rehabilitation; prescription of physical activity for health promotion and for sports performance are important for the science, not only for the gait but for every type of movement and exercise.

In this special issue, men and women were compared in sports, physical activity, and standing posture. P. G. Mouroço and colleagues show that men and women achieve fast front crawl swimming with different strategies; while men use arm stroke, women swim fast using whole-body mean forces. M. Persiani and colleagues showed that men consistently had larger COP parameters than women. They applied lateral optic flow perturbances and found that they cause asymmetry in postural balance and different lateralization of postural controls in men and women. H. Jaafar and colleagues evaluated force-velocity relationship during cycling and arm cranking exercise. They compared the reliability of maximal power, maximal pedal rate, and maximal force in active men and women and showed that they are different according to sex. It is interesting to see how close or far human performance is according to biology.

In rehabilitation, evaluation of functional movements is the key element in building a rehabilitation program. M. Caimmi and colleagues provide a set of normative data from healthy subjects of different ages doing upper-limb functional tasks (reaching and hand-to-mouth movements). Their results are important for upper-limb evaluation in different populations. Normative data in quiet standing is important as well. Therefore, A. C. Alonso and colleagues showed that anthropometric variables can explain part of center of pressure sway quiet standing.

Developing techniques to evaluate body motion during walking is necessary to understand how people move. C. Buckley and colleagues measured upper body movement during walking and associated it with postural control in persons with Parkinson's disease (PD). They showed that persons with PD attenuate body accelerations differently compared to controls. E. Bergamini and colleagues proposed a method to evaluate wheelchair propulsion based on biomechanics and successfully applied the results to regulate a training program for young wheelchair athletes.

But technology is not only applied in rehabilitation to improve measurements; it is applied as an intervention. C. Luque-Moreno and colleagues reviewed and compared VR intervention in stroke, looking for frequently used outcome measures. Multimodal approach, combining VR and conventional physiotherapy, presented the best results in balance and gait. VR mostly improves gait speed, balance, and motor function in stroke patients.

Exercise-based interventions improve performance. N. Hedayatpour and D. Falla reviewed neuromuscular

adaptations to eccentric exercise, depicting histochemical, metabolic, and neural adaptations due to eccentric training. Besides, exercise changes human biology. W. Y. V. Mak and W. K. C. Lai showed that short duration of resistance exercise may provoke a transient increase in central arterial stiffness in healthy young men. They analyzed acute effect of Valsalva manoeuvre during resistance exercise on arterial stiffness.

So, this special issue is very rich and joins different views of the motion study in the same publication, empowering the knowledge for all professionals of this area.

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Luiz Mochizuki
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Research Article

Strength, Multijoint Coordination, and Sensorimotor Processing Are Independent Contributors to Overall Balance Ability

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For young adults, balance is essential for participation in physical activities but is often disrupted following lower extremity injury. Clinical outcome measures such as single limb balance (SLB), Y-balance (YBT), and the single limb hop and balance (SLHB) tests are commonly used to quantify balance ability following injury. Given the varying demands across tasks, it is likely that such outcome measures provide useful, although task-specific, information. But the extent to which they are independent and contribute to understanding the multiple contributors to balance is not clear. Therefore, the purpose of this study was to investigate the associations among these measures as they relate to the different contributors to balance. Thirty-seven recreationally active young adults completed measures including Vertical Jump, YBT, SLB, SLHB, and the new Lower Extremity Dexterity test. Principal components analysis revealed that these outcome measures could be thought of as quantifying the strength, multijoint coordination, and sensorimotor processing contributors to balance. Our results challenge the practice of using a single outcome measure to quantify the naturally multidimensional mechanisms for everyday functions such as balance. This multidimensional approach to, and interpretation of, multiple contributors to balance may lead to more effective, specialized training and rehabilitation regimens.

1. Introduction

It is well known that both sensory and motor systems contribute to the ability to maintain balance. Sensory inputs are necessary to detect unstable conditions (i.e., perturbations to the system) and motor contributions are vital to initiate timely and appropriate responses to counteract these perturbations. Clinical outcome measures such as single limb balance (SLB), Y-balance (YBT), and the single limb hop and balance (SLHB) tests are commonly used to quantify balance in individuals when they are healthy [1–4] or following musculoskeletal injury (e.g., ankle sprains and anterior cruciate ligament (ACL) tears) [5–11] or to assess risk for lower extremity injury [6, 12–14]. Results obtained from these tests are used to represent the mechanisms of balance. However, the contributions of sensory inputs and appropriate motor responses necessary to perform well vary across them. Outcome measures that include smaller changes in lower limb or whole-body position are typically considered measures of static stability of balance, whereas measures that include larger changes in position are often referred to as dynamic stability of balance. One may argue that detection of smaller changes in position or motion would be more challenging for the sensory system to detect and less challenging for the motor system to counteract; conversely, large changes in position or motion would be more easily detected by the sensory system and, in turn, place greater demands on the motor system to counteract in terms of strength and multijoint coordination. As a result, interpretation of the outcomes with respect to underlying sensory or motor deficits becomes challenging when considering the range of static and dynamic measures used to quantify balance.

Unperturbed single limb balance during quiet standing balance tests generally result in relatively small joint excursions and are considered measures of static balance. This requires detection of smaller, subtler sensory stimuli and

relatively small motor responses to maintain balance. In contrast, successful performance on balance tests such as the single limb hop and balance and Y-balance tests involve larger changes in position and are considered measures of dynamic balance. The SLHB quantifies the ability to stabilize the center of mass (COM) after completing a forward hop on a single limb. The transition from a dynamic to a static state can be considered a perturbation to the COM, thus making it a measure of dynamic balance. Performance of both SLB and SLHB is quantified using outcome measures related to center of pressure (COP) movement because they represent corrective actions made to maintain balance [15]. Additionally, performance of the YBT is scored by measuring the farthest distance reached with the free limb while maintaining balance on the stance limb. The maximal reach distances in each of the three directions are considered measures of dynamic balance because changing the spatial orientation of the free limb acts as a perturbation to the COM with respect to the base of support (BOS), or stance limb. For more dynamic tests, while detection of larger joint excursions may be less challenging to the sensory system they also require greater motor responses with respect to lower extremity strength and multijoint coordination [2, 16]. Accordingly, positive correlations between lower extremity strength and performance during these tests suggest that the ability to detect underlying sensorimotor deficits may be limited during these more dynamic tasks [2, 17].

While balance tests are thought to provide insight into sensorimotor processing, it is difficult to test these mechanisms in isolation during traditional balance tests. Therefore, we introduce the Lower Extremity Dexterity (LED) test, which has been proven to quantify sensorimotor processing to control instabilities while controlling for the confounding factors of strength and whole-body equilibrium [18, 19]. The test is based on the principles of the upper extremity Strength-Dexterity (SD) test, which is a repeatable and informative paradigm that has successfully quantified differences in finger dexterity attributed to age, sex, and numerous clinical impairments [18, 20–23]. The SD test quantifies sensorimotor processing for dynamic finger function because it is independent of strength [21, 24] and engages distinct cortico-striatal-cerebellar networks in a context-sensitive way [25, 26]. Building on this paradigm, the LED test quantifies the ability of the isolated lower limb to dynamically stabilize an unstable interface with the ground by controlling the force vectors and motions of the foot [18, 19]. Performance of the LED test is a measure of lower extremity sensorimotor processing that is independent of strength [21], predictive of agility performance in soccer athletes [27], and informative of age- and sex-related effects [18, 28]. Understanding the relationships between LED test and clinical outcome measures can provide insight into the sensitivity of these measures for detecting sensorimotor deficits. Moreover, considering the LED test together with outcome measures will help elucidate how sensorimotor processing contributes to balance.

It stands to reason that balance likely requires a combination of strength, multijoint coordination, and sensorimotor processing that are quantified to varying degrees using numerous outcome measures, several of which are described above. Given the varying demands across tests, it is likely that traditional balance tests provide useful, although test-specific, information regarding the contributors to balance. However, the extent to which these factors contribute to balance and how these outcome measures relate to them are not clear. Therefore, the purpose of this study was to determine the relationships and hierarchy among these outcome measures for balance, strength, and sensorimotor processing in healthy and active young adults.

2. Materials and Methods

Thirty-seven young adults (18 F, 19 M) between the ages of 18 and 30 years (mean ± standard deviation; age: 24.7 ± 2.7 yrs; body mass: 74.4 ± 14.2 kg; height: 1.8 ± 0.1 m) and engaged in recreational sports activities agreed to participate in this study. Participants were excluded if they had (1) any lower extremity injury or surgery within the last 12 months, (2) a current upper or lower extremity injury with persistent pain and/or inability to fully participate in sport, (3) a concurrent pathology or morphology that can cause pain or discomfort during physical activity, or (4) any physical, cognitive, or other condition that would impair their ability to perform the tasks proposed in this study. Prior to participation, testing procedures were explained to the participants and informed consent was obtained as approved by the Institutional Review Board of the University of Southern California Health Sciences Campus. Testing was conducted in the Division of Biokinesiology and Physical Therapy's Human Performance Laboratory located in the Competitive Athlete Training Zone, Pasadena, CA.

2.1. Procedures. Participants attended a single session during which anthropometric measurements (height, weight, and leg length) were collected and foot dominance was self-selected based on participant response to which foot they preferred to kick a ball for maximal distance. Each group completed the following battery of tests, described in detail below, in random order: LED, SLB, SLHB, and YBT. In addition, individuals performed the Vertical Jump (VJ) test to assess lower extremity strength and power.

2.2. Instrumentation. Reflective kinematic markers were placed on the skin over the sacrum and bilaterally on the participant's shoes at the positions best projecting the anatomical landmarks of heel and toe. Three-dimensional motion analysis was performed using a marker-based, 11-camera digital motion capturing system (250 Hz; Qualisys, Gothenburg, Sweden). Ground reaction force (GRF) data were obtained using a 1.20 × 0.60 m force plate (1500 Hz; AMTI, Newton, MA, USA) embedded into the floor surface. These data were collected synchronously using motion capture software (Qualisys Track Manger, v2.6, Gothenburg, Sweden) during the VJ and SLHB tests. The LED test system consisted of a helical compression spring (Century Springs Corp., Los Angeles, CA) mounted on a single-axis force sensor (Transducer Techniques, Temecula, CA) on a stable base with a platform affixed to the free end. The vertical component of the GRF was sampled with a data acquisition

system (2000 Hz; Measurement Computing, Norton, MA) and recorded and displayed in real time with custom software.

2.3. Vertical Jump Test. Participants were instructed to stand adjacent to a Vertec Jump Measurement device (Sports Imports, Hilliard, OH) (positioned on the same side of their self-reported dominant hand) with their feet on the force plate shoulder width apart. After squatting to a comfortable position, they were instructed to perform a maximal vertical jump. Participants were allowed to use their arms to augment performance and they were asked to use the dominant hand to displace the highest possible horizontal swivel vane to encourage maximum jump height. Power was calculated as the product of the vertical GRF and the vertical velocity of the reflective marker placed over their sacrum using BTS SMART-Analyzer software (BTS Bioengineering, Milan, Italy). The outcome measure, peak power (W/kg, normalized to body mass (BM)), was identified for each trial and averaged across three trials for analysis.

2.4. Y-Balance Test. The YBT, a simplified version of the Star Excursion Balance Test, is a reliable measure of dynamic balance featuring the anterior, posterior-medial (PM), and posterior-lateral (PL) components [3]. The anterior direction is defined as directly in front of the participant and the PM and PL directions are located 135 degrees from the anterior direction, separated by 45 degrees, making the "Y" shape described in the name [3]. Participants were asked to stand and maintain balance on their dominant leg and reach as far as possible with the free limb in each direction initiating from the start position. Participants performed three trials in each direction with 40 seconds of rest between reach directions. Trials were terminated early if a participant (1) failed to maintain single-leg balance, (2) used the free limb for stance support, or (3) failed to return to the start position. Participants were provided with a visual demonstration prior to testing and tested in the following order: anterior and then PL and then PM. As the outcome measure, average distances reached in each direction as a percent of leg length (LL) were considered dependent variables for analysis (YBT_A, YBT_{PL}, and YBT_{PM}, resp.). LL was measured in standing with a tape measure from the left greater trochanter to the floor.

2.5. Single Limb Hop and Balance Test. During the SLHB, upon verbal command, participants performed a single limb forward hop of a distance (normalized to their LL) with their dominant leg while their arms were folded across their chest. Upon landing, they were instructed to maintain single limb standing balance with arms still folded across their chest. In accordance with several groups [11, 13], the outcome measures, COP variability in the medial-lateral (ML) and anterior-posterior (AP) directions, COP_{ML} and COP_{AP}, respectively, were considered dependent variables for analysis. COP excursion measurements are representative of body sway and provide information about the ability of motor system to control the COM. While all humans exhibit some level of body sway as measured by COP variability, greater COP variability has been linked to instability and falls [29, 30]. As with the previous tests, the average across three trials was used to indicate performance level.

2.6. Single Limb Balance Test. During the SLB, participants were asked to maintain balance on their dominant leg with their arms folded across their chest and eyes closed for a total of 15 seconds. Participants were positioned on a force plate and upon verbal command asked to lift their nondominant foot off the floor (knee bent at approximately 60°) and close their eyes. Trials were terminated early upon ground contact with the nondominant limb or when participants opened their eyes. As with the SLHB, the mean of the three trials was reported and the outcome measures of COP variability in the ML and AP directions were considered dependent variables for analysis.

2.7. Lower Extremity Dexterity Test. A detailed description of LED test methodology is provided in prior publications [18, 19, 27, 28]; therefore, only a brief description is provided here. Participants were positioned in an upright partially seated posture on a bicycle saddle intended to stabilize the body and minimize extraneous use of the contralateral limb and upper extremities during testing. Visual feedback was provided via computer monitor and participants were instructed to slowly compress the spring with their foot with the goal to raise the force feedback reference line as high as possible and maintain that maximal level of compression for at least ten seconds [18, 19, 27, 28]. After familiarization, at least 10 trials were performed on the self-reported dominant limb. The outcome variables, mean compression force (LED_F) and a measure of force variability defined by the root-mean square (RMS) of the force signal during the steady-state hold (LED_{RMS}), were processed using custom Matlab software (v2013b, MathWorks, Natick, MA) and were considered dependent variables for analysis.

2.8. Statistical Analysis. This study considered five tests and 10 total outcome measures as dependent variables detailed above: YBT (3), SLHB (2), SLB (2), LED (2), and VJ (1). Principal components analysis (PCA) was performed to identify the best linear fit to the data using a series of perpendicular vectors or principal components (PCs) [31]. Within each PC vector (i.e., column), the structure of the correlations and nonzero numerical values in each column quantify the relative positive or negative correlations among variables [31]. To put it simply, we used PCA as a method of examining the contributions of the outcomes measures to balance and the associations among the outcome measures. Due to the differences in units and normal distributions among variables, and for comparison purposes, we calculated the standard score (z-score) of each variable and used their standardized normal distribution values as the PCA dataset [32]. The PCs are presented in descending order quantifying their contributions to balance such that the first principal component explained the largest amount of variance. We note that the first five PCs captured at least 80% of the total variance; therefore, we limited our analysis to them: first five PCs. SPSS version 22 (IBM, Armonk, NY) and Matlab were

Table 1: Mean performance data from all subjects.

Metric	Variable	Mean ± SD
VJ	Power (W/kg, % BM)	48.1 ± 9.6
YBT	YBT_A (% LL)	63.4 ± 4.8
YBT	YBT_{PM} (% LL)	106.6 ± 11.3
YBT	YBT_{PL} (% LL)	102.4 ± 10.1
SLHB	COP_{ML} (mm/s)	0.03 ± 0.01
SLHB	COP_{AP} (mm/s)	0.03 ± 0.01
SLB	COP_{ML} (mm/s)	0.02 ± 0.01
SLB	COP_{AP} (mm/s)	0.01 ± 0.003
LED	LED_F (N)	130.7 ± 13.4
LED	LED_{RMS} (N/s)	0.08 ± 0.03

Table 2: Principle component loadings.

Variable	1st PC	2nd PC	3rd PC	4th PC
VJ	*0.67*	−0.03	*0.60*	−0.54
YBT_A	*0.62*	0.07	−0.52	−0.15
YBT_{PM}	*0.80*	−0.50	0.40	0.41
YBT_{PL}	**1.00**	−0.06	0.23	0.04
SLHS COP_{ML}	−0.19	**1.00**	*0.87*	0.03
SLHS COP_{AP}	−0.18	*0.86*	**1.00**	0.20
SLS COP_{ML}	*0.61*	*0.86*	*−0.70*	0.04
SLS COP_{AP}	*0.68*	*0.80*	*−0.66*	0.17
LED_F	0.52	−0.37	*0.60*	*0.94*
LED_{RMS}	−0.50	0.18	−0.57	**1.00**
% Contribution	**26.07%**	**23.53%**	**14.57%**	**10.49%**
Cumulative	**26.07%**	**49.59%**	**64.17%**	**74.66%**

Normalized loadings for ease of comparison; italics font in each column indicates (≥0.60) positive and negative correlations, respectively, with the dominant variable in bold.

used for this analysis and the significance level was set at $p \leq 0.05$.

3. Results

The means and standard deviations of all dependent variables are presented in Table 1. Outcome measures on all of the tests, by all subjects, were within normal ranges when compared to previously published data [3, 12, 18, 33, 34]. Our PCA data are presented in numerical form (Table 2). Loading values quantify the strength and direction of the relationships between variables and range between −1 and 1, where 1 is total positive correlation, 0 is no correlation, and −1 is total negative correlation.

The 1st PC explained 26% of the total variance in balance with the highest loadings assigned to YBT_{PL} and YBT_{PM} (1.00 and 0.80, resp.). Furthermore, we report additional moderate, positive correlations between VJ, YBT_A, and SLB COP_{AP} and COP_{ML} with loading values ranging from 0.68 to 0.61. The 2nd PC explained an additional 24% of the variance with all SLHB and SLB COP variables exhibiting the highest loadings (1.00–0.80, resp.). In the 3rd PC, the SLHB COP measures featured the highest loadings, explaining 14% of the variance. Interestingly, while the relationships between SLHB and SLB COP variables were moderate to strong in both the 2nd and 3rd PCs, they were negatively correlated in the 3rd PC (−0.62 and −0.59), unlike the 2nd, which featured positive correlations. In addition to the disambiguation between static (SLB) and dynamic (SLHB) balance variables we report in the 3rd PC, we further note that LED_F showed a moderate positive association with SLHB variables while LED_{RMS} was positively correlated with SLB variability. We further report moderate positive correlations with VJ and LED_F. The 4th PC explained an additional 11% of the variance in balance and revealed that the LED variables were highly positively correlated (1.00 and 0.94, resp.) with each other and no other metric. Finally, YBT_A solely dominated the 5 PCs and explained 9% of the total variance. In order to further highlight our results, we provided a visual representation of the respective loadings for each of the first five PCs, first presented in Table 2, in Figure 1.

4. Discussion

This is the first study, to our knowledge, to investigate the relationship among multiple balance tests and outcome measures traditionally used to assess balance in young individuals. The battery of measures examined in this study represent a range of static and dynamic tests that are commonly used to assess balance in healthy individuals or following lower extremity injury or to identify those at greater risk for injury [1, 3, 5–7, 9, 12, 14, 29, 30, 35, 36]. The combination of measures of static and dynamic balance, strength, and sensorimotor processing considered in this study allowed the unique opportunity to explore the relationships between the numerous components that we speculate to contribute to overall balance. Understanding the relationships and hierarchy among outcome measures in young healthy individuals using PCA provides some insight into the contributors to balance. In this paper, we present our PCA data in two distinct formats, numerically (Table 2) and graphically (Figure 1). For ease of comparison, we ordered the measures on a continuum from what can be considered more dynamic (YBT) to more static (SLB) balance tests anchored at the extremes by the outcome measures most associated with strength (VJ) and sensorimotor processing (LED) (top to bottom, Tables 1 and 2; left to right, Figure 1). When considered together, 84% of the variance in balance was explained by the first five PCs with each individually contributing to 9–26% of the total variance. The 6th and further PCs each contribute to relatively small percentages (<9%) of total variance and were not considered in our analysis due to the potential for overinterpretation.

Our analysis indicated that balance is best distinguished by a combination of outcome measures from both static and dynamic tests as the SLB and Y-balance tests were the most heavily loaded in the 1st PC. Together, these measures explained 26% of the total variance in balance. YBT_{PL} featured the highest loading and revealed strong and moderate positive relationships with YBT_{PM} and YBT_A, respectively. Multiple studies have reported correlations between lower limb strength [2, 17], range of motion [37, 38], and Y-balance performance in all three directions. Therefore, it is not surprising that there was also a moderate positive

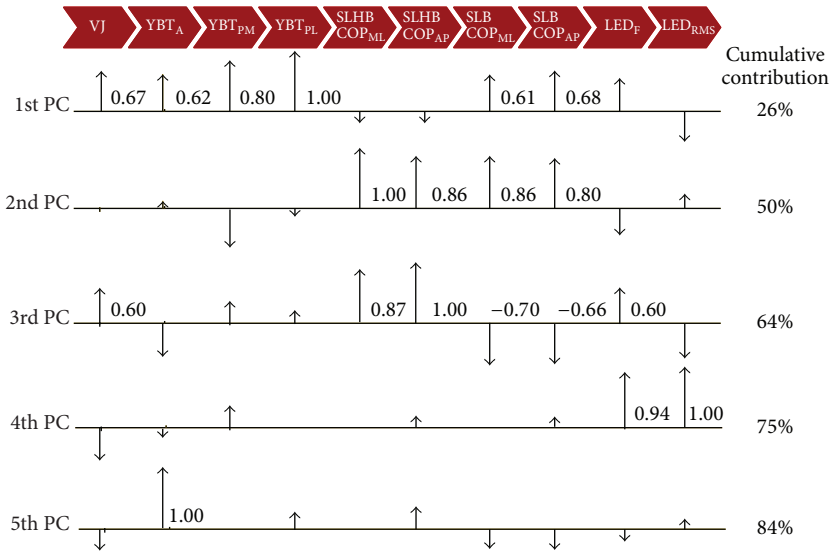

Figure 1: Visualization of PC loadings. The scaled metric loadings for the first five PCs are illustrated above. All loadings are shown, but numerical values are only listed if they are ≥±0.60. The signs of the loadings are indicated by the direction of the arrowheads.

correlation with VJ, a widely accepted estimate of leg power and strength [33, 39, 40]. The inclusion of these measures in the 1st PC suggests that the multijoint coordination and strength required to perform more dynamic tests are important contributors to balance. However, the presence of moderate positive correlations with SLB variability (COP_{ML} and COP_{AP}), the most static balance test, suggests that the detection and correction of smaller perturbations are also important to balance ability. Measurements of COP variability during SLB tests are validated methods of quantifying what is referred to as static balance or stability [1, 29, 34]. Relatively small displacements of the lower limb, particularly at the ankle, are used to maintain balance and are reflected in COP variability [15]. The presence of the SLB variables in the 1st PC seems to indicate moderate dependence on sensory inputs for detection of small perturbations while maintaining balance.

After considering the contribution of these measures to balance, an additional 24% of the variance was explained by grouping of COP variables during both SLHB and SLB in the 2nd PC. It is not surprising that these variables were strongly associated as both are measures of COP variability, which are representative of modulation of ML and AP COP by the motor system. While the mean values for SLHB variability were slightly, although, we emphasize not significantly, greater than the SLB (Table 1), we concede that is due to the more dynamic nature and slightly increased strength demands of the SLHB. When taken together, however, the correlations among the outcome measures from static and dynamic balance tasks support prior research that reported no differences performance on both static and dynamic postural control tasks [29]. Strong positive correlations among these variables suggest that both small and large corrective actions during static and dynamic tests are important overall contributors to balance. Moreover, the negative correlation to YBT_{PM} supports our speculation that COP variables are indicative of separate contributions to balance compared with what is measured during more dynamic, multijoint coordination-, and strength-driven tasks.

In the 3rd PC, which further explained 13% of the total variance, COP velocities in the AP and ML directions during the SLHB were again the leading contributors. Interestingly, in this PC, SLHB measures were moderately negatively correlated with SLB measures, unlike the 2nd PC. The contrasting relationships between COP variables during SLB and SLHB observed between the PCs, as well as the slight differences in mean performance values presented in Table 1, support the notion that COP variability in these two tasks represents similar but distinct mechanisms of balance [1, 4, 12, 14, 30, 36, 41]. The SLHB is a standard objective measure often used to evaluate dynamic balance following training protocols and when examining patients following lower limb injury or surgery [1, 7, 9, 13]. While static balance measures are of clinical relevance, in terms of function, emphasis is often placed on dynamic balance tests (e.g., SLHB and YBT) because they are more representative of activities of daily living (ADLs) and have greater sensorimotor demands. To limit the potential influence of strength and distance hopped on performance of this test, we asked participants to hop a standardized distance equal to the length of their lower limb. The characterization of the SLHB as a more dynamic measure of balance than the SLB is further supported by the moderate positive relationship with VJ. Moreover, the weak and discordant relationship with YBT variables could support the argument that the SLHB is less dynamic than the Y-balance protocol and results in smaller perturbations to the COM within the BOS.

We find it particularly noteworthy that, in the 3rd PC, LED *compression force* (LED_F) was positively correlated with *dynamic* balance variables (SLHB) while LED *force variability* (LED_{RMS}) was more closely associated with *static* balance variables (SLB). The dependent variable for the LED test

has traditionally been the average of the three hold phases with the highest mean compression force (LED_F). This is because the spring becomes increasingly unstable as it is compressed further. Thus, the level of maximal sustained spring compression is informative of the maximal instability that can be controlled by the isolated leg. The springs are designed to reach these high levels of instabilities at very low forces (ca. 150 N for the leg or ca. 10% of body weight). The LED_F has shown to be sensitive to sex differences [18, 28] and age effects [18] and correlate well with whole-body agility [27]. More recently, LED_F has shown strong correlations with single limb cross-country ski distance, which one can easily argue is a *dynamic* measure, but showed no correlation with a *static* single limb balance test [42]. Additionally, the force fluctuations (e.g., RMS) during the hold phases of the SD paradigm for the upper extremity were first introduced as a method of quantifying differences in performance (i.e., sensorimotor processing) attributed to several clinical conditions [18, 22, 23]. Greater RMS indicates larger dynamical dispersion and suggests weaker (or looser) corrective actions by the neuromuscular controller enforcing the sustained compression. Now, in this study, we include force fluctuations during the LED test (LED_{RMS}) as a complementary, but equally important, measure of sensorimotor processing of the lower limb in healthy individuals.

The 4th PC accounted for 11% of the total variance in balance. Strong and positive relationships between both LED variables (LED_F and LED_{RMS}) were noted in this PC, suggesting that the sensorimotor control may uniquely contribute to balance. These results complement previous studies, including numerous of our own featuring the SD paradigm for the fingers, which have found that sensorimotor processing during dexterous tasks (e.g., dexterity) represents a different functional domain than strength or whole-arm coordination [18–21, 24–27, 43, 44]. While no correlations greater than 0.60 were noted with variables of other tests in this PC, LED variables were negatively correlated to VJ (−0.54), a measure of lower extremity strength and power, which further complements our prior work suggesting that lower extremity dexterity is independent of strength [19]. In the 5th PC, YBT_A was the sole contributing variable to the 9% of the total variance explained. While the relative contribution to overall variance explained is comparatively small, the fact that YBT_A shows no correlation with the other YBT variables implies it may represent a different functional dimension than the posterior YBT directions. The anterior direction can be considered primarily uniplanar, whereas the PM and PL directions clearly require coordination of multiple joints across multiple planes. This is also supported by the data in the 1st PC that show strong correlations between the YBT PM and PL directions and only a moderate correlation with the anterior direction and again in the 3rd PC, where YBT_A shows weak negative correlations with the YBT posterior directions.

The results presented in this study speak to the fact that balance is dependent on multiple contributors. We find that the outcome measures of tests can be thought of as quantifying the strength, multijoint coordination, dynamic and static stability, and sensorimotor processing contributors to balance—which we find cannot be assessed independently and simultaneously by any one single outcome measure. This makes it difficult to truly understand the sensorimotor mechanisms of balance, let alone the effects of lower extremity injury on balance ability. This may begin to explain why there are conflicting reports of effects of injury on outcome measures of balance tests or effectiveness of training or rehabilitation protocols for improving these measures. For example, while several studies report differences between control and clinical groups in some or all measures associated with SLB tests [5, 14, 15, 17, 30], others report no differences between or within groups. Previous authors suggest that the inconsistent reports may be attributed to the fact that the SLB test loses sensitivity over the time course of recovery and is not challenging enough to be truly representative of sports-related activities, where balance deficits become more apparent [37, 45, 46]. There are also similar conflicting reports across more dynamic balance tests including the YBT. Multiple groups have reported significant differences between side-to-side YBT outcome measures (e.g., functional reach distances) in participants with chronic ankle instability (CAI) [37, 45]. However, one reported side-to-side differences in participants with CAI, but no group differences between healthy participants and those with CAI [45]. The inconsistencies in the literature in terms of success of both static and dynamic balance tests in the clinic support our hypothesis that these measures provide informative, yet limited, information about the mechanisms of balance ability. It is important to point out that our study was conducted in recreationally active young adults with no recent lower extremity injuries. Our results compel future studies in clinical populations to develop and assess the ability of outcome measures to gauge the efficacy of rehabilitation regimens for lower extremity injuries, including, but not limited to, CAI and ACL tears.

We successfully identified distinct relationships among outcome measures that suggest they together reveal latent functional contributors to balance. After considering the origin, nature, and use of each outcome measure, we propose that the latent contributors to balance they reveal are those of strength, multijoint coordination, and sensorimotor processing. They represent distinct functional domains, which are revealed by the relationships among the loadings in our PCA results. The multiple strong to moderate correlations (loadings) in the 1st PC suggest that a combination of strength, multijoint coordination, and static stability (i.e., detection of small perturbations from the sensory system) is the leading contributors to balance. However, in the subsequent PCs, other contributors gain prominence. The 2nd PC placed strong emphasis on a combination of static and dynamic balance variability. The fact that they are not strongly correlated with the other outcome measures strengthens our assertion that both static and dynamic balance are similar functional features that are distinct from strength or multijoint coordination. These results indicate that the combined corrective actions by the motor system during both static and dynamic balance tests are important contributors to balance. While the SLB and SLHB tests have similar origins and functional features, there are differences

that warrant consideration. The more dynamic nature of the SLHB naturally leads one to assume that there would be different strength and coordination requirements, which is supported by the negative correlations with SLB variables and positive correlation with VJ revealed in the 3rd PC. The opposite loading signs of the SLHB in the 2nd and 3rd PCs speak to the fact that it may be informative of both static and dynamic balance, but the moderate correlation with VJ emphasizes that dynamic stability should be considered in the context of submaximal force performance to reduce the influence of strength, which, as we mentioned previously, can dilute the information gleaned from such dynamic outcome measures. Additionally, the correlations we report between the LED test variables and COP variability during both SLB and SLHB indicate that the LED test may be a useful tool to quantify sensorimotor processing during both static and dynamic balance measures. Finally, our analysis further indicated that sensorimotor processing, as quantified by the LED test, was another distinct contributor to balance (4th PC) that also tended to be independent of strength. This confirms our prior work for both the upper and lower extremity [18–20, 24, 27, 28, 43, 44] and mirrors work about the development of dexterity in children where the SD test was seen as a functional dimension distinct from strength and whole-arm coordination [26]. These results in lower extremity function also complement our findings in the upper extremity [47] despite the obvious evolutionary, anatomical, and functional differences and suggest fundamental, body-wide mechanisms for function. We do acknowledge, however, that sensory or motor constructs (e.g., proprioception, vision, and motor control) were not specifically quantified in this study. We also note that these data represent balance ability in healthy individuals. It is not clear how these results would change if individuals with sensory or motor deficits were included.

Our results support the well-accepted notion that balance is a complex, albeit everyday, task but provide a quantitative context within which to understand its contributors. Thus, we lend evidence to the idea that depending on a single outcome measure to quantify balance, its deficits, and its rehabilitation is arguably deficient. We recommend using a combination of complementary assessments to quantify its multiple contributors: strength, multijoint coordination, stability (both static and dynamic), and sensorimotor processing. This will not only improve assessment accuracy on an individual level, but also facilitate the development of customized rehabilitation or training regimens to target improvements of individual contributors deemed deficient or in most need of attention. Furthermore, the ability of the novel LED paradigm to successfully quantify sensorimotor processing, in addition to the correlations with both static and dynamic balance measures reported in this study, makes it a useful tool to quantify and promote that specific contributor. Thus, it complements the other well-accepted measures of strength and multijoint coordination currently in use in both research and clinical settings. Note that because the LED test requires very low forces and tests the isolated leg while the hip and torso are held steady, it is particularly well suited to clinical, postsurgical, and postinjury populations who cannot perform other outcome measures mostly geared towards healthy athletic young adults.

Outcome Measure Abbreviations

VJ: Vertical Jump test
YBT: Y-Balance test
SLHB: Single Limb Hop and Balance test
SLB: Single Limb Balance test
LED: Lower Extremity Dexterity test.

Disclaimer

The content is solely the responsibility of the authors and does not necessarily represent the official views of the NIH.

Conflict of Interests

Francisco J. Valero-Cuevas holds US Patent no. 6,537,075 on some of the technology used but has no active licensing agreements with any commercial entity. None of the other authors have any financial or personal relationships with other people or organizations that could inappropriately influence this work.

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Review Article

Physiological and Neural Adaptations to Eccentric Exercise: Mechanisms and Considerations for Training

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Eccentric exercise is characterized by initial unfavorable effects such as subcellular muscle damage, pain, reduced fiber excitability, and initial muscle weakness. However, stretch combined with overload, as in eccentric contractions, is an effective stimulus for inducing physiological and neural adaptations to training. Eccentric exercise-induced adaptations include muscle hypertrophy, increased cortical activity, and changes in motor unit behavior, all of which contribute to improved muscle function. In this brief review, neuromuscular adaptations to different forms of exercise are reviewed, the positive training effects of eccentric exercise are presented, and the implications for training are considered.

1. Introduction

Neuromuscular and functional changes induced by exercise are specific to the mode of exercise performed. The degree of mechanical tension, subcellular damage, and metabolic stress can all play a role in exercise-induced muscle adaptations [1–5]. Of the three types of muscle contractions that can be utilized during exercise (concentric, isometric, and eccentric), eccentric exercises are those actions in which the muscle lengthens under tension. During eccentric contractions the load on the muscle is greater than the force developed by the muscle and the muscle is stretched, producing a lengthening contraction. Eccentric exercise is characterized by muscle microlesions and greater mechanical tension as compared to concentric/isometric contractions and therefore may result in greater muscle adaptations. Although all forms of exercise may induce impressive muscle adaptation, it is not always clear which method is best for maximizing adaptation gains. This paper provides a brief overview of studies documenting physiological (metabolic, histochemical) and neural adaptations in response to exercise training, with an emphasis on eccentric exercise.

2. Exercise Training and Physiological Adaptations

High intensity resistance training is associated with significant physiological adaptations within skeletal muscle [6] including changes in the contractile and/or noncontractile elements of muscle. When mechanical overload of muscle occurs, the myofibers and extracellular matrix are disturbed, which in turn stimulates a process of protein synthesis [7]. Mechanical tension induced by high intensity exercise can also increase the rate of metabolic stress and stimulate subcellular pathways involved in protein synthesis such as the mitogen-activated protein kinase pathway, which may

play a role in exercise-induced muscle growth [1, 2]. The total number of sarcomeres in parallel and in series increase resulting in an increase in fascicle length and pennation angle and, consequently, muscle hypertrophy. It has been proposed that stretch combined with overload is the most effective stimulus for promoting muscle growth [8, 9]. During eccentric exercise, skeletal muscle is subjected to both stretch and overload which triggers subcellular damage to the contractile and structural components of skeletal muscle [10, 11]. This subcellular damage induces a sequence of physiological events including the activation of master signaling pathways for gene expression and muscle hypertrophy [1, 8, 10]. Notwithstanding, mechanotransduction (exercise-induced mechanical stimuli) may be the primary mechanism associated with muscle hypertrophy in healthy muscle. This is demonstrated by an increase in the number of sarcomeres in the absence of fiber necrosis following exercise-induced muscle tension [12]. Skeletal muscles sense mechanical information and convert this stimulus into the biochemical events that regulate the rate of protein synthesis. However, since eccentric contractions induce greater mechanical tension on the muscle fibers than concentric exercise, this form of exercise induces a more rapid addition of sarcomeres in series and in parallel as inferred from the increase in muscle cross sectional area (CSA) and pennation angle [13]. Previous studies reported an increase of fiber length in muscles subjected to chronic eccentric work [14], whereas a decrease [14] or a lack of change [15] of fiber length was shown in muscles worked concentrically. Greater muscle hypertrophy following high intensity eccentric exercise was also associated with larger fiber pennation angle [15]. These results indicate that the mechanical stimuli induced by high intensity exercise may be a primary mechanism for muscle hypertrophy. Hortobágyi et al. [16] also observed that muscle mass recovery after immobilization was greatest following eccentric exercise compared to concentric and isometric training, most likely due to the greater mechanical tension produced during eccentric exercise [17]. Similarly, other studies demonstrated that high tension eccentric exercise is more effective than concentric exercise in increasing muscle mass, through changes in histochemical characteristics and metabolic substrates within the skeletal muscle [18].

2.1. Eccentric Exercise and Histochemical Adaptations. The mechanisms underlying the hypertrophic response to exercise may include changes in the hormonal milieu, cell swelling, free-radical production, and increased activity of growth-oriented transcription factors [6, 7]. Mechanical tension, produced by force generation and stretch, is an essential factor to stimulate signaling pathways involved in muscle growth, and the combination of these stimuli appears to have a marked additive effect [9, 19, 20]. Mechanical stimuli can regulate the rate of protein synthesis through changes in binding of a ribosome to the mRNA and/or by modifications in methylguanosine, which in turn encodes proteins that are central to the growth process [21]. Mechanical stimuli may also contribute to muscle hypertrophy through changes in muscle fiber membrane permeability to calcium ions [22]. The increased calcium concentrations within the cytosol of the muscle cell increase the rate of protein synthesis in skeletal muscle [23]. Moreover, titin is a site for calcium binding and is ideally positioned in the muscle sarcomere to sense mechanical stimuli and transform them into biochemical signals, capable of altering sarcomere number and optimal tension during lengthening contractions [24, 25].

During eccentric exercise the contracting muscle is forcibly stretched, producing a higher mechanical tension and muscle microlesions. Mitogen-activated protein kinase is a master signaling pathway for gene expression and muscle hypertrophy [26] and is considered to be the most responsive to mechanical tension and subcellular muscle damage [1]. Mitogen-activated protein kinase links cellular stress with an adaptive response in myocytes, modifying growth and differentiation [7, 27]. Insulin-like growth factor is also considered to be a key factor for muscle hypertrophy and shows enhanced effects in response to mechanical loading [28, 29]. Insulin-like growth factor contributes to muscle hypertrophy through a mechanical response of IGF-1Ea isoform to exercise training and appears to be activated by mechanical signals and subcellular muscle damage [28, 30]. Mechanical stimulation may cause the IGF-1 gene to be spliced toward IGF-1Ea isoform which in turn increases IGF-IEa mRNA expression [31] and muscle hypertrophy [32].

Muscle hypertrophy following eccentric exercise may also be explained by other tension-sensitive anabolic pathways. For example, the effects of testosterone on muscle hypertrophy are enhanced by mechanical loading, either directly by increasing the rate of protein synthesis and inhibiting protein breakdown [33] and/or indirectly by stimulating the release of other anabolic hormones such as Growth hormone [34]. Bamman et al. [35] reported that high intensity eccentric exercise upregulated androgen receptor content in humans and modulation of androgen receptor content appears to occur predominantly in fast-twitch muscle fibers [36]. Accordingly, Ahtiainen et al. [37] reported significant correlations between training intensity, testosterone concentration, and muscle cross-sectional area, indicating that high intensity eccentric exercise-induced elevation in testosterone is an important contributor to muscle hypertrophy.

Growth hormone may contribute to muscle hypertrophy through both anabolic and catabolic processes. An increase in Growth hormone can enhance interaction with muscle cell receptors, facilitating fiber recovery and stimulating a hypertrophic response [38]. Other anabolic signaling pathways including calcium-dependent pathways have been implicated in the regulation of muscle hypertrophy [39].

2.2. Eccentric Exercise and Metabolic Adaptations. Mechanical tension produced by force generation and stretch contributes to muscle ischemia [8, 9] which can lead to metabolic adaptations within the skeletal muscle. During eccentric contractions, passive muscular tension develops because of lengthening of extramyofibrillar elements, especially collagen content in the extracellular matrix which can contribute to an increased acidic environment. Such an environment can contribute to increased fiber degradation and increased sympathetic nerve activity [7], facilitating an adaptive hypertrophic response [2]. Numerous studies indicate that anabolic

exercise induced metabolic stress can have a significant hypertrophic effect [2].

3. Exercise Training and Neural Adaptations

Neural adaptations to training can be defined as changes within the nervous system that allow a trainee to more fully activate prime movers in specific movements and to better coordinate the activation of all relevant muscles, thereby affecting a greater net force in the intended direction of movement [40]. Neural adaptations may occur at the level of the motor cortex, spinal cord, and/or neuromuscular junction following training [41–43]. Adaptations may also occur at excitation- contraction coupling pathways located distal to the neuromuscular junction. The neural adaptations observed following training explain the disproportionate increase in muscle force compared to muscle size during the initial stages of training. For instance, increased muscle activity, recorded with electromyography (EMG), has been observed during the early phase of strength training in association with significant gains in muscle strength, but in the absence of changes of muscle mass or changes in membrane characteristics within the skeletal muscle [44]. Early gains in strength have been attributed to a variety of mechanisms including increased maximal motor unit discharge rates [45, 46], increased incidence of brief interspike intervals (doublets) [47], and decreased interspike interval variability [48].

Numerous other studies have investigated neural adaptations following resistance training. Aagaard et al. [49] observed increases in evoked V-wave and H-reflex responses during maximal muscle contraction after resistance training indicating an enhanced neural drive in the corticospinal pathways and increased excitability of motor neurons. Furthermore, previous studies have demonstrated significant changes in motor unit discharge rate [46], muscle fiber conduction velocity [50], and rate of force development after resistance training [46, 51]. Collectively these studies show that increased strength following resistance training can be attributed to both supraspinal and spinal adaptations (i.e., increased central motor drive, elevated motoneuron excitability, and reduced presynaptic inhibition) [49].

The neural adaptions to resistance training are dependent on type of muscle contractions performed and the neural adaptations and improvement in muscle force vary depending on whether eccentric, concentric, or isometric contractions are executed [46, 52]. The section below focuses on the specific neural adaptations that have been observed with eccentric exercise.

3.1. Eccentric Exercise and Cortical Activity. It is well known that exercise can induce changes in cortical activity [53–55]. These changes can be measured with techniques such as electroencephalography (EEG) and neuroimaging techniques and studies applying these methods have demonstrated that variations in cortical activation patterns depend on exercise mode and intensity [41, 56]. This is perhaps not surprising given that the central nervous system employs a different neural strategy to control skeletal muscle during eccentric contractions versus isometric or concentric muscle contraction. This is evidenced, for example, by the preferential recruitment of fast twitch motor units and different activation levels among synergistic muscles during eccentric compared to concentric contractions [57–59]. Fang et al. [41] showed that cortical activities for movement preparation and execution were greater during eccentric than concentric tasks, most likely due to concurrent modulation (gating by presynaptic input) of the Ia afferent input from the lengthening muscle to reduce the unwanted stretch reflex and subcellular muscle damage [60]. Thus the brain probably plans and programs eccentric movements differently to concentric muscle tasks [41]. Moreover, neuroimaging studies have shown that cortical activities associated with the processing of feedback signals are larger during eccentric than concentric actions, likely due to the higher degree of movement complexity and/or stretch-related transcortical reflexes to control the stretched muscle [61, 62]. Additionally, earlier onset of cortical activation has been observed for eccentric versus concentric contractions [41] which has been attributed to the planning for more movement complexity, modulation of monosynaptic reflex excitability, or carrying out a different control strategy (e.g., motor unit recruitment) for an eccentric action [57, 61, 62].

3.2. Eccentric Exercise and Motor Unit Behavior. During a muscle contraction, the central nervous system controls the production of increased muscle force by either increasing motor unit firing rates and/or the recruitment of additional motor units. Numerous studies have investigated changes in motor unit firing rates after resistance training and have shown that the change in motor unit firing rate is dependent on the type of muscle contraction. Van Cutsem et al. [47] observed increased firing rates of motor units and a more frequent occurrence of short interspike intervals (doublets) following 12 weeks of dynamic contractions of the ankle dorsiflexors. Kamen and Knight [63] also found a 15% increase in motor unit firing rates following 6 weeks of dynamic training of the quadriceps muscles. Similarly, Vila-Chã et al. [45] reported a significant increase in firing rates of vasti motor units after six weeks of resistance training. However, other studies have reported no change in maximal motor unit firing rates following isometric resistance training of the abductor digiti minimi and quadriceps muscles despite a significant increase in absolute force [46, 64, 65]. These studies suggest that maximal motor unit firing rates increase in response to dynamic but not isometric resistance training. It has been proposed that stretch combined with overloading is the most effective stimulus for enhancing motor unit firing rates during dynamic resistance exercise. For instance, Dartnall et al. [66] showed ~40% decline in biceps brachii motor unit recruitment thresholds and 11% increase in minimum motor unit discharge rates immediately after and 24 h after eccentric exercise. Thus, more biceps brachii motor units were active at the same relative force after eccentric exercise.

A potential mechanism responsible for the increased muscle activation following eccentric training has been attributed to the neural regulatory pathways involved in

the excitation and inhibition process. During eccentric contractions, the spinal inflow from Golgi Ib afferents and joint afferents induce elevated presynaptic inhibition of muscle spindle Ia afferents, as demonstrated by reduced H-reflex responses and EMG amplitude during active eccentric versus concentric contractions [67, 68]. The removal of neural inhibition and the corresponding increase in maximal muscle force and rate of force development observed following eccentric resistance training could be caused by a downregulation of such inhibitory pathways, possibly by central descending pathways [69].

3.3. Eccentric Exercise and Muscle Force. Since greater maximum force can be developed during maximal eccentric muscle actions compared to concentric or isometric muscle actions, heavy-resistance training using eccentric muscle actions may be most effective for increasing muscle strength. Eccentric exercise may preferentially recruit fast twitch muscle fibers and perhaps the recruitment of previously inactive motor units [70]. This would lead to increased mechanical tension and as a consequence led to even greater force production [52].

Farthing and Chilibeck [52] reported that 8 weeks of eccentric resistance training resulted in greater muscle hypertrophy and muscle force than training with concentric contractions. In agreement, Kaminski et al. [69] also observed greater improvements in peak torque following eccentric (29%) compared to concentric (19%) training. It has also been shown that ballistic movement with stretch-shortening cycle muscle activation has the greatest effect on enhancing the rate of force development compared to concentric and isometric muscle contractions [71].

4. Considerations

Eccentric exercise is characterized by high force generation and low energy expenditure as compared to concentric and isometric exercises [72, 73] and therefore can be beneficial for clinical treatments. For example, eccentric exercise has been used in rehabilitation to manage a host of conditions including rehabilitation of tendinopathies, muscle strains, and anterior cruciate ligament (ACL) injuries [74, 75]. Although there are positive effects of eccentric exercise as reviewed above, it must be noted that there can also be detrimental effects. For instance, the nonuniform effect of eccentric exercise results in nonuniform changes in muscle activation [11], alternative muscle synergies [76] which may lead to strength imbalances. Studies have confirmed that intensive eccentric exercise may have a differential effect on different muscle regions [4, 5, 11, 77, 78] potentially resulting in an imbalance of muscle activity and alteration of the load distribution on joints. Eccentric exercise is also associated with muscle micro lesions, pain, reduced fiber excitability, and initial muscle weakness [4, 77, 79]. Furthermore, eccentric exercise may impair reflex activity which could lead to compromised joint stability during perturbations [43, 80]. Thus it is important to consider the initial unfavorable effects in addition to the long-term benefits.

5. Conclusion

Eccentric contractions are important to consider for training and rehabilitation programs because of their potential to produce large force with low metabolic cost. Data reported by several studies suggests that stretch combined with overloading, as in eccentric contractions, is the most effective stimulus for promoting muscle growth and enhancing the neural drive to muscle. This is evidenced by greater muscle hypertrophy, greater neural activity, and larger force production following eccentric exercise versus concentric and isometric exercise. Therefore, training that involves true maximal eccentric loadings could be more effective than concentric and isometric training for developing muscle growth and removing neural inhibition, leading to a significant improvement of muscle function.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Wheelchair Propulsion Biomechanics in Junior Basketball Players: A Method for the Evaluation of the Efficacy of a Specific Training Program

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As participation in wheelchair sports increases, the need of quantitative assessment of biomechanical performance indicators and of sports- and population-specific training protocols has become central. The present study focuses on junior wheelchair basketball and aims at (i) proposing a method to identify biomechanical performance indicators of wheelchair propulsion using an instrumented in-field test and (ii) developing a training program specific for the considered population and assessing its efficacy using the proposed method. Twelve athletes (10 M, 2 F, age = 17.1 ± 2.7 years, years of practice = 4.5 ± 1.8) equipped with wheelchair- and wrist-mounted inertial sensors performed a 20-metre sprint test. Biomechanical parameters related to propulsion timing, progression force, and coordination were estimated from the measured accelerations and used in a regression model where the time to complete the test was set as dependent variable. Force- and coordination-related parameters accounted for 80% of the dependent variable variance. Based on these results, a training program was designed and administered for three months to six of the athletes (the others acting as control group). The biomechanical indicators proved to be effective in providing additional information about the wheelchair propulsion technique with respect to the final test outcome and demonstrated the efficacy of the developed program.

1. Introduction

The benefits of practicing sports and physical activities for individuals with disabilities are generally accepted and concern psychosocial health and functional ability as well as general quality of life [1]. This is particularly the case for young people who, through sports practice, can positively reinterpret their role following a disabling injury and regain self-esteem as well as social integration [2, 3]. Participation in sports activities for young people with disability has remarkably increased over the past years and wheelchair basketball, in particular, has become one of the most popular disciplines among wheelchair-based sports [4].

As a consequence of the increase in participation, competitiveness has risen and the development of training protocols which are specific to the discipline and to the population has become more and more important. In this respect, quantitative biomechanical assessment of performance- and injury-related parameters provides information related not only to the overall outcome (product) of the analysed motor task, but also to the way this task is performed (process).

Extensive literature exists which analyzes biomechanical aspects of wheelchair basketball. In particular, the following areas have been widely investigated: wheelchair propulsion technique [5, 6]; the influence on propulsion mechanical efficiency of different parameters such as push frequency [7], push synchrony [7, 8], push symmetry [9], upper limb kinematics [10], and forces exchanged between the hands and the wheelchair hand rims [11]; the optimal release conditions and corresponding arm movement patterns in the free throw [12]; and the optimisation of the wheelchair configuration and ergonomics [13]. All these studies focused on adult wheelchair athletes and although it has been suggested that differences exist in the biomechanics of wheelchair propulsion between adults and children [14], no information is available about junior wheelchair basketball players. In addition, most of the above-mentioned studies were performed in a laboratory setting, using either an ergometer or a treadmill. However, the laboratory-based approach lacks providing a valid representation of the kinematics and kinetics characterising over-ground propulsion [15]. Therefore, in order to supply coaches with the most meaningful information, it is crucial to develop methods and tools able to provide accurate and reliable data in realistic field-based environments [6, 16].

To this aim, wearable inertial measurement units (IMUs) have been proposed as a valid alternative to traditional laboratory-based instruments, allowing for in-the-field performance monitoring without neither constraining athletes' movements nor significantly modifying the original wheelchair configuration. Few studies focused on the use of IMUs to assess biomechanical parameters in wheelchair sports, such as speed [17], wheelchair displacement and orientation [18], and upper body kinematics [19]. However, IMUs have never been used to characterise wheelchair propulsion in junior wheelchair basketball players nor have been flanked to traditional, or *ad hoc* developed, training programs adopted by wheelchair sports coaches.

Concerning training methodologies, only few studies focused on the development and assessment of training programs specific for wheelchair users. One study [20] reported that strengthening, stretching, and aerobic exercises improved propulsion biomechanical economy in terms of both kinematic and kinetic parameters without increasing shoulder and elbow stresses. The same study, however, included only adult wheelchair users without any previous sports experience. In another study [21], the effects of eccentric exercises on muscle soreness and on shooting percentages were investigated in wheelchair basketball players, and it was proven that muscle soreness induced by eccentric training negatively affects the upper limbs motor performance, at least 48 hours after exercise. Again, only adult athletes were included in the work. Only one study [22] focused on paediatric wheelchair users (ranging in age from 4 to 16 years) and reported that, after 8 weeks of resistance training, the performance of the 12-minute test significantly improved. The participants of the study, however, did not have any previous sports experience. Therefore, the conclusions of the above-mentioned works can hardly be extended to the field of junior wheelchair basketball. Nevertheless, the need to identify which critical factors should be taken into account in the development of training programs that are specific for the discipline and, particularly, the target population has been widely acknowledged [2, 3].

In this framework, the purpose of the present study was twofold: (i) to propose a method based on inertial sensing technologies aimed at obtaining a set of biomechanical parameters able to provide performance-related information about in-the-field wheelchair propulsion and (ii) to develop a discipline- and population-specific training program for junior wheelchair basketball athletes and to assess its efficacy using the proposed biomechanical parameters.

2. Methods

2.1. Participants. Twelve junior wheelchair basketball players (Table 1) were included in the study after having provided written informed consent. The participants were recruited from the Santa Lucia Foundation junior wheelchair basketball team and had at least two years of previous wheelchair basketball experience. Medical examination identified four athletes with paraplegia, three with myelomeningocele, two with poliomyelitis, one with spastic diplegia, one with a below-knee amputation, and one with a knee arthroprosthesis due to bone cancer. All participants were right-handed. Criterion for exclusion was presence of any medical contraindications which could prevent the athletes from regularly attending the training sessions over the season. The protocol of the study was approved by the Ethics Committee of the Santa Lucia Foundation.

To assess the efficacy of the proposed training program, participants were randomly divided into an experimental (EG) ($N = 6$) and a control group (CG) ($N = 6$). The homogeneity of the two groups in terms of anthropometrical characteristics (age, mass, and stature), functional classification score (*FCS*), and years of wheelchair basketball practice was assessed by means of an unpaired t-test, after having verified the normality distribution of the data. No significant difference was found ($p > 0.05$) between the two groups for each tested variable.

2.2. Study Design. Three experimental sessions were carried out, during which an instrumented 20-metre sprint test was performed. The first session (ES1) was performed in October, at the beginning of the training season, and was used to fulfil the first aim of the study (i.e., the biomechanical characterisation). The results obtained at this stage were used as a guidance to design a discipline- and population-specific training program (see Section 2.5), which was administered only to EG for three months, during the second half of the competitive season (from February to April). Within the same period, both CG and EG underwent the standard training usually proposed by the team coach. This training was administered twice a week, lasted about 90 minutes, and included exercises aimed at improving all aspects determining wheelchair basketball performances, that is, wheelchair propulsion, wheelchair and ball handling, wheelchair manoeuvrability, and shot and pass accuracy. To

Table 1: Participant characteristics.

Subject	Age [years]	Mass [kg]	Stature [m]	*FCS*	Years of practice	Gender	Group
S1	20	66	1.70	2.5	7	Male	CG
S2	18	48	1.68	3	6	Male	CG
S3	15	42	1.66	2	4	Female	CG
S4	17	61	1.75	0.5	3	Male	CG
S5	12	46	1.45	2	2	Male	CG
S6	17	96	1.78	4.5	3	Male	CG
S7	20	48	1.55	2	7	Male	EG
S8	19	49	1.69	3	7	Male	EG
S9	16	42	1.66	2.5	5	Female	EG
S10	18	74	1.75	0.5	4	Male	EG
S11	13	45	1.45	2	3	Male	EG
S12	20	105	1.90	4.5	3	Male	EG
Descriptive statistics	17.1 ± 2.7	60.2 ± 21.4	1.7 ± 0.1	2.25 ± 1	4.5 ± 1.8		

Characteristics of all participants: descriptive statistics are reported in terms of mean ± standard deviation (SD), except for *FCS* where median ± interquartile range (IQR) is indicated. *FCS*: functional classification score (ranging from 0.5 to 4.5 as proposed by the Italian Paralympic Committee); CG: control group; EG: experimental group.

assess the efficacy of the proposed discipline- and population-specific training program, CG and EG were tested before and after its administration. Specifically, a second experimental session (ES2) was performed in January and was considered as a baseline, whereas a third and final experimental session (ES3) was carried out in May, at the end of the three-month administration period.

The three experimental sessions, as well as the specific and the standard training program, were performed on the basketball court of the Santa Lucia Foundation and each athlete used the wheelchair commonly adopted during both training practice and competition. The same experimental protocol was followed for all the three sessions, as described in the following section.

2.3. Experimental Protocol and Instrumentation. During each experimental session, each athlete was equipped with three wearable IMUs (Opal, APDM Inc., Portland, Oregon, USA). These devices embed three-axial accelerometers (±6 g of full-range scale, 128 samples/s) providing the components of the vector sum of gravitational and inertial linear accelerations along the axes of a coordinate system fixed with the unit. Two IMUs were fixed on the right and left wrists using elastic bands, while the third unit was securely attached to the backrest of the wheelchair using double-sided tape (Figure 1).

Each athlete performed a 20-metre sprint test (20 mS). This test, focusing in particular on start-up and steady state velocities, was selected within those proposed for adult wheelchair basketball players [23] and then validated in terms of reliability for junior athletes [24]. It includes crucial factors in wheelchair basketball performance, such as starting and sprinting, which involve both strength capacity and coordination skills [23, 25]. Athletes started from a static position with the front wheels behind the start line and, after the start signal, pushed themselves for 20 metres as fast as possible. Time was manually recorded using a digital stopwatch ($t_{20\,\mathrm{mS}}$). The test was performed twice and the trial corresponding to the athletes' shortest time was further considered.

Figure 1: Wheelchair- and wrist-mounted IMUs with the relevant systems of reference.

In addition, before the beginning of the competitive season, each subject was medically examined and the peak power output (*PO*) of the upper arms was obtained using an arm crank ergometer (LODE, Groningen, Netherlands). The test protocol was defined according to previous performances obtained by each athlete on the same arm crank ergometer and following the indications of published studies on wheelchair athletes [26, 27]. The initial workload was set to 5 W for the athletes with a *FCS* equal or inferior to 2.5, whereas it was set to 10 W for those athletes with a score higher than 2.5. The workload was increased, every minute thereafter, by 5 or 10 W, according to the *FCS* of each athlete. Participants were instructed to maintain a cranking rate between 60 and 70 revolutions per minute. Testing

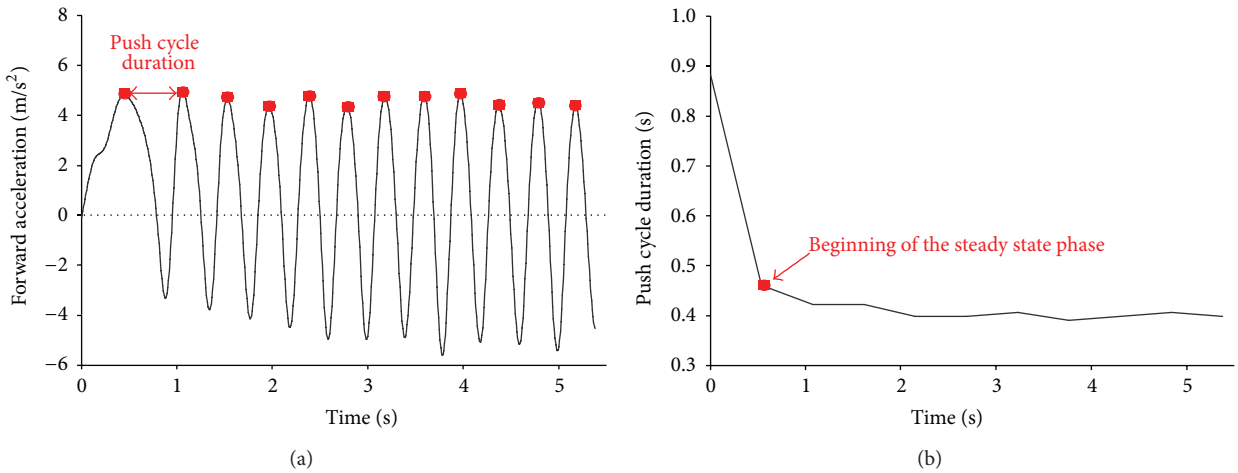

Figure 2: (a) Forward component of the acceleration measured by the wheelchair IMU. Dots indicate relative maxima. The duration of a cycle is also indicated with a double arrow. (b) Trend of the push cycle duration over time. The beginning of the steady state phase is indicated.

was terminated when the subject was unable to maintain the required cranking rate or upon the athlete's volitional fatigue (about 10–12 minutes after the beginning of the test). The normalised *PO* with respect to each athlete's mass was hereafter considered as an indicator of the subject's aerobic physical capacity.

2.4. IMU Data Processing. To remove random noise, the measured accelerations were low-pass-filtered with a cut-off frequency of 12 Hz using a 4th-order zero-lag Butterworth filter [28]. The calibration of the accelerometers was verified by performing *ad hoc* data collection and following the procedure described in [29]. The inclination with respect to gravity of the IMU located on the wheelchair backrest was computed during the stationary phase (which lasted about five seconds) at the beginning of each trial using IMU data and following the pitch-roll-yaw rotation sequence [29]. The IMU measurements were thus expressed, through a rigid transformation, in a local reference frame having one axis aligned vertically and one aligned with the progression direction. As the IMU was rigidly attached to the wheelchair and no wheelies occurred during the 20 mS, it can be assumed that the above-mentioned system of reference maintained a constant orientation throughout the 20 mS except for rotations occurring about the vertical axis.

The forward component of the acceleration measured by the wheelchair IMU was then used to identify the beginning of the steady state phase. To this aim, the signals were further low-pass-filtered with a cut-off frequency of 4 Hz and relative maxima were detected on the curve (Figure 2(a)). The time duration between two consecutive peaks was calculated and associated with the push cycle duration. The beginning of the steady state phase was set as the instant of time where a rapid change in the time duration slope occurred (Figure 2(b)).

Based on previous literature [9, 10, 30] and on the team coach's suggestions, for each session/subject/trial, the following parameters were estimated from the acceleration signals during the steady state phase.

Timing. The push cycle duration (Δt) was defined as the average of the period of time from two consecutive maxima identified on the forward component of the acceleration measured by the IMU located on the backrest of the wheelchair. The push cycle frequency (f) was defined as the number of push cycles per minute.

Progression Force. The progression force (F_p) was defined as the product of m times a_p, where m is the total mass obtained as the sum of the wheelchair, IMU, and athlete's masses and a_p is the peak value of the forward component of the acceleration measured by the IMU located on the wheelchair. This parameter, which was obtained for each push cycle, represents an estimation of the anteroposterior component of the force applied to the centre of mass of the whole system composed by wheelchair, athlete, and IMU. The average of the F_p parameter over the steady state phase of each trial was calculated and considered for the statistical analysis.

Bilateral Symmetry. A parameter associated with the symmetry (sym) between the dominant arm and nondominant arm in pushing the wheelchair was computed for each push cycle according to [30] and considering the peak acceleration measured by the IMUs located on the wrists

$$sym = \frac{a_{p\_\mathrm{dom}}}{\left(a_{p\_\mathrm{dom}} + a_{p\_\mathrm{nondom}}\right)} \cdot 100, \tag{1}$$

where a_{p_dom} and a_{p_nondom} correspond to the peaks of the acceleration magnitude as measured by the IMUs positioned on the wrist of the dominant and nondominant arms, respectively. This parameter was calculated to investigate whether the dominant and nondominant arm wrists presented similar peak accelerations. A value ranging between 45% and 55% indicates good symmetry, whereas a value lower than 45% or higher than 55% reflects greater accelerations for the nondominant or dominant hand rim, respectively [30]. The

average of the *sym* parameter over the steady state phase was calculated and hereafter considered.

Intercycle Variability. The intercycle variability was assessed by computing the coefficient of variation (CV) of Δt, F_p, and *sym* over all consecutive push cycles of the steady state phase. The mean and standard deviation (SD) of the above listed parameters were calculated and the relative coefficients of variation were obtained as follows:

$$CV_{\text{parameter}} = \left(\frac{\text{SD}_{\text{parameter}}}{\text{mean}_{\text{parameter}}} \right) \cdot 100. \quad (2)$$

This index is considered as an indicator of the push-to-push movement consistency and low values of the CV have been associated with the effective execution of repetitive movement patterns [10, 31], related to task specific coordinative skills [32].

2.5. Training Program Definition and Administration. Based on the results of the above-mentioned instrumented 20 mS and on literature evidence related to the demands of wheelchair basketball athletes [20, 23, 33], a training program specific for the population included in the study was developed. This program focused on both strength and coordination training.

Strength exercises involved the following muscle groups: biceps, triceps, middle trapezius, and shoulder abductors and adductors [4]. Due to the presence of athletes affected by spasticity, exercises included limited repetitions (three sets of 15–20 repetitions) [34] and were performed using elastic bands and following the guidelines proposed in the literature [35, 36]. Resistance was modulated by shortening or doubling the elastic bands. All strengthening exercises were performed within the subjects' tolerance and stopped if pain or discomfort was reported.

Coordination exercises focused on the sport specific components of coordination skills and, in particular, on spatial orientation, kinaesthetic differentiation, reaction, adaptation, combination, and rhythm [37]. The intensity of the exercises was increased over the administration time span by using balls of different masses and dimensions (tennis, medicine, soft, and basketball), by combining (limiting) visual or auditive stimuli to specific body movements stressing the cognitive component, and by changing the rhythm of the exercise [38].

Following the guidelines proposed by Faigenbaum et al. [35], the training program was administered to the six athletes of the EG twice a week for three months (12 weeks). Each session lasted 30–35 minutes and was divided into three parts: warm-up, conditioning phase, and cooldown (including stretching and breathing exercises). Within each month, the first and the third weeks were dedicated to coordination exercises whereas the second and the fourth weeks were focused on strength training.

2.6. Statistical Analysis. The statistical analysis was performed using the IBM SPSS Statistics software (IBM Corp., Armonk, NY, USA). The alpha level of significance was set to 0.05 for all statistical tests.

Table 2: Results of the biomechanical characterisation at ES1.

Parameter	Descriptive statistics	Correlation with $t_{20\,\text{mS}}$	
		ρ	p value
$t_{20\,\text{mS}}$ [s]	6.4 ± 1.1		
FCS	2.25 ± 1	−0.675	$p < 0.001$
PO [W/kg]	1.46 ± 0.69	−0.870	$p < 0.001$
Δt [s]	0.47 ± 0.06	0.757	$p < 0.001$
f [push/min]	113.93 ± 14.86	−0.631	$p < 0.001$
F_p [N]	342.10 ± 240.37	−0.242	$p < 0.05$
sym [%]	48.77 ± 3.27	−0.659	$p < 0.001$
$CV_{\Delta t}$ [%]	6.86 ± 6.67	0.548	$p < 0.001$
CV_{F_p} [%]	11.09 ± 6.84	0.448	$p < 0.001$
CV_{sym} [%]	4.58 ± 3.16	0.520	$p < 0.001$

Descriptive statistics and Spearman correlation coefficients of parameters obtained during ES1. Data are expressed as mean ± SD except for *FCS* reported as median ± IQR. $t_{20\,\text{mS}}$: time to complete the 20-metre sprint test; *FCS*: functional classification score; *PO*: upper arms peak power output; Δt: push cycle duration; f: push cycle frequency; F_p: peak progression force; *sym*: bilateral symmetry index; $CV_{\Delta t}$: coefficient of variation of the push cycle duration; CV_{F_p}: coefficient of variation of the peak progression force; CV_{sym}: coefficient of variation of the bilateral symmetry index.

2.6.1. Biomechanical Characterisation. For each subject and each trial, the normal distribution of the IMU-based estimated parameters was verified using the Shapiro-Wilk test of normality. As all parameters were not normally distributed, the Spearman (ρ) correlation coefficient was used to explore the relationship between the 20 mS performance and each estimated parameter. In addition, to investigate which parameter, among those estimated, could be used to predict the value of $t_{20\,\text{mS}}$, a linear stepwise multiple regression analysis was performed using $t_{20\,\text{mS}}$ as dependent variable and Δt, f, F_p, *sym*, $CV_{\Delta t}$, CV_{F_p}, CV_{sym}, and *FCS* and *PO* as independent variables.

2.6.2. Training Program Assessment. For both ES2 and ES3, each subject, and each trial, the normal distribution of the IMU-based estimated parameters was verified using the Shapiro-Wilk test of normality. As all parameters were not normally distributed, a Wilcoxon signed-rank test was performed to investigate whether significant differences existed between ES2 and ES3 sessions, for both EG and CG. In addition, Mann-Whitney U test was carried out to assess whether significant differences existed between EG and CG, for both ES2 and ES3.

3. Results

3.1. Biomechanical Characterisation. Descriptive statistics of $t_{20\,\text{mS}}$, *FCS*, and *PO* and of all the parameters estimated from the IMU data during ES1 are presented in Table 2. Spearman correlation coefficients between $t_{20\,\text{mS}}$ and each above-mentioned parameter are also reported, together with

Table 3: Results of the multiple regression model.

Parameter	Unstandardized coefficients		t	p value	β confidence interval (95%)	
	β	Std. error			Lower bound	Upper bound
Constant	7.344	0.225	32.593	$p < 0.001$	6.896	7.792
PO	−1.021	0.104	−9.799	$p < 0.001$	−1.229	−0.814
CV_{sym}	0.100	0.020	4.998	$p < 0.001$	0.060	0.140
$CV_{\Delta t}$	0.031	0.009	3.634	$p < 0.001$	0.014	0.048
FCS	−0.151	0.052	−2.906	$p < 0.005$	−0.254	−0.048

Unstandardized β coefficients of the regression equation and relevant statistics (standard error, t-test results in terms of t and p value, 95% confidence interval). Constant: constant parameter; *PO*: upper arms peak power output; CV_{sym}: coefficient of variation of the bilateral symmetry index; $CV_{\Delta t}$: coefficient of variation of the push cycle duration; *FCS*: functional classification score.

Table 4: Results of the training program assessment (ES2 and ES3 results).

Parameter	Group	ES2	ES3	Wilcoxon test	
		Mean ± SD	Mean ± SD	p value	Z
$t_{20\,\mathrm{mS}}$ [s]	CG	6.1 ± 1.2	5.9 ± 1.1	0.248	−1.156
	EG	6.1 ± 1.2	6.2 ± 1.1	0.080	−1.753
Δt [s]	CG	0.44 ± 0.05	0.45 ± 0.08	0.705	−0.378
	EG	0.47 ± 0.09	0.44 ± 0.08	0.027*	−2.207
f [push/min]	CG	117.32 ± 14.62	118.30 ± 21.02	0.600	−0.524
	EG	112.65 ± 18.71	119.97 ± 15.74	0.028*	−2.201
F_p [N]	CG	276.56 ± 148.16	289.57 ± 241.16	0.463	−0.734
	EG	330.71 ± 191.10	588.32 ± 312.29	0.028*	−2.201
sym [%]	CG	48.91 ± 3.82	47.86 ± 3.82	0.173	−1.363
	EG	47.77 ± 3.56	48.62 ± 3.96	0.046*	−1.992
$CV_{\Delta t}$ [%]	CG	4.36 ± 3.64	6.65 ± 4.81	0.173	−1.363
	EG	5.00 ± 3.38	4.12 ± 4.59	0.345	−0.943
CV_{Fp} [%]	CG	12.00 ± 6.93	15.24 ± 10.47	0.249	−1.153
	EG	12.16 ± 9.92	7.41 ± 3.12	0.249	−1.153
CV_{sym} [%]	CG	5.57 ± 4.38	4.56 ± 3.41	0.249	−1.153
	EG	4.35 ± 2.74	4.17 ± 2.83	0.917	−0.105

Descriptive statistics and Wilcoxon signed-rank test results (in terms of p value and Z value) for both the CG and the EG, before (ES2) and after (ES3) the training program administration. Significant differences are indicated with an asterisk. Data are expressed as mean ± SD. $t_{20\,\mathrm{mS}}$: time to complete the 20-metre sprint test; Δt: push cycle duration; f: push cycle frequency; F_p: peak progression force; *sym*: bilateral symmetry index; $CV_{\Delta t}$: coefficient of variation of the push cycle duration; CV_{Fp}: coefficient of variation of the peak progression force; CV_{sym}: coefficient of variation of the bilateral symmetry index.

the relevant p value. Strong correlations were obtained for the *FCS*, *PO*, Δt, f, and *sym*. Moderate correlations were found for all the coefficients of variation ($CV_{\Delta t}$, CV_{sym}, and CV_{F_p}), and a weak negative correlation was obtained for F_p. All the reported correlations were statistically significant.

The results of the multiple regression analysis are reported in Table 3. Four predictors were included in the model: *PO*, CV_{sym}, $CV_{\Delta t}$, and *FCS* ($R^2 = 0.826$, $R^2_{\text{Adjusted}} = 0.818$, standard error = 0.484, F(4,7) = 102.205, $p < 0.001$). All of them significantly contributed to predicting values of $t_{20\,\mathrm{mS}}$ and accounted for more than 80% of $t_{20\,\mathrm{mS}}$ variance. Both *PO* and *FCS* had significant negative regression scores, while CV_{sym} and $CV_{\Delta t}$ had significant positive scores.

3.2. Training Program Assessment. The mean and SD of $t_{20\,\mathrm{mS}}$ and of all the IMU-based estimated parameters, together with the results of the comparison between ES2 and ES3 for both the CG and the EG, are reported in Table 4. No significant difference between ES2 and ES3 was found for the CG. Conversely, for the EG, significant differences were obtained for Δt, f, F_p, and *sym* when comparing those parameters before and after the administration of the training program. No significant difference was found between the two groups, for both ES2 and ES3.

4. Discussion

This study presents a method for the quantitative characterisation of a wheelchair basketball field test (20-metre sprint test) based on the use of inertial measurement units (IMUs). A list of biomechanical parameters was defined as performance indicators, which allowed for the development of a sports- and population-specific training program. The method was applied to a team of junior wheelchair basketball athletes and was used to assess the efficacy of the developed training program which was administered for three months.

4.1. Biomechanical Characterisation. The proposed instrumented 20-metre sprint test (20 mS) allowed extracting a list of biomechanical indices associated with the performance of the test ($t_{20\,\mathrm{mS}}$). Differently from common performance scores based on the overall outcome (product) of field tests, the estimated biomechanical indices allowed obtaining additional information related to the way this outcome was obtained (process). In particular, from IMU data, parameters were estimated related to the propulsion timing, the progression force that accelerates the wheelchair, the symmetry between the right and left arms in pushing the wheelchair, and the intercycle variability of these parameters.

The results about $t_{20\,\mathrm{mS}}$ are in general agreement with the existing literature, whereas the observed discrepancies are mainly attributed to the different participants' age [24, 39] or to the slightly different testing protocol, which, in other studies, involves the use of the ball during wheelchair propulsion [25]. Similarly, the results about the timing parameters (Δt and f) confirm previous findings [8, 40]. Further comparisons with other studies can be hardly performed, as the testing protocols rarely include velocity above 2 m/s [10, 11].

In terms of aerobic power, the present results are in agreement with previous findings about elite wheelchair athletes [27, 41]. According to the existing literature [27], a strong correlation exists between the maximal oxygen consumption $VO_{2\,\max}$ and *PO*. Thus, an estimation of the $VO_{2\,\max}$ can be obtained from the *PO* results, so as to have an overall picture of the athletes' aerobic physical fitness. For the participants of the present study, mean values of predicted $VO_{2\,\max}$ were about 1.7 l/min for female and 2.1 l/min for male athletes. These values are comparable with those reported in [41], where elite wheelchair basketball athletes were considered, indicating that the tested sample had a good level of aerobic physical fitness.

The results of the correlation analysis provide interesting insights into the importance of the biomechanical characterisation of the wheelchair propulsion during the 20 mS. All parameters display a statistically significant relationship with the performance index, $t_{20\,\mathrm{mS}}$ (correlation coefficients ranging from 0.24 to 0.87), indicating that the estimated indicators not only provide information about the athletes' wheelchair propulsion technique, but also are descriptive of the overall test performance.

In particular, the peak power output (*PO*), the *FCS*, both timing parameters (Δt and f), and the bilateral symmetry index (*sym*) strongly correlate with $t_{20\,\mathrm{mS}}$. These results, which confirm the findings of previous studies [8, 10], reveal that better performances are related to the ability of the athletes to generate high power with the upper arms as well as to increase the push frequency (thus decreasing the push cycle duration) and the push symmetry between the right and left arms during propulsion. In particular, as far as the push symmetry is concerned, four main aspects emerge from the existing literature: (i) symmetrical and synchronous pushing modes are associated with greater wheelchair velocity and push power [8]; (ii) the presence of upper-extremity asymmetry when pushing the wheelchair may contribute to the development of injury [9]; (iii) a close relationship exists between upper arms coordination and both technical efficiency and injury prevention [33]; (iv) symmetry indices are often used to describe upper arm coordination in different sport activities [42]. The results of the present work, therefore, confirm the importance of the push symmetry as a valuable performance- and injury-related indicator.

Moderate positive correlations were obtained for all the computed coefficients of variation (*CV*), showing that the lower the intercycle variability, the lower the $t_{20\,\mathrm{mS}}$. This is in agreement with the idea that a high level of expertise results in the capacity to reproduce a movement like an automatism [43] and that low values of *CV* are associated with the effective execution of repetitive movement patterns [10, 31]. This does not imply that expert athletes are able to reproduce identical movement patterns but rather that they are capable of picking up several sources of information (visual, haptic, and acoustic) to perform different movements and to use the so-called coexisting modes of coordination [32] to achieve consistent and effective functional performance outcomes [43].

A significant negative relationship between $t_{20\,\mathrm{mS}}$ and the peak progression force (F_p) was found, although less evident than that of other parameters. In particular, this correlation suggests that, as expected, the higher the force that accelerates the wheelchair, the better the test outcome. The relationship between the wheelchair propulsion efficiency and the force applied by each arm on the hand rim, as well as the symmetrical/asymmetrical distribution of this force, is well established in the literature. In particular, the greater the force is (or the power output), the more mechanically efficient the wheelchair propulsion is [10, 20]. The weak correlation between F_p and $t_{20\,\mathrm{mS}}$ found in the present study can be related to the fact that F_p does not coincide with the actual force applied by the athlete on the hand rims, but it is rather an estimation of the anteroposterior component of the force applied to the centre of mass of the whole system composed by wheelchair, athlete, and IMU. Still, this parameter proved to be significantly related to the athletes' performance and can be estimated directly in the field and in real training conditions.

According to the results of the correlation analysis, the regression model identified four significant parameters able to predict $t_{20\,\mathrm{mS}}$, that is, *PO*, CV_{sym}, $CV_{\Delta t}$, and *FCS*. Based on these results, two aspects were selected as targets of the designed discipline- and population-specific training program: (i) strength, here represented by *PO* [10, 20], and (ii) coordination, here represented by CV_{sym} and $CV_{\Delta t}$ [32]. This is in agreement with the existing literature which identifies wheelchair propulsion dynamics and eye-hand coordination as the two most important constructs for successful wheelchair basketball performance [23, 25].

4.2. Training Program Assessment. In the present study, an *ad hoc* training program aimed at improving both strength capacity and coordination skills was developed. On the one hand, no significant difference was found for $t_{20\,\mathrm{mS}}$ after three months of administration, neither for the CG nor for the EG, indicating that the athletes' performance during the 20 mS did not change from ES2 to ES3. This result is in agreement

with a previous study [22] which assessed the efficacy of resistance training on six paediatric wheelchair users. In this study, no significant difference was found in the performance of a 50-metre dash test after 8 weeks of training. It should be noticed that, in the present study, the scores of the athletes were similar to those displayed by elite athletes of the same discipline [39] in all three experimental sessions. This aspect, together with the good aerobic physical fitness of the athletes, suggests that there may be limited room for improvement and that three months of program administration could be too short to produce significant changes in the 20 mS final outcome. In addition, the time taken by each subject to complete the test was recorded manually with a digital stopwatch as provided by the established protocol [10, 23–25]. Therefore, the accuracy in measuring the time to complete the test is of the same order as the reaction time typical of thumb movements, that is, about 0.3 s [44], and the detected changes of $t_{20\,\mathrm{mS}}$ are, indeed, within this margin. If actual variations in $t_{20\,\mathrm{mS}}$ take place, they are below the accuracy threshold and, thus, hardly detectable with this instrumentation.

On the other hand, when considering the IMU-based biomechanical parameters related either to strength or coordination, differences between ES2 and ES3 were obtained for the EG only. In particular, significant improvements were displayed for Δt, f, F_p, and sym, showing that EG athletes modified their propulsion technique increasing the push cycle frequency, the force expressed to accelerate their wheelchair, and adopting a more symmetrical pushing mode. In addition, all the CV parameters slightly decrease from ES2 to ES3 ($p > 0.05$), suggesting a more effective execution of the repetitive pushing movements, typical of expert athletes [31].

These results indicate that although the administered training program does not influence the final outcome of the 20 mS, it affects the way this outcome is obtained by the athletes. This supports the hypothesis that biomechanical analysis can effectively provide additional performance indicators to the coaches relative to the way specific movements are performed, not limiting the analysis on the final product of the selected motor task.

4.3. Limitations of the Study. The investigation included a relatively small sample size, particularly when considering the control and experimental groups separately (i.e., during ES2 and ES3). Also, the study involved participants with different pathologies and variable level of spinal cord lesions. It is plausible that these two aspects play an important role in determining the lack of significant differences between CG and EG after the training program administration. Still, significant differences for a list of biomechanical parameters were identified within each group when comparing the results obtained before and after the administration of the program. It is possible that between-group differences might have been pointed out if a larger cohort of athletes would have been admitted to the program.

Conversely, no statistical difference was detected when comparing the times to complete the 20 mS. This is probably related to the inadequate level of accuracy of the manual digital stopwatch, which could be improved by using photocells or a laser gun. Moreover, it is not excluded that differences in the test score could emerge from a program administration period greater than three months.

In the present study, the progression force was estimated by analysing the acceleration signals measured by the IMU located on the wheelchair. F_p represents, therefore, an estimation of the anteroposterior component of the force applied to the centre of mass of the whole system composed of wheelchair, athlete, and IMU. Although this parameter does not represent the actual force applied by each arm on the wheel hand rim, it proved to be effective in revealing the changes occurring after the training program administration. Furthermore, contrary to traditional experimental protocols which need either to modify the wheelchair with instrumented wheels (increasing the wheelchair mass) or to use a static ergometer, F_p can be estimated directly in the field and in real training conditions by using only one IMU located on the back of the wheelchair.

5. Conclusions

This paper fills an existing gap in the field of junior wheelchair basketball. A methodology for the biomechanical assessment of the wheelchair propulsion was developed and a list of biomechanical indices associated with the performance of a 20-metre sprint test was obtained by means of wheelchair- and wrists-mounted inertial measurement units. These indices proved to correlate with the test performance and provide quantitative information about the way athletes obtained such performance. Therefore, they were used to define a sports- and population-specific training program focused on strength and coordination training. The proposed biomechanical methodology was then used to assess the efficacy of the defined training program after three months of administration. The estimated indices were effective in identifying both strength and coordination improvements following the training administration.

Both the biomechanical assessment method and the training program proved to be well perceived by the athletes and to be applicable in training conditions. Special attention, in fact, was paid to the organisational and practical aspects of the experimental protocol and of the program administration. It is worth underlining that the results of the present study were achieved thanks to the effective interaction within the multidisciplinary research group, which allowed addressing and answering the needs of both coaches and physiotherapists, through the complementary expertise of biomechanists, in terms that were valuable to the former professionals.

Symbols

IMU:	Inertial measurement unit
CG:	Control group
EG:	Experimental group
ES1, ES2, and ES3:	First, second, and third experimental sessions, respectively
20 mS:	20-metre sprint test

FCS: Functional classification score
PO: Upper arms peak power output
$t_{20\,\mathrm{mS}}$: Time to complete the 20-metre sprint test
Δt: Push cycle duration
f: Push cycle frequency
F_p: Peak progression force
a_p: Peak acceleration
sym: Bilateral symmetry index
CV: Coefficient of variation
SD: Standard deviation
IQR: Interquartile range.

Conflict of Interests

The authors declare no conflict of interests.

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Research Article

Attenuation of Upper Body Accelerations during Gait: Piloting an Innovative Assessment Tool for Parkinson's Disease

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The objective of the current investigation was to explore whether upper body accelerations obtained during gait provide sensitive measures of postural control in people with Parkinson's disease (PD). Thirteen people with PD (70 ± 11 years) and nineteen age-matched controls (70 ± 7 years) walked continuously for two minutes while wearing three inertial sensors located on their lower back (L5), shoulder level (C7), and head. Magnitude (root mean square (RMS)), attenuation (attenuation coefficient), and smoothness (Harmonic ratios, HR) of the accelerations were calculated. People with PD demonstrated greater RMS, particularly in the mediolateral direction, but similar harmonic ratio of head accelerations compared to controls. In addition, they did not attenuate accelerations through the trunk and neck as well as control participants. Our findings indicate that measuring upper body movement provides unique information regarding postural control in PD and that poor attenuation of acceleration from the pelvis to the head contributes to impaired head control. This information is simple to measure and appears to be sensitive to PD and, consequently, is proposed to benefit researchers and clinicians.

1. Introduction

People with Parkinson's disease (PD) walk with a gait pattern that is characterised by slowness (bradykinesia), muscle rigidity, and postural instability [1–3]. As the disease progresses, postural control deteriorates and predisposes people with PD to falls [4–6]. Current measures of postural control, based on the ability to maintain upright balance during quiet stance, poorly reflect real life situations when people with PD are at risk of falling. Consequently, researchers and clinicians are promoting the measurement of postural control during gait [7, 8].

The recent development of small and inexpensive wireless inertial sensors has helped facilitate routine measurement of postural control during gait in the clinic, laboratory, and the community. Emerging evidence suggests that measuring upper body acceleration during gait using inertial sensors can objectively quantify differences in gait patterns between those with and without PD [9, 10]. It has also been shown that upper body accelerations are sensitive to differences between PD fallers and nonfallers [11]. Specifically, these studies have revealed a deterioration of the smoothness of trunk accelerations in people with PD as measured by harmonic ratios, which was more pronounced in those with a history of falls.

Despite emerging evidence that maintaining head stability during gait is a key determinant of postural control [12–16], accelerations of the head have been neglected in these previous studies examining upper body acceleration in PD. One potential reason head stability is important is that the head contains the visual and vestibular systems, which are critical for navigation and preplanning of adaptive motor strategies [13]. Head stability may have added importance for people with PD because they rely heavily on vision to

maintain their postural control [5]. Recent evidence suggests that vision during gait is affected in PD [17] and that the smoothness of trunk accelerations is also altered [9, 10]. However, it has not yet been established whether PD affects the stability of the head during gait. A key mechanism in maintaining head stability is the attenuation of accelerations through the trunk. People with PD often develop axial rigidity, which may impair their ability to attenuate the accelerations that are applied to the lower limbs during gait from impacting on head stability. The measurement of attenuation of accelerations through the upper body has previously been investigated as a strong postural control indicator for children, adults, and elderly individuals [18–21] but has not yet been examined in people with PD.

The objective of the current investigation was to explore whether upper body accelerations obtained during gait provide sensitive proxy measures of postural control in people with Parkinson's disease (PD). More specifically, the aims of this study were to assess the magnitude, attenuation, and smoothness of upper body accelerations in people with PD compared to age-matched controls. We tested the hypotheses that: people with PD would demonstrate impaired smoothness and attenuation of accelerations. To address these aims, accelerations of the head, trunk, and pelvis were assessed during gait in a cohort of people with PD and an age-matched control group.

2. Materials and Methods

2.1. Participants. A subsection of community dwelling older adults and people with PD were tested as part of the ongoing ICICLE-PD (Incidence of Cognitive Impairment in Cohorts with Longitudinal Evaluation—Parkinson's Disease) Gait study [22, 23]. Participants were excluded if they had any neurological (other than PD), orthopaedic, or cardiothoracic conditions that may have markedly affected their walking or safety during the testing sessions. In addition, PD participants had to be diagnosed with idiopathic PD according to the UK Parkinson's Disease Brain Bank criteria and were excluded if they presented with significant memory impairment (Mini Mental State Exam (MMSE) ≤ 24 [24]), dementia with Lewy bodies, drug induced parkinsonism, "vascular" parkinsonism, progressive supranuclear palsy, multiple system atrophy, corticobasal degeneration, or poor command of English. This study was conducted according to the Declaration of Helsinki and had ethical approval from the Newcastle and North Tyneside research ethics committee. All participants signed an informed consent form.

2.2. Experimental Protocol. All participants walked for two minutes at their preferred pace around a 25 m circuit, fully described in [27]. Spatiotemporal gait variables (walking speed, step time, step length, and step width) were measured using a 7 m long Gaitrite pressure activated electronic walkway (Platinum model Gaitrite, software version 4.5, CIR systems, United States of America). Upper body accelerations were measured using three OPAL inertial sensors sampling at 128 Hz (APDM Inc, Portland, OR, USA) located at 5th lumbar vertebra to represent the pelvis level (P), the 7th cervical vertebra to represent the shoulder level (S) and upon the back of the head (H). The Gaitrite and the OPAL system were synchronised and the data was collected using the same A/D converter.

2.3. Data Analysis. To ensure only steady-state, straight-line walking was analysed, only the portion of the acceleration data recorded while participants who were in contact with the Gaitrite walkway was used. As detailed in Mazzà et al. [20], prior to collecting the gait data, a calibration trial was captured using a sensor placed on the floor to create a global reference frame for the laboratory. Thereafter, the local reference frame of each sensor was reoriented for each time sample to the newly established global reference frame [19, 28]. Following, the acceleration data was further segmented based upon the foot contact and foot off values obtained from the Gaitrite walkway. Then, the mean value of the signal was removed and a low-pass fourth order Butterworth filter with a cut-off frequency of 10 Hz was applied [19]. Data for each stride was normalised to 100 data points using linear interpolation. All signals were processed using MATLAB (version 8.1.0).

2.4. Magnitude of Acceleration. The magnitude of accelerations was calculated using the root mean square (RMS) of the accelerations, measured by each sensor for each stride in the Anteroposterior (AP), Mediolateral (ML), and Vertical (V) directions.

2.5. Attenuation of Acceleration. The ability to attenuate accelerations through the upper body was quantified using the attenuation coefficient. The attenuation coefficient describes the ability to reduce accelerations from inferior to the superior anatomical locations and was calculated using the RMS values for each stride.

The attenuation coefficients were computed using the RMS values of the head ($\mathrm{RMS_H}$), shoulder ($\mathrm{RMS_S}$), and pelvis ($\mathrm{RMS_P}$) as follows [18–20]:

$$C_{\mathrm{PH}} = \left(1 - \frac{\mathrm{RMS_H}}{\mathrm{RMS_P}}\right) \times 100,$$
$$C_{\mathrm{PS}} = \left(1 - \frac{\mathrm{RMS_S}}{\mathrm{RMS_P}}\right) \times 100, \quad (1)$$
$$C_{\mathrm{SH}} = \left(1 - \frac{\mathrm{RMS_H}}{\mathrm{RMS_S}}\right) \times 100$$

with C_{PH} representing the attenuation from the pelvis to the head, C_{PS} representing the attenuation from the pelvis to the shoulder, and C_{SH} representing the attenuation from the shoulder to the head. Each equation provides a percentage representing the amount of acceleration that is attenuated from the inferior sensors to the superiorly located sensor. A positive coefficient indicates reduced acceleration at the superiorly located sensor relative to the inferiorly located sensor. A negative coefficient value indicates a greater acceleration at the superiorly located sensor.

Table 1: The mean (±SD) participant characteristics and spatial-temporal gait variables for the PD and control group.

	PD ($n = 13$)	Control ($n = 19$)	P (t-test)
Age (years)	69.7 ± 11.1	70.2 ± 6.7	0.90
Height (m)	1.70 ± 0.10	1.72 ± 0.10	0.99
Mass (Kg)	77.9 ± 13.3	83.2 ± 14.2	0.30
BMI	26.1 ± 3.3	28.0 ± 4.5	0.20
MDS UPDRS III	35.6 ± 12.6	NA	NA
Hoehn and Yahr stage	HY II: 11; HY III: 2	NA	NA
Gait speed (m/s^2)	1.22 ± 0.22	1.32 ± 0.15	0.14
Step time (s)	0.54 ± 0.21	0.54 ± 0.44	0.97
Step length (cm)	0.66 ± 0.12	0.71 ± 0.07	0.15
Step width (cm)	0.09 ± 0.03	0.09 ± 0.02	0.46

*Significant difference at $P < 0.05$.
BMI: body mass index.
MDS UPDRS III: Movement Disorders Society Revised Unified Parkinson's Disease Rating Scale–Movement Subsection [25].
HY: Hoehn and Yahr stage [26].

2.6. Smoothness of Accelerations. We quantified the smoothness of upper body accelerations using the harmonic ratio (HR). The HR accurately describes the step-to-step symmetry within a stride but for upper body gait analysis is also commonly referred to as a measure smoothness [29]. The HR was calculated via discrete Fourier transform for each of the acceleration components measured at the H, S, and P levels in the AP, ML, and V directions [30]. The fundamental frequency was set equal to the stride frequency.

For the AP and V Components, the HR Was Defined as:

$$\mathrm{HR} = \frac{\Sigma \text{ Amplitudes of even harmonics}}{\Sigma \text{ Amplitudes of odd harmonics}} \quad (2)$$

For the ML Component, the HR Was Defined as:

$$\mathrm{HR} = \frac{\Sigma \text{ Amplitudes of odd harmonics}}{\Sigma \text{ Amplitudes of even harmonics}} \quad (3)$$

Higher values of HR are associated with a higher similarity between the pattern of the upper body movements occurring during the right and left steps and are therefore favourable [9, 31]. Following calculation, the HR's were normalised to each participant's gait speed [9, 14].

2.7. Statistical Analysis. A series of two-tailed paired t-tests were used to test the difference between groups for the magnitude, attenuation, and smoothness of accelerations. The level of significance was set at $P = 0.05$. Given the exploratory nature of this study, the P value was not adjusted for multiple comparisons.

3. Results

The characteristics of the participants are reported in Table 1. All the participants with PD were tested within 18–54 months post diagnosis. No significant differences were found between the two groups in terms of anthropometric characteristics or spatiotemporal gait values.

3.1. Magnitude of Acceleration. Significantly higher ML head accelerations were observed in people with PD compared to controls (1.08 ± 0.29 m/s^2 versus 0.86 ± 0.21 m/s^2, $P = 0.024$) but not at the pelvis or the shoulder level. There were no other significant between-group differences although AP and V head accelerations tended to be greater in the PD group (Table 2).

3.2. Attenuation of Acceleration. People with PD did not attenuate AP or ML accelerations as well as controls (Figure 1). For C_{PH}, a significant difference existed between PD and the control participants in the ML direction (0.12 ± 34.7% versus 33.8 ± 21.3%, $P = 0.003$). For C_{PS}, a significant difference existed between PD and controls in the AP (16.0 ± 15.6% versus 33.1 ± 12.4%, $P = 0.002$), as well as the ML direction (5.5 ± 24.5% versus 27.7 ± 18.6%, $P = 0.009$). For C_{SH}, a significant difference existed between the PD and the control group in the ML direction (−3.6 ± 15.5% versus 9.4 ± 15.3%, $P = 0.031$).

3.3. Harmonic Ratio. The HRs normalised to gait speed showed no significant differences between the PD and control participants (Table 3).

4. Discussion

Our current investigation provides evidence that upper body accelerations obtained during gait provide sensitive measures of postural control in people with Parkinson's disease (PD). As hypothesised, the results of this study showed that people with PD walked with altered upper body accelerations compared to age-matched controls. In particular, people with PD walked with greater magnitude of ML head accelerations and demonstrated impaired attenuation of accelerations from the pelvis and neck to the head. In contrast to our hypothesis, smoothness of upper body accelerations as measured by the HR was not significantly affected in this sample of PD.

To our knowledge, this is the first study to show impaired head stability in people with PD using inertial sensors. A greater magnitude of ML head acceleration was found for

Table 2: The mean (±SD) root mean square (RMS) for the PD and the control participants calculated at the head (H), shoulder (S), and the pelvis (P) levels.

Sensor location	Component	PD	Control	P (t-test)
H	AP	1.02 ± 0.24	0.92 ± 0.20	0.22
	ML	1.08 ± 0.29	0.86 ± 0.21	0.02*
	V	2.15 ± 0.74	2.41 ± 0.47	0.26
S	AP	1.03 ± 0.18	0.96 ± 0.16	0.31
	ML	1.05 ± 0.24	0.96 ± 0.17	0.25
	V	2.04 ± 0.64	2.28 ± 0.46	0.24
P	AP	1.28 ± 0.38	1.47 ± 0.33	0.14
	ML	1.17 ± 0.36	1.41 ± 0.42	0.11
	V	2.16 ± 0.70	2.35 ± 0.47	0.37

*Significant difference at $P < 0.05$.
H: head; S: shoulder level; P: pelvis.
AP: anterior/posterior; ML: medial/lateral; V: vertical.

Table 3: The mean (±SD) Harmonic ratios normalised to gait speed for the PD and the control participants calculated at the head (H), shoulder (S), and the pelvis (P) levels.

Sensor location	Component	PD	Control	P (t-test)
H	AP	0.71 ± 0.36	0.53 ± 0.23	0.11
	ML	1.22 ± 0.56	1.02 ± 0.38	0.27
	V	2.03 ± 0.57	2.18 ± 0.60	0.50
S	AP	0.70 ± 0.23	0.66 ± 0.22	0.64
	ML	0.80 ± 0.50	0.80 ± 0.22	0.95
	V	2.34 ± 0.77	2.51 ± 0.72	0.55
P	AP	1.22 ± 0.38	1.13 ± 0.48	0.61
	ML	1.05 ± 0.69	0.80 ± 0.38	0.22
	V	2.02 ± 0.60	2.17 ± 2.17	0.58

H: head; S: shoulder level; P: pelvis.
AP: anterior/posterior; ML: medial/lateral; V: vertical.

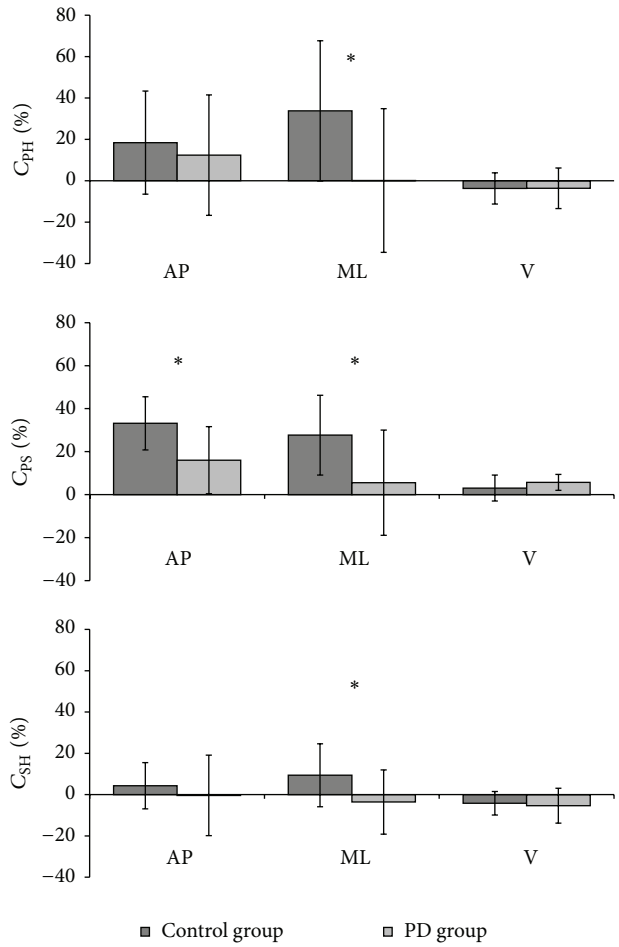

Figure 1: Mean (±SD) values of the attenuation coefficients (C_{PH}, C_{PS}, and C_{SH}) of the three acceleration components (AP = anterior/posterior, ML = medial/lateral; V = vertical), computed for the control and group with PD. *$P < 0.05$.

the PD group. This was interpreted as a result of poor postural control for the PD participants and a failure to stabilise their head in space [18, 21]. High values for the head accelerations have been previously described as a reduced ability to stabilise the head in space. This is particularly crucial for people with PD because of their aforementioned increased dependence upon visual input for correcting postural control [5]: higher accelerations are likely disturbing their visual system, leading to an impaired ability to preplan effective motor strategies [13], causing an increased likelihood to fall. Although they might be a useful measure of postural control, RMS values of head accelerations are known to be dependent upon step length and gait speed [31]. Despite no significant differences being observed for these parameters between the PD and the control group in this sample, it is common that PD affects both gait speed and step length [1–3]. As a result, the magnitude of accelerations may lack sensitivity when used for discriminating PD patients, which in other studies have been shown to possess a decreased step length and gait speed, when compared to age-matched controls.

Alternatively, being computed as a ratio between accelerations measured during the same trial [18], the coefficients of attenuation do not suffer from being speed dependent. In the current investigation; the coefficients of attenuation provided insight into why the PD participants demonstrated greater accelerations at the head. Participants with PD were less able to attenuate accelerations through the trunk, as shown by impaired pelvis-shoulder attenuation coefficients, which were reduced on average by at least a half in the PD cohort, both in the AP and in the ML direction. It is not possible to fully explain why the people with PD did not attenuate accelerations well through the upper body; however, it may be associated with *en bloc* movement and axial rigidity. It has previously been stated that increased rigidity may cause underlying changes in the physiological and mechanical functioning of the axial muscles which results in *en bloc* movement, where the head, trunk and pelvis move together

as one rigid unit [12, 32]. It might be assumed that the same mechanisms could be responsible for poor attenuation of accelerations through the spine in PD. However, more research is certainly needed to test this hypothesis and explain the mechanisms ruling altered head accelerations and poor attenuation in PD, as well as the implications of poor head stability on vision and postural control.

Interestingly, the findings regarding attenuation coefficients were strongest in the ML direction. Similar results were found even when analysing healthy elderly subjects [21]. The fact that instability was predominantly found in the ML direction, suggests that when utilising a coefficient of attenuation, the ML direction is potentially most informative of an impaired walking stability. Consequently, assessments in the ML direction may be best for proxy measures of postural control in PD.

In contrast with our hypothesis and previous studies [9], the smoothness of upper body accelerations (harmonic ratios) was not significantly affected in our sample of PD participants. The discrepancy with previous studies is unlikely due to methodological differences, as the studies were similar in design. It is possible, however, that we were statistically underpowered to detect group difference, as suggested by a 25% reduction of AP HRs and 16% reduction of ML HRs at the head in the PD group that did not reach statistical significance ($P = 0.106$). Further research is required to determine the effectiveness of harmonic ratios as a sensitive measure to PD at different stages of their disease progression, as well as its ability to predict future falls.

Clinicians require objective measures to assess postural control during locomotion in people with PD to supplement standard clinical assessments and conventional rating scales which are not sensitive to subtle postural control disturbance [31, 33, 34]. Our findings indicate that it is feasible to measure the magnitude, attenuation, and smoothness of upper body accelerations in people with PD using body worn sensors. The rapid technological development of inertial sensors may afford a quick, clinically appropriate, and cost effective method to measure postural control in the clinic and community settings [5]. Specifically, the attenuation coefficient is a promising measure that is sensitive to PD; however, larger longitudinal studies are needed to assess its ability to monitor disease progression, determine intervention efficacy, and inform clinical management [5, 34, 35].

5. Conclusion

The current investigation suggests that assessing upper body acceleration offers additional and unique information about postural control during gait in people with PD. In particular, the magnitude of ML head accelerations and attenuation of upper body acceleration appear sensitive to PD and consequently hold promise as useful proxy measures that can be utilised in clinical and community settings.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Acute Effect on Arterial Stiffness after Performing Resistance Exercise by Using the Valsalva Manoeuvre during Exertion

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Background. Performing resistance exercise could lead to an increase in arterial stiffness. *Objective*. We investigate the acute effect on arterial stiffness by performing Valsalva manoeuvre during resistance exercise. *Materials and Methods*. Eighteen healthy young men were assigned to perform bicep curls by using two breathing techniques (exhalation and Valsalva manoeuvre during muscle contraction) on two separate study days. Carotid pulsed wave velocity (cPWV) was measured as an indicator to reflect the body central arterial stiffness using a high-resolution ultrasound system, and its value was monitored repeatedly at three predefined time intervals: before resistance exercise, immediately after exercise, and 15 minutes after exercise. *Results*. At the 0th minute after resistance exercise was performed using the Valsalva manoeuvre during exertion, a significant increase in cPWV (4.91 m/s ± 0.52) compared with the baseline value (4.67 m/s ± 0.32, $P = 0.008$) was observed, and then it nearly returned to its baseline value at the 15th minute after exercise (4.66 m/s ± 0.44, $P = 0.010$). These findings persisted after adjusting for age, body mass index, and systolic blood pressure. *Conclusion*. Our result suggests short duration of resistance exercise may provoke a transient increase in central arterial stiffness in healthy young men.

1. Background

Resistance exercise can be performed using various breathing-control techniques during exertion, and the most commonly adopted breathing-control techniques involve performing exhalation or the Valsalva manoeuvre during exertion. To perform Valsalva manoeuvre, a person must inhale fully and exhale against a closed airway [1]. This process involves the contraction of the diaphragm, abdominal muscles, and other expiratory muscles [2]. As these muscle groups contract, the intra-abdominal pressure increases. Although such an increase in the intra-abdominal pressure is an extravascular event, the pressure is transferred and exerted onto the arterial wall of the abdominal aorta, leading to saturation in the elastin fibre of the abdominal aorta wall and a subsequent increase in the central arterial stiffness [3–5].

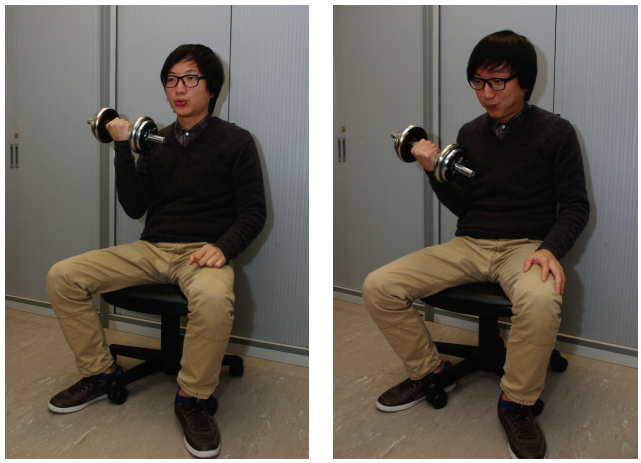
Previous studies have demonstrated that performing Valsalva manoeuvre during isometric exercise or heavy weight lifting can alter the hemodynamic function and cause a substantial increase in systolic blood pressure [6, 7]. However, when Valsalva manoeuvre was performed during resistance exercise, systolic blood pressure appeared to be unaffected [8]. Therefore, whether performing resistance exercise could affect central arterial stiffness is still controversial.

2. Objectives

The primary objective of this study was to identify the effect of performing Valsalva manoeuvre during resistance exercise on central arterial stiffness. We hypothesized that one bout of resistance exercise in the form of bicep curls combined with Valsalva manoeuvre during exertion can lead to an increase in central arterial stiffness in young men.

3. Materials and Methods

3.1. Subject. Eighteen healthy young men aged between 20 and 24 years were successfully recruited with informed consent. Subjects who had no history of diabetes, hypertension,



(a) (b)

Figure 1: The setup of bicep curls. A volunteer performing bicep curls on Day 1 (a) exhales during exertion and inhales when extending the arm and releasing the dumbbell and on Day 2 (b) performs Valsalva manoeuvre during exertion and exhales when extending the arm and releasing the dumbbell. Strict instruction was provided to every volunteer about the breathing technique during exercise.

cardiovascular disease, chronic kidney disease, or respiratory disease were included in the study. Subjects who took any type of regular medication and smokers were excluded from this study.

3.2. Ultrasound Measurement. The local institutional review board approved the experimental protocol. The study included two main experimental sessions. All of the subjects were asked not to consume any food or drink containing caffeine 1 day prior to the study. The two experimental sessions were conducted 1 week apart from each other to eliminate any accumulated effect of postexercise that might affect the outcome of the subsequent experimental session. In each session, all of the subjects were allowed to rest on a bed for 10 minutes before any measurements were taken. The one-repetition maximum (1-RM) test was performed on Day 1 to determine the muscle strength of the subjects based on the maximal weight that the subject could lift. The corresponding weight of the dumbbell (75% of 1-RM) was used in the two experimental sessions [9].

In each experimental session, the subjects were instructed to perform ten sets of repeated bicep curls. Each set of repeated bicep curls consisted of completing ten bicep curls within 20 seconds, and a resting period of 90 seconds was given between each set of repeated bicep curls. On Day 1, the subjects were asked to exhale during exertion and inhale when extending the arm and releasing the dumbbell. On Day 2, they were asked to perform Valsalva manoeuvre during exertion and exhale when extending the arm and releasing the dumbbell (Figure 1).

In both sessions, the brachial systolic and diastolic blood pressures were measured using the arm that did not perform the bicep curl. Brachial blood pressure and cPWV were measured at three predefined time intervals: before the exercise (baseline), immediately after the final set of bicep curls was completed (0th min after exercise), and 15 minutes after the final set of bicep curls was completed (15th min after exercise). For obtaining reproducible results, cPWV was always measured at a site 1 cm proximal to the bifurcation of the left common carotid arteries by using the RF-based Quality Arterial Stiffness (RFQAS) software on an Esaote MyLab Sat ultrasound system with a high-frequency linear transducer (Esaote SL3323, 13-6 MHz) (Figure 2). All of the ultrasound examinations were conducted by a single well-trained researcher (VM) to minimize interrater variability.

3.3. Data Analysis. All of the collected data were analysed using IBM SPSS (version 21.0.0) software. The average of three cPWV measurements during resistance exercise under different breathing techniques was calculated at three predefined time points (baseline, 0 min after exercise, and 15 min after exercise) and was expressed as mean ± SD and analysed using one-way repeated-measure ANOVA with Fisher's least significant difference post hoc test. The significance level in this study was set at $P < 0.05$.

4. Results

The interrater and the intrarater intraclass correlation in the measurement of the cPWV were 0.980 and 0.994, respectively. The subject characteristics are summarized in Table 1. At the 0th minute after resistance exercise was performed using Valsalva manoeuvre during exertion, a significant increase in cPWV (4.91 m/s ± 0.52) compared with the baseline cPWV (4.67 m/s ± 0.32, $P = 0.008$) was observed, and the value then returned to a value near the baseline value at the 15th minute after exercise (4.66 m/s ± 0.44, $P = 0.010$) (Table 2). However, although a small increase in cPWV was observed at the 0th minute after performing resistance exercise by using the exhalation technique during exertion, the rise was shallow and no significant difference in cPWV ($P = 0.156$) among the three time intervals existed (Table 3).

(a) (b)

Figure 2: *Ultrasound Scanning of the Carotid Artery*. The setup of the ultrasound measurement of cPWV at a site 1 cm proximal to the bifurcation of the left common carotid artery (a). The measurement interface of RFQAS program (b).

Table 1: The subject characteristics in the study.

Number of subjects	18
Variables	Mean ± SD
Age (yr)	21 ± 1
Weight (kg)	56.0 ± 7.5
Height (cm)	169 ± 6
Brachial systolic pressure (mm Hg)	107 ± 11
Brachial diastolic pressure (mm Hg)	62 ± 8
1-RM (lbs)	20.5 ± 3.3

Table 2: Change in cPWV before and after resistance exercise was performed using the Valsalva manoeuvre during exertion.

Time interval	Preexercise	Postexercise	
	Baseline	0th min	15th min
cPWV (m/s)	4.67 ± 0.32*	4.91 ± 0.52	4.66 ± 0.44**

cPWV at three predefined time intervals when subjects performed resistance exercise by using Valsalva manoeuvre during exertion. One-way repeated-measure ANOVA with Fisher's least significant difference post hoc test (with adjustment of age, body mass index, and systolic blood pressure) demonstrated that significant difference in cPWV exists between before exercise and at 0th min (* $P = 0.008$) and between 0th min and 15th min after exercise (** $P = 0.010$).

Table 3: Change in cPWV before and after resistance exercise was performed using the exhalation technique during exertion.

Time interval	Preexercise	Postexercise	
	Baseline	0th min	15th min
cPWV (m/s)	4.38 ± 0.42	4.64 ± 0.53	4.57 ± 0.55

cPWV at three predefined time intervals when subjects performed resistance exercise by using exhalation technique during exertion. One-way repeated-measure ANOVA (with adjustment of age, body mass index, and systolic blood pressure) demonstrated that no significant difference in cPWV ($P = 0.156$) exists among the three time intervals.

5. Discussion

Resistance exercise is a frequently used mode of exercise, particularly among young men, because it enhances muscle strength and body shape. Although patients with coronary heart disease are instructed to avoid performing resistance exercise because it could affect their blood pressure profile and cause arterial stiffness [10, 11], recent studies have proposed that performing aerobic exercises supplemented with resistance exercises could improve the health conditions of stable-condition coronary heart disease patients [10, 12]. Therefore, in addition to being considered as a training plan for healthy people, resistance exercise has also been adopted as a complementary component of rehabilitation programs for patients with cardiovascular diseases [11–13].

Numerous parameters are used to describe arterial stiffness, including pulse wave velocity (PWV), the augmentation index (AIx), distensibility coefficient (DC), and compliance coefficient (CC) [14–16]. Currently, all of these parameters can be easily quantified using the ultrasound imaging method, and PWV is used extensively in scientific research related to cardiovascular function [17]. PWV reflects the speed of the pulsatile blood flow travelling a predefined distance [18]. To quantify central arterial stiffness, the carotid-PWV approach is the most favoured approach because it provides arterial stiffness measurements that can be easily reproduced [15, 19–21]. The result of the present study suggested that performing resistance exercise by using the Valsalva manoeuvre modulated central arterial stiffness, but this change was not significant when resistance exercise was performed using the exhalation technique during exertion.

Therefore, although performing resistance exercise by using Valsalva manoeuvre during exertion is useful and beneficial to experienced weight lifters or athletes when lifting heavy weights with the provision of additional strength and support to the body trunk [2, 8], beginners and amateurs should not use Valsalva manoeuvre in resistance exercises during exertion. Arterial stiffness, one of the markers reflecting the health status of vessel walls [16, 20, 22–24], defines the ability of vessel walls to expand and recoil in response to changes in blood pressure [25]. In a healthy elastic arterial system, high-pressure blood flow can be withstood by the arterial vessel wall through distention during each ventricular contraction [5, 25]. However, when the arteries stiffen, the arterial wall lacks the ability to distend sufficiently. A higher blood pressure must then be generated through stronger contractions of the myocardium to provide an equivalent amount of blood. Ultimately, the long-term practice of resistance

exercise by using Valsalva manoeuvre could possibly increase the workload of the myocardium and cause hypertrophy or even promote fibrosis of the myocardium because of the prolonged and frequent abrupt changes in hemodynamics.

6. Conclusion

We prospectively investigated the effect of performing resistance exercise on arterial stiffness by monitoring cPWV for 15 minutes after exercise. An acute and significant increase in central arterial stiffness was observed immediately after the subjects performed resistance exercise by using the Valsalva manoeuvre during exertion. However, a similar trend in which central arterial stiffness did not reach a significant level was observed when the subjects performed resistance exercise using the exhalation technique during exertion. Further clinical study is needed to see whether repeated resistance exercise by using Valsalva manoeuvre during exertion could lead to hypertrophy or fibrosis of the myocardium. In addition, since the scope of the present study was limited to including the male subjects and assessing one muscle group, future studies on arterial stiffness in women and other types of resistance exercise for the quadriceps and other muscle groups are warranted.

Disclosure

The authors hereby declared that the whole paper had not been submitted elsewhere for publication, and all authors have contributed significantly and are in agreement with the content of the paper.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Reliability of Force-Velocity Tests in Cycling and Cranking Exercises in Men and Women

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The present study examined the reliability of the force-velocity relationship during cycling and arm cranking exercises in active males and females. Twenty male and seventeen female physical education students performed three-session tests with legs and three-session tests with arms on a friction-loaded ergometer on six different sessions in a randomized order. The reliability of maximal power (P_{max}), maximal pedal rate (V_0), and maximal force (F_0) were studied using the coefficient of variation (CV), the intraclass correlation coefficient (ICC) and the test-retest correlation coefficient (r). Reliability indices were better for men ($1.74 \leq \text{CV} \leq 4.36$, $0.82 \leq \text{ICC} \leq 0.97$, and $0.81 \leq r \leq 0.97$) compared with women ($2.34 \leq \text{CV} \leq 7.04$, $0.44 \leq \text{ICC} \leq 0.98$, and $0.44 \leq r \leq 0.98$) and in cycling exercise ($1.74 \leq \text{CV} \leq 3.85$, $0.88 \leq \text{ICC} \leq 0.98$, and $0.90 \leq r \leq 0.98$) compared with arm exercise ($2.37 \leq \text{CV} \leq 7.04$, $0.44 \leq \text{ICC} \leq 0.95$, and $0.44 \leq r \leq 0.95$). Furthermore, the reliability indices were high for P_{max} and F_0 whatever the expression of the results (raw data or data related to body dimensions). P_{max} and F_0 could be used in longitudinal physical fitness investigations. However, further studies are needed to judge V_0 reliability.

1. Introduction

Maximal anaerobic power can be measured on friction-loaded cycle ergometers or isokinetic ergometers. Many protocols have been proposed for maximal power measurement: all-out tests against a single load (e.g., the Wingate test) [1, 2], relationship between torque and pedal rate on an isokinetic ergometer [3, 4], relationship between load and peak velocity [5], and force-velocity relationship during a single all-out test against a pure inertial load [6] or inertial + braking load [7–9].

On friction-loaded ergometer, maximal power corresponds to power at peak velocity or is computed during the acceleration phase taking into account the power necessary to increase the flywheel kinetic energy [10]. The relationship between pedal rate (V) and braking force (F) or torque (T) can be described by a linear relationship [3, 5–9, 11]. Linear force-velocity relationships have been described for all-out exercises performed on a cycle ergometer not only with the legs (i.e., cycling exercise) but also with the arms (i.e., cranking exercise). The individual characteristics of the force-velocity or torque-velocity relationship can be defined by two parameters: V_0 (the intercept with the pedal rate axis which has the dimension of a maximal pedal rate) and F_0 or T_0 (the intercepts with the force or torque axis, which have the dimension of a maximal force or a maximal torque). Maximal power (P_{max}) corresponds to an optimal pedal rate (V_{opt}) equal to $0.5V_0$ and an optimal load or torque equal to $0.5F_0$ or $0.5T_0$.

Previous studies reported that P_{max} [8] or peak power during a Wingate test [12–15] are significantly correlated with the percentage of the fast muscle fibers in the vastus lateralis. Furthermore, a significant positive correlation was observed between P_{max} and triceps surae musculotendinous stiffness at relative peak torque corresponding to the optimal cycling rate [16]. On the other hand, the value of V_{opt} during sprint cycling was significantly correlated with vastus lateralis myosin heavy chain II composition in a study comparing old

and young participants [17]. The proportion of fast twitch fibres expressed in terms of cross-sectional area was highly correlated with V_{opt} ($r = 0.88$, $P < 0.001$) [18], and the authors of this latter study suggested that V_{opt} would be the most accurate parameter to explore the fibre composition of the knee extensor muscle from cycling tests. The value of F_0 in cycling depends on the strength and the rate of force development of muscle knee extensors [19]. The Wingate optimal braking force can also be determined from the result of a cycling force-velocity test as this braking force is close to $0.5F_0$ [5, 20].

Therefore, it could be interesting to determine the parameters of the force-velocity relationships (V_0, F_0, or T_0) in addition to P_{max} on a cycle ergometer. Furthermore, the study of the changes in power-velocity relationship during an annual training cycle has been proposed in volleyball players [21], which assumes that the results of the force-velocity tests on cycle ergometers are reliable. The reliability of the cycling all-out tests has mainly been investigated by studying either the test-retest correlation coefficients ($r_{test\text{-}retest}$) or the intraclass correlation coefficient (ICC) or the standard errors of estimations (SEE) or the coefficients of variation (CV) for the indices of maximal power (Wingate peak power or P_{max}) with the different protocols [1–4, 6, 9, 22–27]. In contrast, the reliability of the parameters of the force-velocity relationship (slope, T_0, F_0, and V_0) has been investigated in a few studies, only [4, 6, 26]. Moreover, the validity of the statistical tests in these studies on reliability was probably questionable [28].

In a review on the reliability of power in physical performance tests, Hopkins et al. [29] suggested that nonathletic females might be less reliable than nonathletic males, probably because the nonathletic females may be less physically active than the nonathletic males. Similarly, cranking exercises are probably less familiar than cycling exercises and the effect of familiarisation sessions might be more important for force-velocity tests with the arms.

Thus, the aim of the present study was to examine the reliability of P_{max}, V_0, and F_0 during force-velocity tests. In light of the literature observations, we hypothesized that reliability is lower in women than in men and for cranking force-velocity tests than for cycling tests.

2. Materials and Methods

2.1. Participants. Twenty healthy males (24.20 ± 2.69 years, 1.80 ± 0.06 m, and 76.48 ± 8.93 kg) and seventeen healthy females (23.53 ± 2.12 years, 1.68 ± 0.06 m, and 61.18 ± 9.58 kg) volunteered to participate in this study. The participants were all active physical education students but none of them were familiarized with sprint cycling or arm cranking before participation in the study. Before any data collection, all participants were fully informed of the possible risk and discomfort associated with the experimental procedures and gave written informed consent. The experimental protocol was approved by the Institutional Review Board of the University and carried out according to the guidelines of the Declaration of Helsinki.

2.2. Procedures. The participants performed three session tests with the legs and three session tests with the arms on six different sessions in random order. All the tests were performed within a period of four weeks with at least 48 hours between the sessions. Participants were instructed to avoid any strenuous activity between sessions and to follow their usual diet throughout the experimental period. All tests were performed at the same time of day to minimize the effects of circadian rhythms [30] and with similar standard environmental conditions for all participants (mean temperature and humidity: 22 ± 0.1°C and 35 ± 0.4%, resp.). Body mass and height measures of all subjects were examined before each testing session.

The participants performed a standard warm-up consisting of 5 min cycling (80 W and 50 W for men and women, resp.) before the leg tests or arm cranking (50 W and 20 W for men and women, resp.) for the arm tests, with two short accelerations (3-s) at the end of the third min and the fifth min. After 5 minutes of passive recovery, participants performed the force-velocity test which consisted of repetitive short maximal sprints of 6-s against increasing braking forces. The braking forces administrated at the beginning of the sprints cycling were 2 kg and 1.5 kg for men and women, respectively, while during arm cranking the loads were equal to 1.5 kg and 1 kg for men and women, respectively. Then, the braking force was increased after 5 min of passive recovery (sprints cycling: 2 and 1.5 kg for men and women, resp.; arm cranking: 1.5 and 1 kg for men and women, resp.) until the participant was unable to reach a peak velocity higher than 100 rpm. The same order of braking force application was respected across session tests.

All force-velocity tests were performed on a friction-loaded cycle ergometer with weights (Monark 864, Monark Exercise AB, Vansbro, Sweden) adjustable for both leg and arm exercises [31, 32]. During sprint cycling exercises, participants were seated on the cycle ergometer equipped with toe clips and well-fastened straps to avoid losing the pedals. The same riding position was used throughout the study. Participants were instructed to cycle in seated position to avoid the effect of postural changes [33–35]. During arm cranking exercises, the pedals were replaced with handles and the cycle ergometer was fixed on a metal frame. The participants were standing on their feet in front of the ergometer during the exercises. The center of the pedal axis was approximately 20 cm lower than the level of the shoulder axis. All sprints were performed from the same initial pedal position. Participants were encouraged by the same investigator to reach the maximal velocity rate as quickly as possible. Instantaneous pedal rate in cycling or cranking was monitored throughout a PC computer by means of an encoder placed on the cycle ergometer flywheel. Then, the velocity was averaged over 1-s intervals.

The peak velocity (V) was measured for each braking force (F) and was used to calculate the linear force-velocity relationship for cycling exercises according to the least squares method:

$$V = a - bF. \quad (1)$$

The above relationship was transformed as follows [33]:

$$V = V_0\left(1 - \frac{F}{F_0}\right). \quad (2)$$

In this equation, V_0 and F_0 corresponded to the intercepts with the velocity axis and force axis, respectively ($V_0 = a$ and $F_0 = a/b$). Since a linear relationship between F and V was assumed, $P_{\max}$ corresponded to an optimal velocity and an optimal braking force equal to $0.5V_0$ and $0.5F_0$, respectively. Hence, $P_{\max}$ was calculated as follows [5, 33]:

$$P_{\max} = 0.5V_0 \times 0.5F_0 = 0.25V_0F_0. \quad (3)$$

The performance variables were expressed in absolute units and according to dimensional scaling. V_0 was expressed in absolute unit (rpm) and relative to body height ($\text{rpm}\cdot\text{BH}^{-1}$). F_0 was expressed in absolute unit (kg) and relative to body mass raised to the power of 0.67 ($\text{kg}\cdot\text{BM}^{-0.67}$). $P_{\max}$ was expressed in absolute unit (W) and relative to body mass ($\text{W}\cdot\text{BM}^{-1}$).

2.3. Relation between the Variabilities of F_0 and V_0. The variability of F_0 and V_0 between the second and first sessions ($\Delta F_{0\ 2\text{-}1}$ and $\Delta V_{0\ 2\text{-}1}$) and between the third and second sessions ($\Delta F_{0\ 3\text{-}2}$ and $\Delta V_{0\ 3\text{-}2}$) was calculated according to the following formulas:

$$\begin{aligned}
\Delta F_{0\ 2\text{-}1} &= 100\frac{F_{02}}{F_{01}},\\
\Delta F_{0\ 3\text{-}2} &= 100\frac{F_{03}}{F_{02}},\\
\Delta V_{0\ 2\text{-}1} &= 100\frac{V_{02}}{V_{01}},\\
\Delta V_{0\ 3\text{-}2} &= 100\frac{V_{03}}{V_{02}}.
\end{aligned} \quad (4)$$

2.4. Statistical Analyses. Statistical procedures were carried out using Statistica 7.1 Software (StatSoft, France). Data of V_0, F_0, and $P_{\max}$ are presented as mean and standard deviation (mean ± SD). Before statistical analysis, each performance variable was tested for normality with the Shapiro-Wilk test. With the assumption of normality confirmed, systematic change in performance from trials 1 to 3 was examined using one-way ANOVA with repeated measures and a Tukey's post hoc test. All significance thresholds were set at $P < 0.05$.

Absolute reliability, which concerns the consistency of individual's scores [36], was determined using the standard error of measurement SEM and the coefficient of variation (CV) using the following formulas [37]:

$$\begin{aligned}
\text{SEM} &= \frac{\text{SD}_{\text{diff}}}{\sqrt{2}},\\
\text{CV}\ (\%) &= \frac{\text{SEM}}{\text{Mean}} \times 100,
\end{aligned} \quad (5)$$

where SD_{diff} was the standard deviation of the differences between consecutive session tests (i.e., sessions 1 and 2 and sessions 2 and 3).

Relative reliability, which concerns the consistency of individual's position in the group relative to others [36], was assessed using the intraclass correlation coefficient of two-way random effects model with single measure for each pair of consecutive session tests (i.e., sessions 1 and 2 and sessions 2 and 3) as follows:

$$\text{ICC}\ (2,1) = \frac{\text{MS}_P - \text{MS}_E}{\text{MS}_P + \text{MS}_E + 2\left(\text{MS}_T - \text{MS}_E\right)/n}. \quad (6)$$

In this formula MS_P represents the participant mean square, MS_E represents the error mean square, k is the number of trials, MS_T represents the trials mean square, and n is the number of participants. The ICC is considered as high for values above 0.90, moderate for values between 0.80 and 0.90, and low for values below 0.80 [38].

In addition, the test-retest correlation coefficient ($r_{\text{test-retest}}$) was calculated for each pair of consecutive session tests in order to compare the results of the present study to the data in the literature [29]. The Bland-Altman plots were used to check for heteroscedasticity [28].

3. Results

3.1. Variations in Body Mass (BM). For the arm tests, the differences in BM between the sessions were equal to −0.08 ± 0.754 ($\Delta S2 - S1$), 0.305 ± 0.669 ($\Delta S3 - S2$), and 0.225 ± 0.916 kg ($\Delta S3 - S1$) in men and 0.129 ± 0.512 ($\Delta S2 - S1$), 0.006 ± 0.553 ($\Delta S3 - S2$), and 0.124 ± 0.529 kg ($\Delta S3 - S1$) in women.

For the leg tests, the differences in BM between the sessions were equal to 0.090 ± 0.704 ($\Delta S2 - S1$), 0.255 ± 0.737 ($\Delta S3 - S2$), and 0.345 ± 0.944 kg ($\Delta S3 - S1$) in men and 0.288 ± 0.499 ($\Delta S2 - S1$), −0.206 ± 0.536 ($\Delta S3 - S2$), and 0.08 ± 0.591 kg ($\Delta S3 - S1$) in women.

3.2. V_0, F_0, and $P_{\max}$ in the Three Sessions. The individual values of F_0 and V_0 measured in the three sessions are presented in Figure 1. The branches of hyperbolae (i.e., continuous and dashed curves) in Figure 1 correspond to the participants with different combinations of F_0 and V_0 but the same value of $P_{\max}$. The means ± SD and ranges of $P_{\max}$, F_0, V_0, $P_{\max}\cdot\text{BM}^{-1}$, $F_0\cdot\text{BM}^{-1}$, $F_0\cdot\text{BM}^{-0.67}$, and $V_0\cdot\text{BH}^{-1}$ measured in the different sessions are presented in Tables 1 and 2 and Figures 1 and 2. In Table 1 and Figure 1, BM corresponded to the body mass measured during each session whereas BM was equal to the average of the three measures of BM in Figure 2.

All the differences between men and women were highly significant ($P < 0.001$) even when the data were related to body mass ($P_{\max}\cdot\text{BM}^{-1}$, $F_0\cdot\text{BM}^{-1}$, and $F_0\cdot\text{BM}^{-0.67}$). The significance level of the difference in $V_0\cdot\text{BH}^{-1}$ between men and women was equal to $P < 0.05$, only.

3.3. Reliability. The one-way ANOVA with repeated measure showed a significant main effect of trial on V_0 in men ($F_{(2,38)} = 11.48$, $P < 0.001$ and $F_{(2,38)} = 6.93$, $P < 0.01$, for cycling

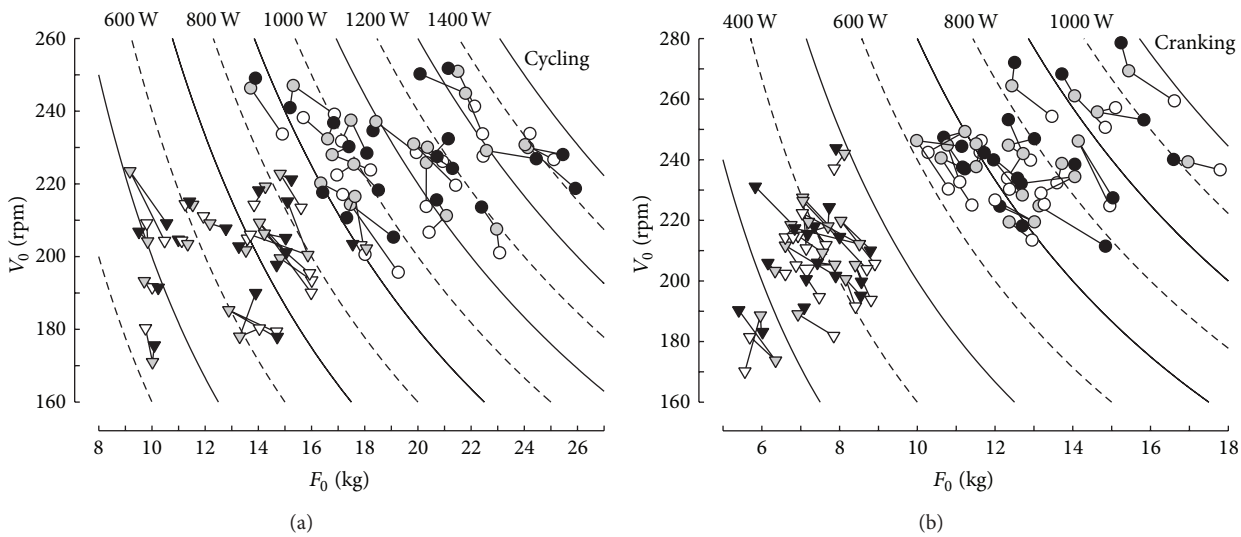

Figure 1: Individual values of V_0 and F_0 corresponding to the force-velocity relationships in cycling (a) and cranking (b) at the first (empty symbols), second (grey symbols), and third (black symbols) trials. The three values of each participant are linked by broken lines. Circles and triangles correspond to men and women, respectively.

Table 1: Parameters P_{max}, F_0, and V_0 (means, SD, and range) computed from the force-velocity tests performed with legs or arms by men in sessions 1, 2, and 3.

			Session 1	Session 2	Session 3
Legs	V_0	rpm	223 ± 14 (196–241)	230 ± 12 (208–251)	228 ± 13 (205–242)
		rpm·BH^{-1}	1.24 ± 0.09 (1.08–1.38)	1.28 ± 0.08 (1.11–1.43)	1.27 ± 0.08 (1.14–1.45)
	F_0	kg	19.8 ± 2.9 (14.9–25.1)	19.3 ± 3.0 (13.7–24.1)	19.7 ± 3.3 (13.9–25.9)
		kg·BM^{-1}	0.26 ± 0.03 (0.21–0.32)	0.25 ± 0.03 (0.21–0.31)	0.26 ± 0.03 (0.22–0.31)
		kg·BM$^{-0.67}$	1.09 ± 0.12 (0.89–1.30)	1.06 ± 0.13 (0.88–1.29)	1.07 ± 0.13 (0.88–1.33)
	P_{max}	W	1105 ± 174 (871–1423)	1107 ± 173 (844–1387)	1122 ± 182 (865–1451)
		W·BM^{-1}	14.5 ± 1.8 (11.4–17.7)	14.5 ± 1.6 (11.4–17.6)	14.6 ± 1.5 (11.8–17.8)
Arms	V_0	rpm	237 ± 12 (213–259)	243 ± 14 (219–269)	242 ± 17 (211–279)
		rpm·BH^{-1}	1.32 ± 0.08 (1.21–1.46)	1.35 ± 0.08 (1.24–1.48)	1.35 ± 0.09 (1.20–1.52)
	F_0	Kg	13.1 ± 1.9 (10.2–17.8)	12.9 ± 1.7 (10.0–17.0)	13.1 ± 1.7 (10.7–16.6)
		kg·BM^{-1}	0.17 ± 0.02 (0.14–0.21)	0.17 ± 0.02 (0.13–0.20)	0.17 ± 0.02 (0.14–0.21)
		kg·BM$^{-0.67}$	0.71 ± 0.08 (0.57–0.86)	0.70 ± 0.07 (0.55–0.81)	0.71 ± 0.07 (0.60–0.86)
	P_{max}	W	777 ± 136 (620–1077)	781 ± 122 (615–1039)	792 ± 123 (660–1061)
		W·BM^{-1}	10.1 ± 1.2 (8.2–12.1)	10.2 ± 1.1 (8.2–12.3)	10.3 ± 1.1 (8.4–12.0)

and cranking, resp.) and women ($F_{(2,32)} = 4.55$, $P < 0.05$ and $F_{(2,32)} = 6.10$, $P < 0.01$, for cycling and cranking, resp.). Tukey's post hoc tests revealed that V_0 at session 1 was significantly lower by comparison to sessions 2 and 3. In contrast, there was no significant main effect of sessions on F_0 and P_{max} for arms and legs in men and women ($P > 0.05$).

The CV (%) of V_0, F_0, and P_{max} are presented in Tables 3 and 4. The highest CV values were obtained for F_0 by comparison with V_0 and P_{max}. The greatest CV values were observed for cranking exercises in female participants.

The values of $r_{test\text{-}retest}$ are presented in Tables 3 and 4. The values of $r_{test\text{-}retest}$ increased for the correlations between sessions 2 and 3 when compared with the correlations between sessions 1 and 2. Except F_0 with the arms in women, the lowest $r_{test\text{-}retest}$ were observed for V_0.

For the correlations between the results of the first and second sessions, the values of $r_{test\text{-}retest}$ for F_0 were significantly different between cycling and cranking but in the female group, only ($P = 0.030$ for F_0; $P = 0.036$ for F_0 related to BM$^{-0.67}$). Similarly, the values of $r_{test\text{-}retest}$ between the first and second sessions were significantly different between male and female groups for F_0 and P_{max} ($P = 0.007$ for F_0, $P = 0.005$ for F_0 related to BM$^{-0.67}$, and $P = 0.047$ for P_{max} in watts). For the correlations between the results of the second and third sessions, the values of $r_{test\text{-}retest}$ for F_0 and P_{max} were significantly different between cycling and cranking but in the female group, only ($P = 0.01$ for F_0; $P = 0.006$ for F_0 related to BM$^{-0.67}$ and $P = 0.023$ for P_{max} in watts). All the other comparisons of $r_{test\text{-}retest}$ between men and women or cycling and cranking were not significantly different.

Table 2: Parameters P_{max}, F_0, and V_0 (means, SD, and range) computed from the force-velocity tests performed with legs or arms by women in sessions 1, 2, and 3.

			Session 1	Session 2	Session 3
Legs	V_0	rpm	200 ± 12 (179–215)	203 ± 15 (171–223)	203 ± 13 (176–221)
		rpm·BH^{-1}	1.19 ± 0.09 (1.00–1.35)	1.21 ± 0.10 (1.04–1.36)	1.21 ± 0.09 (1.00–1.32)
	F_0	kg	13.3 ± 2.6 (9.8–17.9)	12.9 ± 2.4 (9.2–18.1)	13.2 ± 2.3 (9.5–17.5)
		kg·BM^{-1}	0.22 ± 0.02 (0.19–0.25)	0.21 ± 0.02 (0.16–0.26)	0.21 ± 0.02 (0.19–0.25)
		kg·BM$^{-0.67}$	0.84 ± 0.09 (0.69–1.01)	0.82 ± 0.10 (0.69–1.01)	0.83 ± 0.09 (0.69–0.99)
	P_{max}	W	662 ± 130 (430–907)	655 ± 136 (428–914)	668 ± 131 (443–893)
		W·BM^{-1}	10.8 ± 1.1 (8.3–12.4)	10.7 ± 1.4 (7.5–12.9)	10.9 ± 1.3 (8.4–13.3)
Arms	V_0	rpm	203 ± 17 (170–237)	210 ± 16 (174–242)	209 ± 16 (183–244)
		rpm·BH^{-1}	1.21 ± 0.11 (1.03–1.37)	1.25 ± 0.10 (1.07–1.41)	1.25 ± 0.10 (1.41–1.10)
	F_0	kg	7.4 ± 1.0 (5.6–9.0)	7.3 ± 0.8 (6.0–8.5)	7.3 ± 1.0 (5.4–8.7)
		kg·BM^{-1}	0.12 ± 0.01 (0.10–0.14)	0.12 ± 0.01 (0.09–0.15)	0.12 ± 0.01 (0.09–0.14)
		kg·BM$^{-0.67}$	0.47 ± 0.04 (0.39–0.54)	0.47 ± 0.04 (0.38–0.55)	0.46 ± 0.05 (0.38–0.54)
	P_{max}	W	375 ± 61 (237–466)	386 ± 59 (276–491)	380 ± 63 (257–482)
		W·BM^{-1}	6.2 ± 0.8 (4.6–7.7)	6.4 ± 0.8 (5.1–7.6)	6.3 ± 0.9 (4.9–7.7)

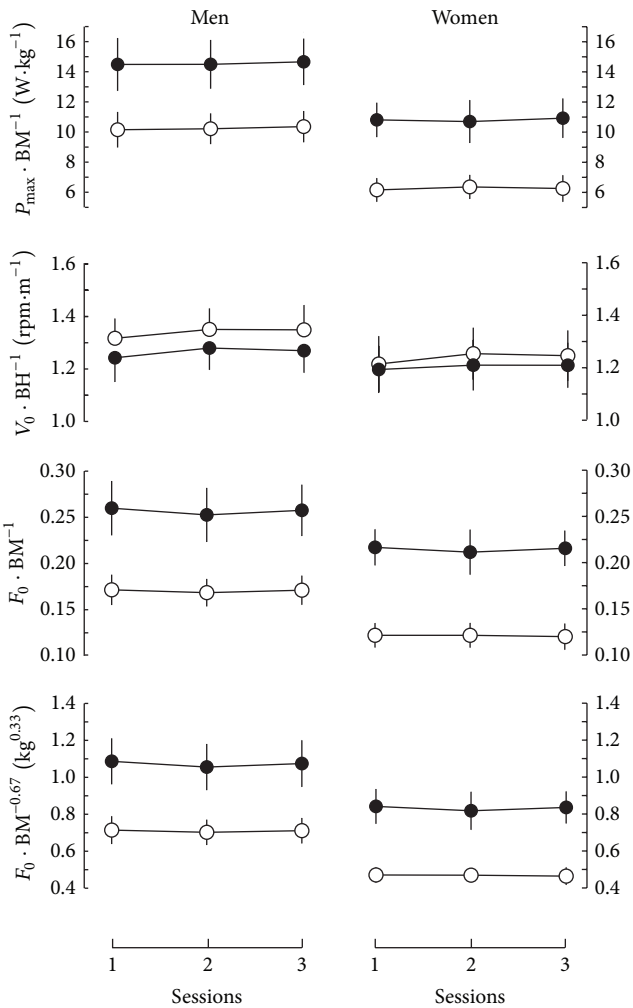

Figure 2: Results of the force-velocity tests (means ± SD) in the three sessions related to body dimensions (F_0 related to BM and BM$^{-0.67}$, V_0 related to BH, and P_{max} related to BM). Black points, exercises performed with the legs; empty circle, exercises performed with the arms.

Table 3: Differences between sessions 1 and 2; coefficients of variation (CV), intraclass correlation coefficients (ICC), and test-retest correlation coefficients ($r_{\text{test-retest}}$) for V_0, F_0, and $P_{\max}$ for the leg or arm force-velocity tests in men and women.

			Men		Women	
			Legs	Arms	Legs	Arms
SEM	V_0	rpm	4.28	5.30	5.80	6.67
		rpm·BH^{-1}	0.02	0.03	0.03	0.04
	F_0	kg	0.59	0.48	0.73	0.58
		kg·BM$^{-0.67}$	0.03	0.03	0.05	0.04
	$P_{\max}$	W	29.10	24.9	24.5	21.7
		W·BM^{-1}	0.38	0.32	0.41	0.35
CV (%)	V_0	rpm	1.89	2.21	2.88	3.23
		rpm·BH^{-1}	1.89	2.25	2.90	3.23
	F_0	kg	3.01	3.69	5.60	7.84
		kg·BM$^{-0.67}$	2.95	3.75	5.50	7.52
	$P_{\max}$	W	2.63	3.19	3.71	5.69
		W·BM^{-1}	2.61	3.18	3.83	5.60
ICC	V_0	rpm	0.79	0.75	0.80	0.78
		rpm·BH^{-1}	0.93	0.78	0.85	0.80
	F_0	kg	0.95	0.93	0.91	0.60
		kg·BM$^{-0.67}$	0.91	0.86	0.77	0.25
	$P_{\max}$	W	0.97	0.96	0.97	0.86
		W·BM^{-1}	0.95	0.93	0.90	0.79
$r_{\text{test-retest}}$	V_0	rpm	0.89	0.84	0.82	0.84
		rpm·BH^{-1}	0.93	0.85	0.86	0.85
	F_0	kg	0.96	0.94	0.91	0.60
		kg·BM$^{-0.67}$	0.94	0.87	0.79	0.24
	$P_{\max}$	W	0.97	0.97	0.97	0.87
		W·BM^{-1}	0.95	0.93	0.92	0.80

The ICC of each performance variable across sessions 1 and 2 and sessions 2 and 3 in male and female participants are presented in Tables 3 and 4. The values of ICC improved for sessions 2 and 3 by comparison with sessions 1 and 2. Excepting F_0 with the arms in female participants, the lowest ICC values were observed for V_0.

3.4. Relation between the Variabilities of F_0 and V_0. The variability of F_0 ($\Delta F_{0\ 2\text{-}1}$ or $\Delta F_{0\ 3\text{-}2}$) was significantly correlated with the variability of V_0 ($\Delta V_{0\ 2\text{-}1}$ or $\Delta V_{0\ 3\text{-}2}$) as shown in Figure 3:

in women:

$$\begin{aligned}
\Delta F_0\text{arms}_{2\text{-}1} &= 263 - 1.57\,\Delta V_0\text{arms}_{2\text{-}1},\\
r &= 0.695; \quad P = 0.002,\\
\Delta F_0\text{arms}_{3\text{-}2} &= 274 - 1.76\,\Delta V_0\text{arms}_{3\text{-}2},\\
r &= 0.742; \quad P < 0.001,\\
\Delta F_0\text{legs}_{2\text{-}1} &= 235 - 1.36\,\Delta V_0\text{legs}_{2\text{-}1},\\
r &= 0.773; \quad P < 0.001,\\
\Delta F_0\text{legs}_{3\text{-}2} &= 215 - 1.12\,\Delta V_0\text{legs}_{3\text{-}2},\\
r &= 0.644; \quad P = 0.005,
\end{aligned} \tag{7}$$

in men:

$$\begin{aligned}
\Delta F_0\text{arms}_{2\text{-}1} &= 184 - 0.83\,\Delta V_0\text{arms}_{2\text{-}1},\\
r &= 0.503; \quad P = 0.024,\\
\Delta F_0\text{arms}_{3\text{-}2} &= 219 - 1.17\,\Delta V_0\text{arms}_{3\text{-}2},\\
r &= 0.624; \quad P = 0.003,\\
\Delta F_0\text{legs}_{2\text{-}1} &= 184 - 0.83\,\Delta V_0\text{legs}_{2\text{-}1},\\
r &= 0.503; \quad P = 0.024,\\
\Delta F_0\text{legs}_{3\text{-}2} &= 219 - 1.17\,\Delta V_0\text{legs}_{3\text{-}2},\\
r &= 0.624; \quad P = 0.003.
\end{aligned} \tag{8}$$

4. Discussion

In the present investigation, we studied the reliability of $P_{\max}$, V_0, and F_0 during cycling and arm cranking exercises in active men and women. In order to study the reliability

Table 4: Differences between sessions 2 and 3; coefficients of variation (CV), intraclass correlation coefficients (ICC), and test-retest correlation coefficients ($r_{\text{test-retest}}$) for V_0, F_0, and $P_{\max}$ for the leg or arm force-velocity tests in men and women.

			Men		Women	
			Legs	Arms	Legs	Arms
SEM	V_0	rpm	3.97	5.74	4.76	6.01
		rpm·BH^{-1}	0.02	0.03	0.03	0.04
	F_0	kg	0.65	0.56	0.50	0.51
		kg·BM$^{-0.67}$	0.01	0.03	0.03	0.03
	$P_{\max}$	W	29.8	26.3	19.1	20.6
		W·BM^{-1}	0.38	0.32	0.27	0.33
CV	V_0	rpm	1.74	2.37	2.35	2.87
		rpm·BH^{-1}	1.74	2.37	2.34	2.91
	F_0	kg	3.34	4.36	3.85	7.01
		kg·BM$^{-0.67}$	3.26	4.21	3.56	7.04
	$P_{\max}$	W	2.67	3.35	2.88	5.37
		W·BM^{-1}	2.63	3.16	2.50	5.17
ICC	V_0	rpm	0.90	0.87	0.88	0.86
		rpm·BH^{-1}	0.93	0.87	0.91	0.87
	F_0	kg	0.95	0.89	0.95	0.69
		kg·BM$^{-0.67}$	0.92	0.82	0.90	0.44
	$P_{\max}$	W	0.97	0.95	0.98	0.89
		W·BM^{-1}	0.94	0.92	0.95	0.85
$r_{\text{test-retest}}$	V_0	rpm	0.90	0.88	0.89	0.86
		rpm·BH^{-1}	0.93	0.88	0.91	0.86
	F_0	kg	0.96	0.89	0.96	0.70
		kg·BM$^{-0.67}$	0.92	0.81	0.92	0.44
	$P_{\max}$	W	0.97	0.95	0.98	0.89
		W·BM^{-1}	0.94	0.92	0.97	0.85

of these parameters, force-velocity tests on cycle ergometer were separately repeated three times in different sessions for each exercise. It was assumed that reliability was lower (1) in women than in men and (2) for cranking force-velocity tests than for cycling tests. The results of the present study were in agreement with this hypothesis: the reliability indices were better for the men and the leg indices when compared with the women and arm indices (Tables 3 and 4). Whatever the force-velocity parameter (V_0, F_0, and $P_{\max}$), familiarisation sessions might be more important for women and arm tests as indicated by the lower values of CV in men and leg tests when the results of the first and second sessions were compared (Table 3).

The reliability of $P_{\max}$ was similar to the reliability of the different indices of maximal power in previous studies. For example, the reliability of the results of the Wingate is good for the peak power ($r_{\text{test-retest}} > 0.90$) and the mean power ($r_{\text{test-retest}}$ between 0.91 and 0.93) [1, 2, 22], in contrast with the reliability of the fatigue index ($r_{\text{test-retest}} = 0.43$). Similarly, the reliability of the power indices measured with the different force-velocity protocols was high when measured with isokinetic cycle ergometers [3, 4, 9], friction-loaded ergometers [23, 24, 26], or the inertial load method [6, 25]. In a study by Winter et al. [23], the maximal power computed during the acceleration phase (PP_{corr}) estimated according to Lakomy [10] was 10% higher than $P_{\max}$ but the reliability of PP_{corr} was lower ($r_{\text{test-retest}}$: 0.530 for PP_{corr} versus 0.972 for $P_{\max}$ in men, and 0.922 for PP_{corr} versus 0.952 for $P_{\max}$ in women). In the same study of Winter et al. [23], the CV values of PP_{corr} were higher in men (6.9% for PP_{corr} versus 2.7% for $P_{\max}$) but not in women (3.7% for PP_{corr} versus 4.2% for $P_{\max}$). Furthemore, according to Winter et al. [23], these results of optimization procedures (i.e., the method of Vandewalle et al. [5]) add further support and have securer fundations than those enjoyed by correction procedures [10]. For arm exercises, Smith et al. [39] reported CV values of 4.5% for PP_{corr} and 2.8% for $P_{\max}$. It is likely that the lower reliability of PP_{corr} is explained by oscillations of P_{corr} (product of V and F_{corr} that takes into account not only the braking force but also the force necessary for the flywheel acceleration). On isokinetic cycle ergometers, the coefficients of variation of the slope and intercept of

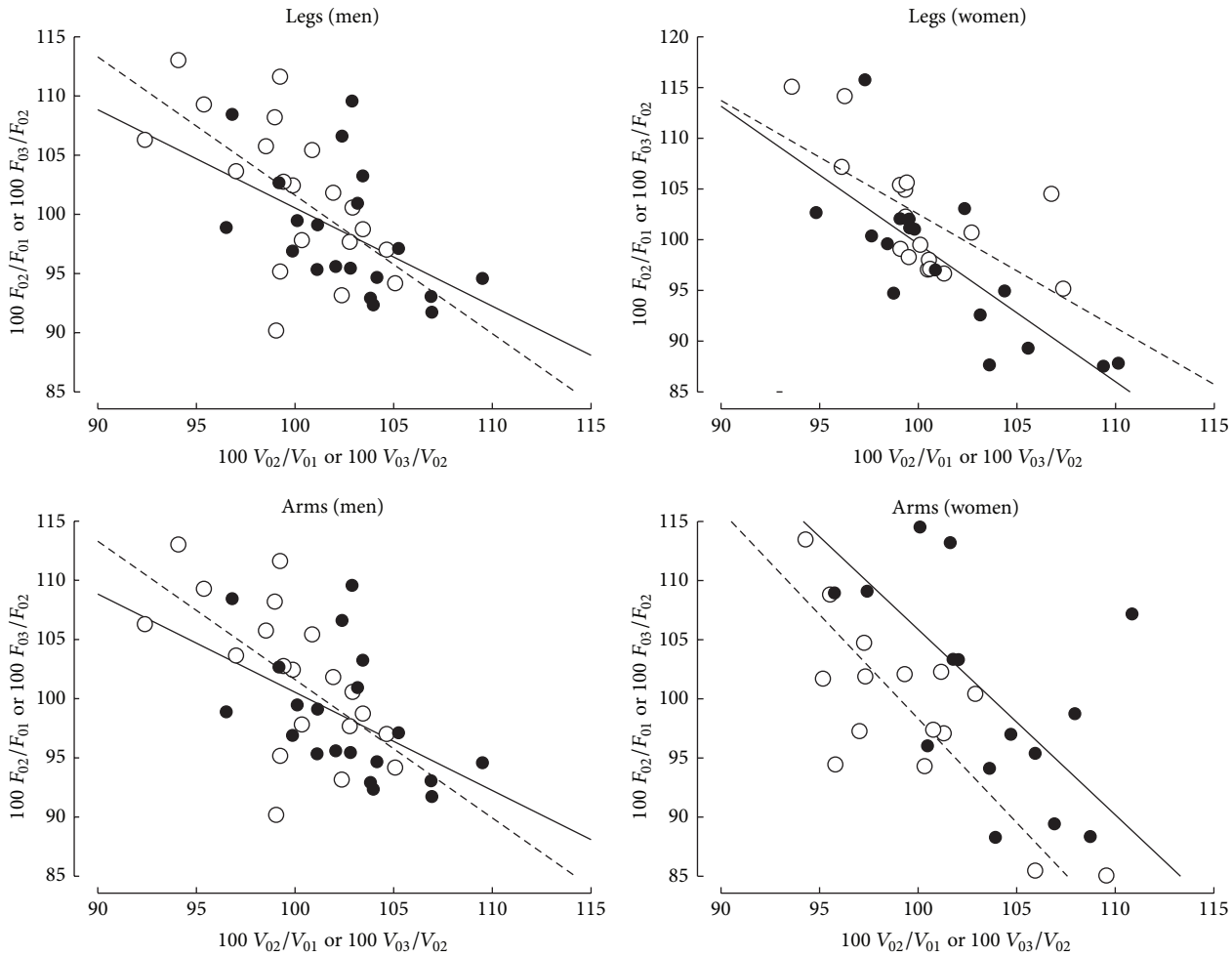

Figure 3: Relationships between intersession differences in F_0 (ordinates) and intersession differences in V_0 (abscissae) for the leg and arm force-velocity tests in men and women. Continuous lines and black points: differences between the first and second sessions. Dashed line and empty circles: differences between the second and third sessions.

the regression between torque and pedal rate were 13.7 and 10.5%, respectively [4].

The values of CV of V_0, F_0, and P_{max} in the present study were similar to the values of CV for the different parameters measured with the inertial method (4 trials on the same day): 3.3% for PP_{corr}, 2.7% for V_0, and 4.4% for T_0 [6]. For friction-loaded ergometers, the reliability of the force-velocity parameters in cycling has been tested in male physical education students [26]. For F_0 and P_{max}, SEE was lower than 5% and $r_{test\text{-}retest}$ or ICC were higher than 0.90 as in the present study for the cycling force-velocity test in the male participants. However, the comparison and the validity of the reliability indices must take into account the characteristics of the data [28, 37]. The data are said to be homoscedastic when the random error does not depend on the size of the measured value. Homoscedastic errors are generally expressed in the same units as those of their measurements and they can be analysed with conventional parametric analyses. SEM is valid when the data are homoscedastic. The data are said to be heteroscedastic when the random error increases as the measured values increase. Heteroscedastic data should be measured on a ratio scale (e.g., percentage) and be investigated with an analysis based on nonparametric analyses (i.e., rank tests). CV is valid even when the data are heteroscedastic. The heterogeneity of values between participants influences the results of the reliability tests.

(1) The coefficient of correlation of test-retest ($r_{test\text{-}retest}$) is sensitive to the heterogeneity of data between participants.

(2) The effect of heteroscedascity on the observed "errors" in a test-retest is low when the data range is narrow.

The spread of the data between participants is different for V_0, F_0, and P_{max} expressed in percentage of the group averages even when they are related to body dimension (Table 5). Heteroscedasticity was expected for V_0, F_0, and P_{max} raw data. However, this expectation was not confirmed with Bland-Altman plots of these data, especially in men (Figure 4). The data ranges of parameters V_0, F_0, and P_{max} were lower than 62% in men (Table 5), which could partly explain that heteroscedascity was not suggested by the Bland-Altman plots of V_0, F_0, and P_{max} raw data (Figure 4). In

Table 5: Ranges of parameters V_0, F_0, and P_{max} expressed in percentage of the means of the male or female groups.

		Men			Women		
		Session 1	Session 2	Session 3	Session 1	Session 2	Session 3
Legs	V_0	20.5	18.9	20.4	17.9	25.9	22.6
	$V_0 \cdot BH^{-1}$	24.7	25.2	24.1	28.9	26.4	27.0
	F_0	51.5	54.0	61.0	61.0	68.9	61.0
	$F_0 \cdot BM^{-0.67}$	38.1	38.6	41.8	37.7	43.6	35.9
	P_{max}	50.0	49.0	52.3	70.5	74.1	67.3
	$P_{max} \cdot BM^{-1}$	43.5	42.4	40.9	37.7	51.0	45.3
Arms	V_0	19.4	20.6	27.7	32.9	32.5	29.1
	$V_0 \cdot BH^{-1}$	19.5	18.0	24.5	27.9	27.6	24.6
	F_0	57.6	54.3	45.3	46.4	34.7	45.8
	$F_0 \cdot BM^{-0.67}$	40.9	37.0	37.1	30.6	36.7	33.8
	P_{max}	58.8	54.4	50.7	61.1	55.7	59.1
	$P_{max} \cdot BM^{-1}$	37.7	40.1	35.9	51.0	39.9	45.1

women, the data ranges were larger than in men when the ranges were expressed as percentages of the means (Table 5) but the correlations of the absolute values of the differences versus the means of the results in the first and second sessions (Figure 5) were not significant. All other things being equal, the differences between sessions are probably lower in well-motivated individuals and experts in cycling and the average of their performances in sessions 1 and 2 should be higher (and inversely for the nonexperts and not motivated individuals). Therefore, the effects of motivation and expertise can alter the results of the Bland-Altman plot in this kind of physical tests.

As in the study by Attiogbé et al. [26], the values of $r_{test\text{-}retest}$ and ICC were lower for V_0 than for F_0 and P_{max}, which can be partly explained by the smaller variance of this parameter. Indeed, the range of V_0 is smaller (Table 5) than the range of F_0 and P_{max}. The small variance of V_0 in the present study is probably an expression of the small variance of V_0 when compared with the variances of F_0 and P_{max} in a general athletic population [35]. The small range of V_0 also probably explains that the values of CV in men and women were lower for V_0 than for F_0 and P_{max} in the cycling as well as the cranking force-velocity tests. Excepting the study by Buśko [21], there is no data about the changes in V_0 during an annual training cycle and, therefore, it is difficult to know whether its reliability is good enough for the estimation of the training effect on this parameter.

The ranges of F_0 and P_{max} were similar but the values of $r_{test\text{-}retest}$ or ICC were higher for P_{max} than for F_0 (and V_0). It is likely that the variations in V_0 and F_0 between sessions are not totally independent (Figure 3). Indeed, the values of V_0 and F_0 are extrapolated from the relationship between braking force and peak velocity. An underestimation of the peak velocity corresponding to the highest braking force induces a rotation of the F-V regression line (i.e., a more negative slope) and, consequently, an overestimation of V_0 in addition to an underestimation of F_0. Inversely, an underestimation of the peak velocity corresponding to the lowest braking force induces a less negative slope of the F-V regression line and, consequently, and overestimation of F_0 in addition to an underestimation of V_0. The value of P_{max} depends on F_0 and V_0 and the effect of an underestimation of V_0 on P_{max} should be compensated by the effect of an overestimation of F_0, and *vice versa*. This could partly explain why the values of $r_{test\text{-}retest}$, ICC, or CV were better for P_{max} than for F_0.

The values of V_0, F_0, and P_{max} were lower in women than in men. The differences in BH and BM were not the only explanations of the lower values of V_0, F_0, and P_{max} in women. Indeed, these differences were still significant when force-velocity parameters were related to BH or BM ($V_0 \cdot BH^{-1}$, $F_0 \cdot BM^{-0.67}$, and $P_{max} \cdot BM^{-1}$). This gender effect could partly be explained by a difference in muscle fiber composition as, for example, the higher percentage of the cross-sectional area that corresponds to the slow fibers in women [40–42]. The lower values of $F_0 \cdot BM^{-0.67}$, $F_0 \cdot BM^{-1}$, and $P_{max} \cdot BM^{-1}$ might partly be explained by a lower percentage of lean body mass in women. The lower values of $r_{test\text{-}retest}$ in women cannot be explained by a lower range of the individual data (Table 5). The lower reliability in women might partly be explained by the effect of menstrual cycle, but it is possible that this effect is less important in trained women because training might reduce the cyclical hormonal fluctuations [29].

The variability of F_0 and P_{max} depends on the variability of BM when these data are related to body mass ($F_0 \cdot BM^{-1}$, $F_0 \cdot BM^{-0.67}$, and $P_{max} \cdot BM^{-1}$). In spite of the instructions about diet, hydration, and training, the standard deviations of the differences in BM between the sessions were not negligible (<1.25% of BM).

5. Methodological Considerations

To the best of our knowledge, this is the first study examining the reliability of force-velocity tests on cycle ergometer during sprint cycling and arm cranking exercises in active men and women. One of the limitations inherent to the experimental

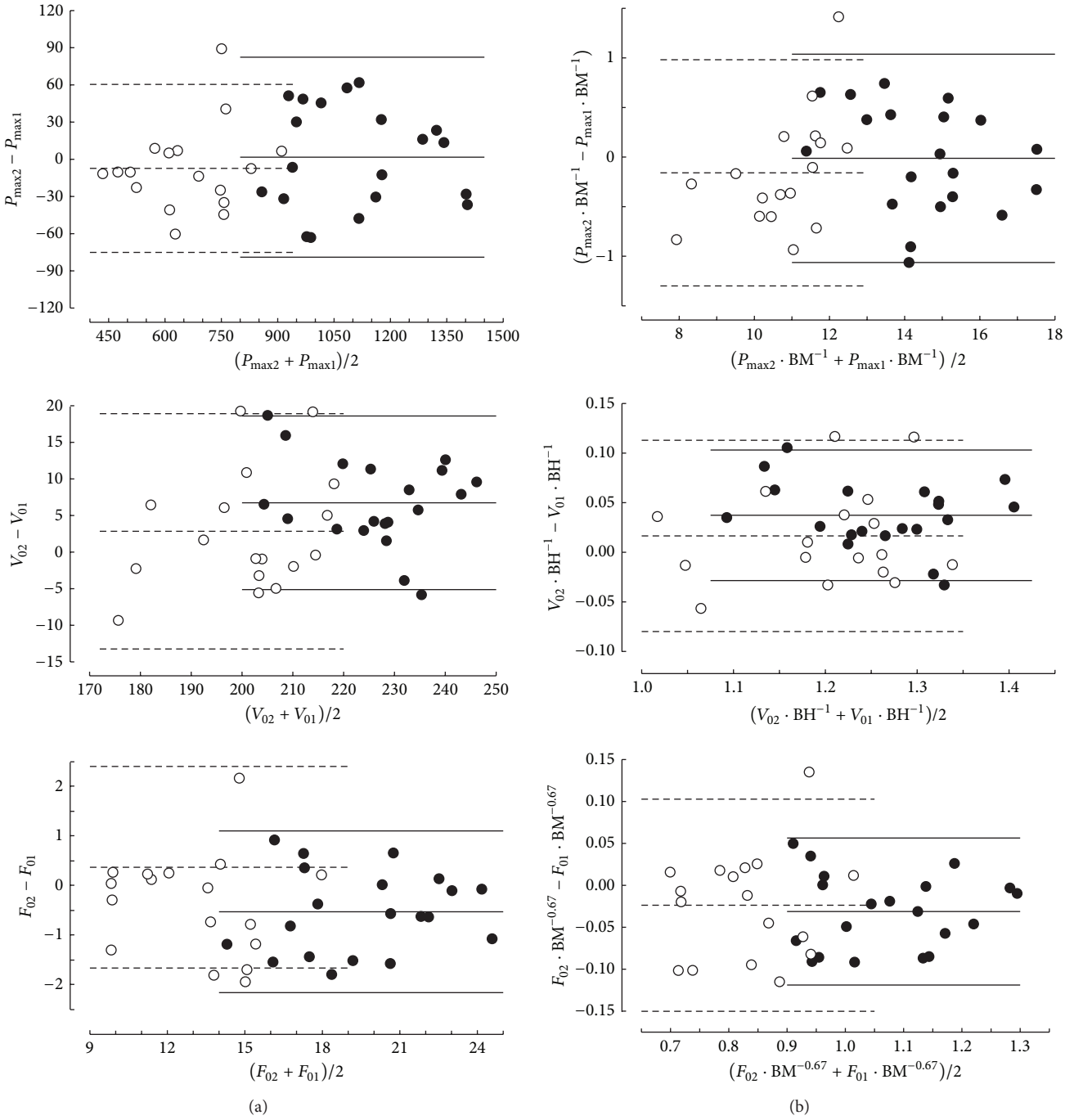

Figure 4: Bland and Altman plots of the results of differences in parameters P_{max}, V_0, and F_0 ((a) raw data; (b) data related to body dimensions) between sessions 1 and 2, in men (black points and continuous lines) and women (empty circles and dashed lines).

protocol in the present study is that the crank length was the same for all participants. The usual crank length is probably higher than the optimal length in small participants, which could partially explain the lower reliability in women. Therefore, familiarization sessions are required in small participants.

6. Conclusion

The present study showed high reliability of P_{max} and F_0, allowing the use of these parameters in longitudinal evaluations. Furthermore, the reliability of P_{max} was better than that of F_0 whatever the expression of the results (expressed

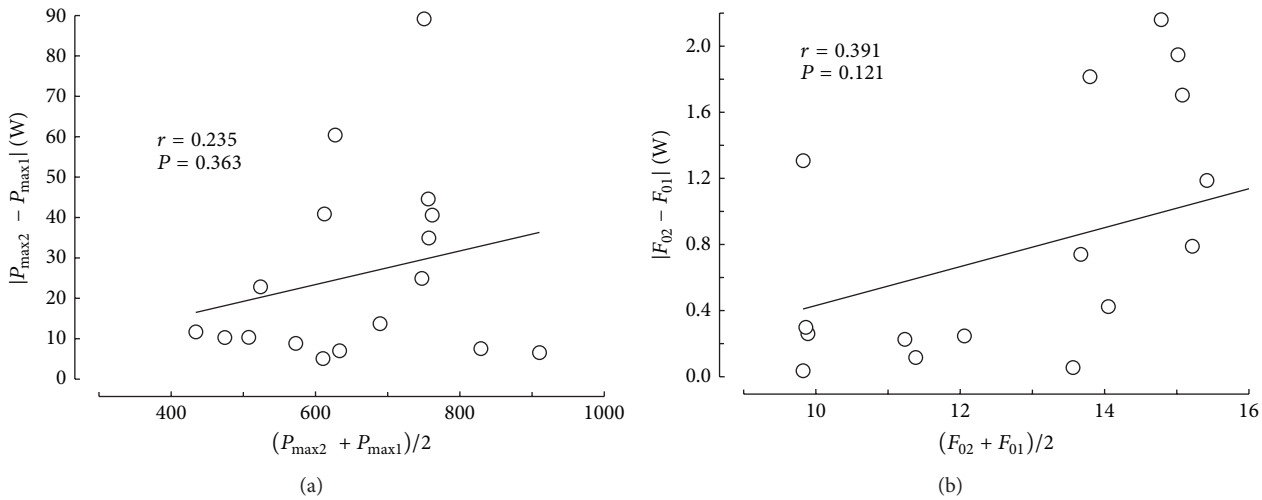

Figure 5: Plot of the absolute differences between the results of sessions 1 and 2 (ordinates) and the individual means (abscissae) for P_{max} (a) and F_0 (b) in women.

in absolute unit or data related to body dimension). The reliability indices were also better in men and cycling force-velocity tests than in women and cranking force-velocity tests. Further studies are needed to judge the reliability of V_0.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Laterality of Stance during Optic Flow Stimulation in Male and Female Young Adults

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During self-motion, the spatial and temporal properties of the optic flow input directly influence the body sway. Men and women have anatomical and biomechanical differences that influence the postural control during visual stimulation. Given that recent findings suggest a peculiar role of each leg in the postural control of the two genders, we investigated whether the body sway during optic flow perturbances is lateralized and whether anteroposterior and mediolateral components of specific center of pressure (COP) parameters of the right and left legs differ, reexamining a previous experiment (Raffi et al. (2014)) performed with two, side-by-side, force plates. Experiments were performed on 24 right-handed and right-footed young subjects. We analyzed five measures related to the COP of each foot and global data: anteroposterior and mediolateral range of oscillation, anteroposterior and mediolateral COP velocity, and sway area. Results showed that men consistently had larger COP parameters than women. The values of the COP parameters were correlated between the two feet only in the mediolateral axis of women. These findings suggest that optic flow stimulation causes asymmetry in postural balance and different lateralization of postural controls in men and women.

1. Introduction

The human upright stance is characterized by continuous movements of the body similar to an inverted pendulum [1]. Vision provides the nervous system with information regarding the position and movements of elements present in the environment relatively to the body, playing an important role in the postural stabilization. The characteristic pattern of visual stimuli that provides information of self-motion and the environmental structure is defined as "optic flow" [2, 3]. The optic flow originates from the focus of expansion (FOE), a point of the visual scene that corresponds to the final destination of self-motion. The neural mechanisms integrate visual, vestibular, and proprioceptive inputs of self-motion perception to generate the typical body oscillation defined as body sway. The body sway is regulated by the neuromotor system and is considered a consequence of small postural oscillations. These small postural oscillations reflect the regulatory activity of the several control loops of stabilization of an unstable structure, such as the human body, for maintenance of balance [4, 5].

Until now, several studies have focused on the maintenance of balance control, looking at the variation of the center of pressure (COP) trajectory. The COP analysis with bilateral force plate can be useful for assessing postural behavior related to each foot in healthy individuals [6, 7]. Few studies have addressed the laterality or asymmetry during quiet stance; however, these studies were performed with the eyes open or closed or under two-dimensional visual stimulation [8–10].

In a previous paper, we showed that foveal, peripheral, and full field optic flow stimulations evoke different muscular activations in the right and left leg and different directions of oscillation in men and women [11]. Thus, the aim of this paper was to verify whether the different oscillations caused by foveal, peripheral, and full field optic flows depend on the variations of specific COP parameters in each leg. Results showed that optic flow significantly affected the COP

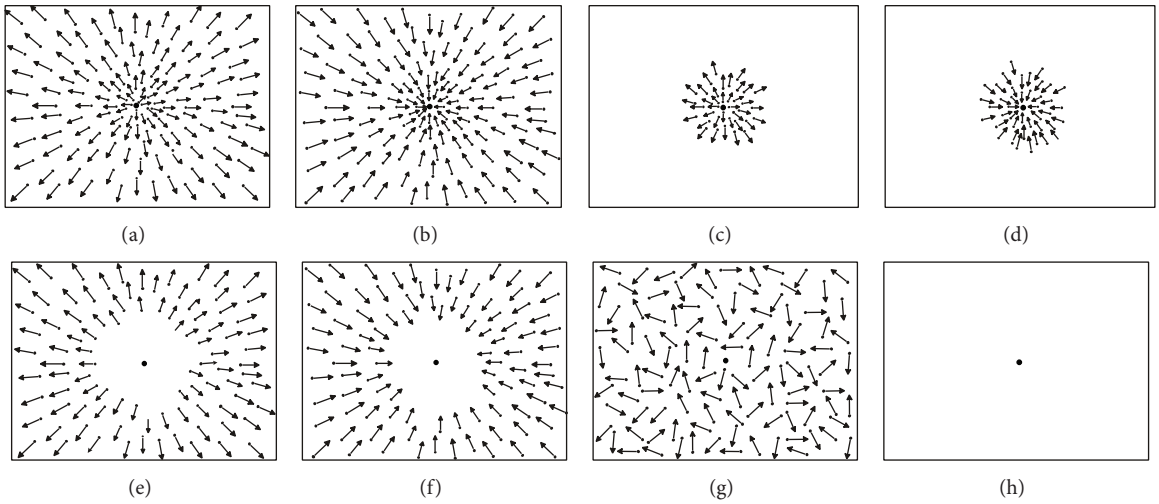

Figure 1: Optic flow stimuli. Arrows represent the velocity vectors of moving dots. (a) Full field expansion. (b) Full field contraction. (c) Foveal expansion. (d) Foveal contraction. For the foveal stimuli, the stimulated area had a radius of 7°. (e) Peripheral expansion. (f) Peripheral contraction. For the peripheral stimuli, the blank area in the center had a radius of 20°. (g) Random motion stimulus. (h) Baseline (fixation in the dark).

parameters, and each foot had a specific contribution to postural control that was not evident in the global data.

2. Methods

For this study, we reexamined the data of the experiments performed on 24 right-handed and right-footed subjects [11], 12 women and 12 men, ranging from 20 to 30 years (average age was 24.5). Average height and weight plus standard deviation (SD) was 167 ± 5 cm and 62 ± 5 kg for women and 178 ± 6 cm and 72 ± 5 kg for men. All subjects voluntarily participated in the experiments. The experimental protocol was approved by the Institutional Ethic Committee of the University of Bologna. Recordings were performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki. The subjects included in the sample practiced a moderate physical activity (i.e., no more than 1 hour three times a week). None of the subjects had any history of gait or posture disorders or injuries in the previous two years; all of them had normal or corrected-to-normal vision.

2.1. Experimental Procedure and Stimuli. The experimental paradigm and visual stimuli are identical to those described in a previous publication [11]. The experiments were performed in a dark room. Optic flow stimuli and fixation point were presented by a retro video projector (Sony VPL EX3) positioned 415 cm away from a translucent screen that covered 135 × 107° of visual field, placed 115 cm from the subjects' eyes.

Optic flow stimuli consisted of white dots (1.3 cd/m^2) of 0.4° size, which moved at a speed of 5°/s. The stimuli were expanding and contracting flows originating from a central FOE. The fixation point consisted of a white dot of 0.6° always positioned in the middle of the screen. The focus of expansion was always in the center of the screen. Expansion and contraction optic flows were presented as full field (Exp and Contr, resp., Figures 1(a) and 1(b)), foveal (ExpF and ContrF, resp., Figures 1(c) and 1(d)), or peripheral (ExpP and ContrP, resp., Figures 1(e) and 1(f)) stimuli. A random dot motion stimulus was used as a control (random, Figure 1(g)). For baseline trials, stabilometric activity was recorded while the subject fixated on the fixation point without visual stimulation (Figure 1(h)). Stimuli were created using MATLAB Psychophysical Toolbox (The MathWorks Inc.) and had the same dot density with respect to the retinal stimulated area. Full field stimulus is 1155 dots. Foveal stimulus is 36 dots. Peripheral stimulus is 992 dots.

The stabilometric data were acquired using two Kistler force platforms (number 9286BA). Subjects were instructed to place a foot on each platform before the beginning of each trial. The platforms were marked to normalize posture and to control the subject's distance from the screen. Subjects had to look at a white fixation point (0.6°), which was also the FOE of the optic flow stimuli, always positioned in the center of the screen and adjusted to the height of each subject. The stimulus was present during the entire trial duration and trial onset was determined by the stimulus onset.

2.2. Data Analysis. We acquired 5 trials for each stimulus condition and 4 trials at baseline (i.e., fixation in the dark without visual stimulation, Figure 1(h)). Each stimulus lasted about 30–35 s. Stabilometric signals were recorded at 1000 Hz, then low-pass filtered at 15 Hz, and resampled at 250 Hz. We recorded ground reaction forces and COP measures from each foot by the two platforms. We analyzed both anteroposterior COP and mediolateral COP of each foot using SMART Analyzer (BTS Bioengineering Inc.) and MATLAB (The MathWorks, Inc.). Subsequently, we computed the global COP according to the following formula [12]:

$$\mathrm{COP}_{\mathrm{global}} = \frac{\mathrm{COP}_{\mathrm{L}} * R_{\mathrm{VL}}}{(R_{\mathrm{VL}} + R_{\mathrm{VR}})} + \frac{\mathrm{COP}_{\mathrm{R}} * R_{\mathrm{VR}}}{(R_{\mathrm{VL}} + R_{\mathrm{VR}})}, \quad (1)$$

where R_{VL} and R_{VR} are the vertical reaction forces from left and right feet, respectively. The analysis was performed in the first 25 s of each trial.

In this study, we computed five measures referring to the COP of each foot and COP_{global}: (1) the anteroposterior range of oscillation (APO), which is the difference between the maximum and minimum range of oscillation in anteroposterior direction [13], (2) the mediolateral range of oscillation (MLO), which is the difference between the maximum and minimum range of oscillation in the mediolateral direction [13], (3) the anteroposterior COP velocity (VelAP), (4) the mediolateral COP velocity (VelML), the two latter measurements reflecting the total distance travelled by the COP over time on each axis [13–16], and (5) the COP area (Area), quantified within the 95% confidence ellipse, which is the enclosed area covered by the COP as it oscillates within the base of support [17].

We first computed the percentage of loading in the right and left foot using Smart-Analyzer software (BTS Bioengineering Inc.) and MATLAB (The MathWorks Inc.). The values of the percentage of loading were then analyzed with a multivariate ANOVA (within-subject factor: stimuli; between-subject factors: side and gender).

Then, we analyzed the COP parameters APO, MLO, VelAP, VelML, and Area using Sway and Smart-Analyzer software (BTS Bioengineering Inc.) and MATLAB (The MathWorks Inc.). The analysis was performed separately for measurements of each limb and global. To analyze the influence of optic flow stimuli on postural control, we performed a repeated-measure ANOVA in which optic flow stimuli and side (right, left, and global) were the within-subject factors, while gender was the between-subjects factor.

After having assessed the effects of stimuli, side, and gender, we then analyzed in depth the relationship between the left and right feet in response to visual stimuli using a bivariate Pearson linear correlation analysis.

Lastly, we looked at the degree of variation of the right and left foot in the five COP parameters using the coefficient of variation (CV) computed as the ratio of the standard deviation to the mean. The CV was computed for each trial of each stimulus in each subject. Then, values for all subjects in each condition and group were averaged.

3. Results

3.1. Limb Loading. To quantify the asymmetry, we first computed the limb loading. Mean values of the percentage of loading are shown in Figure 2. Women had an almost equal load, while men consistently loaded the left leg more than the right. The results of the multivariate ANOVA (see Methods) showed an effect of side in all stimuli ($F(8, 35) = 10,57$, $p < 0.001$) and an interaction effect of side × gender in all stimuli ($F(8, 35) = 7,74$, $p < 0.001$). No main effect of gender was found ($F(8, 35) = 0.31$, $p = 0.95$).

3.2. Effect of Stimuli, Side, and Gender on Postural Responses. All COP parameters showed significant main effects of stimuli, side, and gender as summarized in Table 1. VelML did not show a significant main effect of gender but showed significant interaction effects (stimulus × gender and stimulus × gender × side). Area showed an interaction effect between stimulus and side.

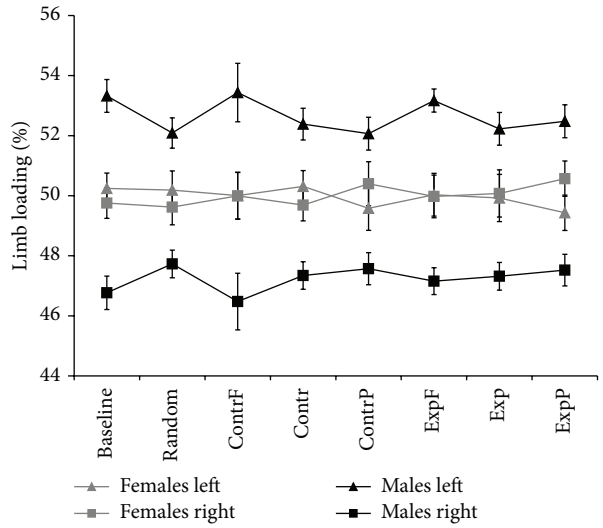

Figure 2: Average values of left and right percentage of loading in the right and left foot of men and women. Data are shown for all stimuli and baseline. Each data point shows mean ± standard error (SE). ContrF: foveal contraction, Contr: full field contraction, ContrP: peripheral contraction, ExpF: foveal expansion, Exp: full field expansion, and ExpP: peripheral expansion.

The results of the between-subjects analysis (ANOVA, see Methods) showed that, among the COP parameters, MLO showed more differences between men and women. The gender effect was examined in each stimulus of the right and left leg allowing the analysis in 16 conditions. A significant effect was found in almost all stimuli (14/16). The two nonsignificant effects were found in expansion ($p = 0.14$) and foveal contraction ($p = 0.08$) of the left foot. No difference emerged in the stimuli of the MLO_{global}. In APO, however, significant gender effects were found for foveal ($p < 0.024$) and peripheral contraction ($p < 0.028$) stimuli in the left leg, while no differences were found in the APO_{global}. The VelAP and VelML showed similar results: in VelAP, a significant gender effect was found in the left foot only for baseline, random, and foveal stimuli ($p < 0.05$), while in VelML significant gender differences were observed in the left foot for baseline, random, and peripheral contraction stimuli ($p < 0.05$). Similar to MLO, the Area parameter showed a significant gender effect in the right and left foot in 13 out of 16 stimuli (ANOVA, $p < 0.05$). The three nonsignificant effects were found for expansion ($p = 0.09$) and peripheral contraction ($p = 0.08$) of the left foot and foveal contraction of the right foot ($p = 0.22$). No differences were found for the $Area_{global}$ parameter. Figure 3 shows the mean values of the COP parameters in both feet and the global data for men and women. All parameters yielded larger values in men. The left foot had larger values of APO and Area (Figures 3(a) and 3(e)), while the right foot showed higher values in MLO, VelAP, and VelML (Figures 3(b)–3(d)).

Table 1: Full statistical information for the repeated-measure ANOVA in which optic flow stimuli and side (right, left, and global) were the within-subject factors, while gender was the between-subjects factor.

	APO	MLO	VelAP	VelML	Area
Side	$F(2; 42) = 7.70$; MSE = 148.28; **p = 0.009***	$F(2; 42) = 15.05$; MSE = 56.24; **p = 0.002***	$F(2; 42) = 5.38$; MSE = 3.72; **p = 0.015***	$F(2; 42) = 5.08$; MSE = 3.35; **p = 0.05***	$F(2; 42) = 123.77$; MSE = 883.66; **p < 0.001***
Sex	$F(1; 21) = 5.96$; MSE = 489.87; **p = 0.024***	$F(1; 21) = 14.63$; MSE = 151.1; **p = 0.009***	$F(1; 21) = 5.68$; MSE = 27.13; **p = 0.041***	$F(1; 21) = 6.94$; MSE = 4.86; $p = 0.07$	$F(1; 21) = 4.22$; MSE = 1165.7; **p = 0.05***
Stimulus	$F(7; 147) = 11.73$; MSE = 76.48; **p < 0.001***	$F(7; 147) = 5.43$; MSE = 103.56; **p = 0.035***	$F(7; 147) = 6.49$; MSE = 2.53; **p = 0.002***	$F(7; 147) = 5.72$; MSE = 0.22; **p = 0.001***	$F(7; 147) = 5.30$; MSE = 193.62; **p = 0.001***
Stimulus × sex	$F(7; 147) = 0.55$; MSE = 76.48; $p = 0.706$	$F(7; 147) = 1.54$; MSE = 103.56; $p = 0.259$	$F(7; 147) = 1.34$; MSE = 2.53; $p = 0.281$	$F(7; 147) = 2.71$; MSE = 0.22; **p = 0.036***	$F(7; 147) = 0.25$; MSE = 193.62; $p = 0.891$
Side × sex	$F(2; 42) = 0.45$; MSE = 148.28; $p = 0.524$	$F(2; 42) = 0.31$; MSE = 56.24; $p = 0.67$	$F(2; 42) = 1.16$; MSE = 7.2; $p = 0.335$	$F(2; 42) = 0.38$; MSE = 4.75; $p = 0.635$	$F(2; 42) = 0.8$; MSE = 709.51; $p = 0.393$
Stimulus × side	$F(14; 294) = 1.57$; MSE = 41.95; $p = 0.171$	$F(14; 294) = 1.92$; MSE = 130.85; $p = 0.2$	$F(14; 294) = 0.98$; MSE = 0.7; $p = 0.417$	$F(14; 294) = 1.67$; MSE = 1.97; $p = 0.26$	$F(14; 294) = 2.08$; MSE = 72.47; **p = 0.012***
Side × stimulus × sex	$F(14; 294) = 0.5$; MSE = 41.95; $p = 0.781$	$F(14; 294) = 0.55$; MSE = 130.85; $p = 0.556$	$F(14; 294) = 1.31$; MSE = 0.7; $p = 0.289$	$F(14; 294) = 2.17$; MSE = 0.3; **p = 0.027***	$F(14; 294) = 0.13$; MSE = 229.03; $p = 0.977$

Significant values are in bold and marked with an asterisk.

3.3. Correlation Analysis. A bivariate Pearson correlation was used to test whether the relationship between the right and left foot in each COP parameter was linear. The analysis was performed separately for men and women on left versus right foot for all stimuli and baseline values of each COP parameter. In women (Figure 4(a)), significant linear correlations between the two feet were found only in MLO (baseline: $R(9) = 0.659$, $p = 0.05$; random: $R(11) = 0.737$, $p = 0.01$; foveal contraction: $R(11) = 0.67$, $p = 0.02$; contraction: $R(11) = 0.634$, $p = 0.036$; peripheral contraction: $R(11) = 0.731$, $p = 0.011$; peripheral expansion: $R(12) = 0.778$, $p = 0.003$). The values of the right and left foot in the COP other parameters showed very low correlation coefficients, often negative (Figure 4(a)). Men, however, showed few significant correlations between right and left foot COP values (Figure 4(b)) but the two feet seem to have more similar movements than those of women (APO random: $R(11) = 0.603$, $p = 0.049$; APO peripheral expansion: $R(11) = 0.644$, $p = 0.032$; VelAP foveal contraction: $R(11) = 0.733$, $p = 0.01$; VelAP contraction: $R(11) = 0.641$, $p = 0.033$; VelAP foveal expansion: $R(12) = 0.877$, $p < 0.001$; MLO foveal contraction: $R(10) = 0.631$, $p = 0.05$; MLO contraction: $R(9) = 0.72$, $p = 0.029$; Area contraction: $R(11) = 0.688$, $p = 0.019$; Area foveal expansion: $R(11) = 0.736$, $p = 0.01$).

3.4. Variation in the COP Parameters. To examine the variability of postural adjustments during optic flow stimulation, we computed the CV for the five COP parameters in the right and left foot. MLO consistently showed greater variability than APO. Baseline stimuli always had the highest CV, indicating that the absence of visual stimulation caused a greater instability. In women, different variability was observed in the left and right foot: MLO_{left} always showed higher CV than MLO_{right}, while, in almost all stimuli, APO_{right} showed higher CV than APO_{left} (Figure 5(a)). In men, MLO had still higher variations than APO; however, they were smaller when compared to those of women (Figure 5(b)). The greatest variations were observed in the COP velocity (Figures 5(c) and 5(d)). In both men and women, VelML always showed greater variations than VelAP, suggesting that subjects consistently experienced a loss of balance control on the mediolateral axis. Both genders showed greater variability for Area of the left foot for the majority of stimuli (Figures 5(e) and 5(f)). Men showed the greatest variability of the COP Area.

As these observations on the CV were largely descriptive, the CV values were further analyzed to quantify the variability related to gender and foot. A one-way ANOVA, with side as between-subject factor and stimuli as within-subject factor, was performed separately for men and women. Significant differences between the left and right feet were found only in women in VelML for all visual stimuli (foveal contraction: $F(1, 23) = 4.69$, MS = 630.29, $p = 0.041$; contraction: $F(1, 23) = 20.73$, MS = 166.48, $p < 0.001$; peripheral contraction: $F(1, 21) = 15.23$, MS = 144.67, $p = 0.001$; foveal expansion: $F(1, 21) = 24.05$, MS = 187.27, $p < 0.001$; expansion: $F(1,23) = 13.61$, MS = 125.18, $p = 0.001$; peripheral expansion: $F(1, 23) = 12.7$, MS = 125.26, $p = 0.002$; random: $F(1, 22) = 5.04$, MS = 696.75, $p = 0.036$;

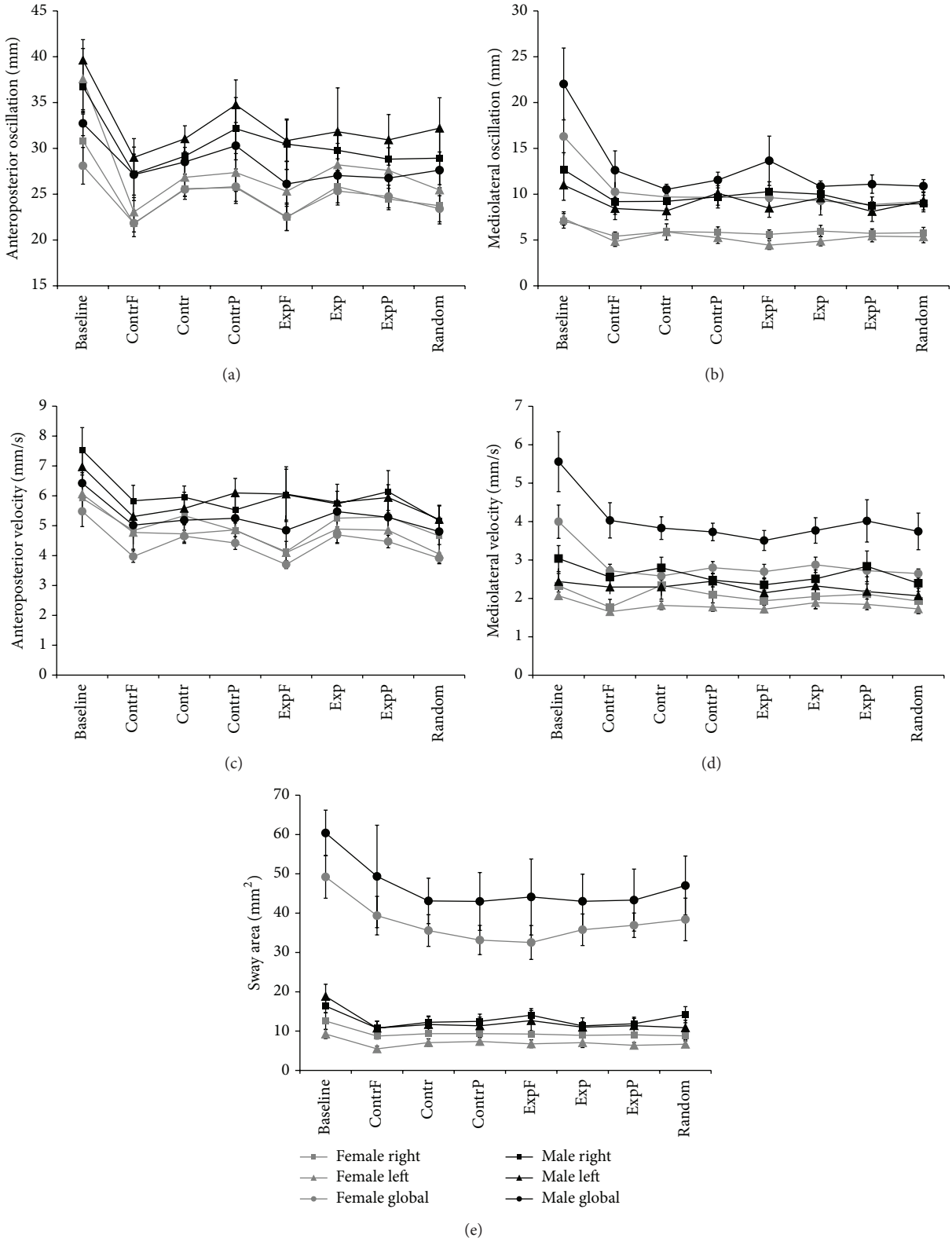

Figure 3: Average values of COP parameters in the left and right limb and global data. Values are shown for men and women during optic flow stimuli and baseline. (a) Anteroposterior range of oscillation (APO). (b) Mediolateral range of oscillation (MLO). (c) Anteroposterior velocity (VelAP). (d) Mediolateral velocity (VelML). (e) Sway area (Area). Each data point shows mean ± standard error (SE). Conventions are as in Figure 2.

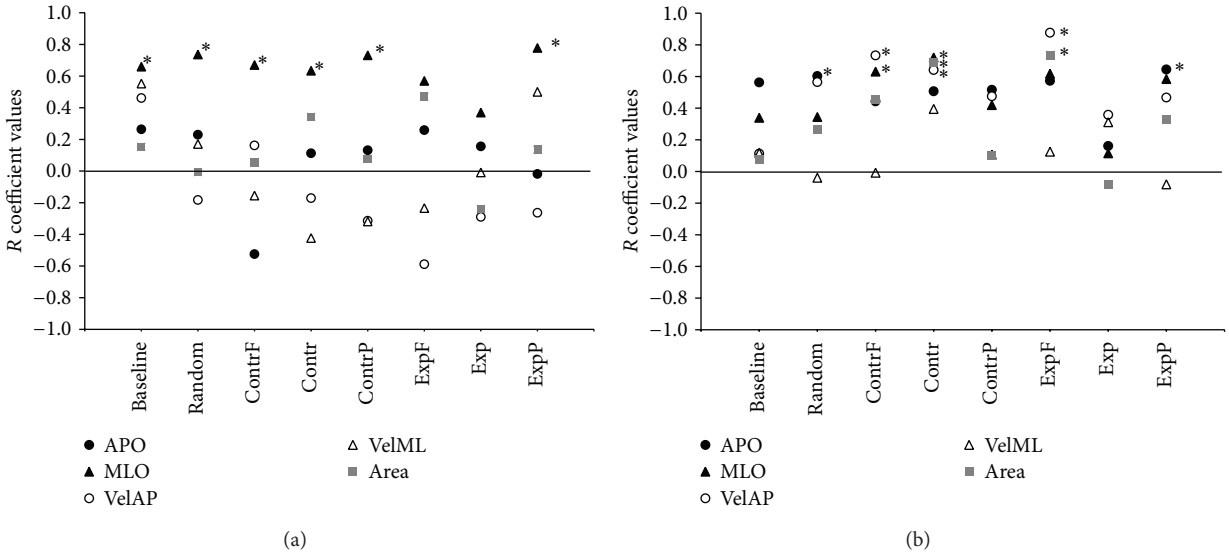

Figure 4: Correlation coefficients for the correlation analysis between the right and left foot. (a) Women. (b) Men. Asterisks indicate significant values (bivariate Pearson correlation, $p < 0.05$). Conventions are as in Figures 2 and 3.

baseline: $F(1, 23) = 3.84$, MS = 55.72, $p = 0.063$). No significant differences between the two feet were found in the other parameters in men.

4. Discussion

The optic flow is a key input for maintaining postural stability during self-motion [18]. The human body is fundamentally asymmetrical, manifesting in the functional anteroposterior and mediolateral asymmetries observed in balance control [19]. The aim of this study was to investigate whether the body sway during foveal, peripheral, or full field optic flow stimulation is lateralized and whether anteroposterior and mediolateral components of specific COP parameters of the right and left foot differ between the two genders.

4.1. Limb Load Asymmetry. An important issue in studying postural asymmetry is limb loading. Some evidence seems to support the idea that healthy subjects unequally distribute their weight across the two feet in conditions of eyes open and closed [9, 20]. Our female subjects showed an almost even limb loading while men loaded the left limb more than the right. This is a first indication of gender differences in the postural control during optic flow stimulation. The left loading preference of men, irrespective of handedness and footedness, can be explained by the different muscular activity. Indeed, the electromyography recordings performed in the previous study showed that men showed the greatest activation in the left thigh muscles [11]. This is also supported by the fact that male soccer players have a better standing balance on the nondominant leg, probably as a consequence of many hours of soccer practice during which they maintain standing balance for a few seconds on the nondominant leg for kicking the ball with the dominant foot to have more precision [21]. Practicing physical activity seems to enhance the inter-leg differences, because it has been shown that dominance does not interfere in the evaluation of single-foot balance among healthy sedentary individuals [22].

4.2. Contribution of Individual Leg on Postural Control. Footedness entails postural asymmetry [23]. All subjects were right-footed [11]. This, together with our analysis model, allowed us to broaden the knowledge on the contribution of each leg to postural control during the view of optic flow stimuli. Some authors suggest differential effects on the recurrent dynamics of the individual leg COPs and COP_{global} trajectories [24, 25]. The detailed analysis of left, right, and global data shows that each leg contributes individually to side-by-side postural control, which is not obvious when analyzing the global data. As pointed out by King and coworkers [24, 25], the degree of asymmetry between left leg and right leg COP dynamics differed across all postural stances and COP_{global} dynamics. Analyzing each foot separately revealed variation of postural control in terms of different variability between the left and right foot parameters. The present study emphasizes asymmetries between the two feet in the postural maintenance showing different dynamics between the two feet in each parameter.

4.3. Gender Asymmetry. The present results suggest that optic flow stimuli produced different COP oscillations, velocities, and area dimensions. These results point out important characteristics of the feet asymmetry; the fact that the two feet exhibit different values in distinct parameters may indicate that each foot has its own role in balance control. As suggested by Anker and coworkers [14], the muscles of the unloaded leg lose their capacity to generate effective stabilizing ankle torques, while the velocity of COP under the loaded leg increases, reflecting the generation of compensatory ankle moments. Our subjects showed no significant relationship between limb dominance and the side of load preference

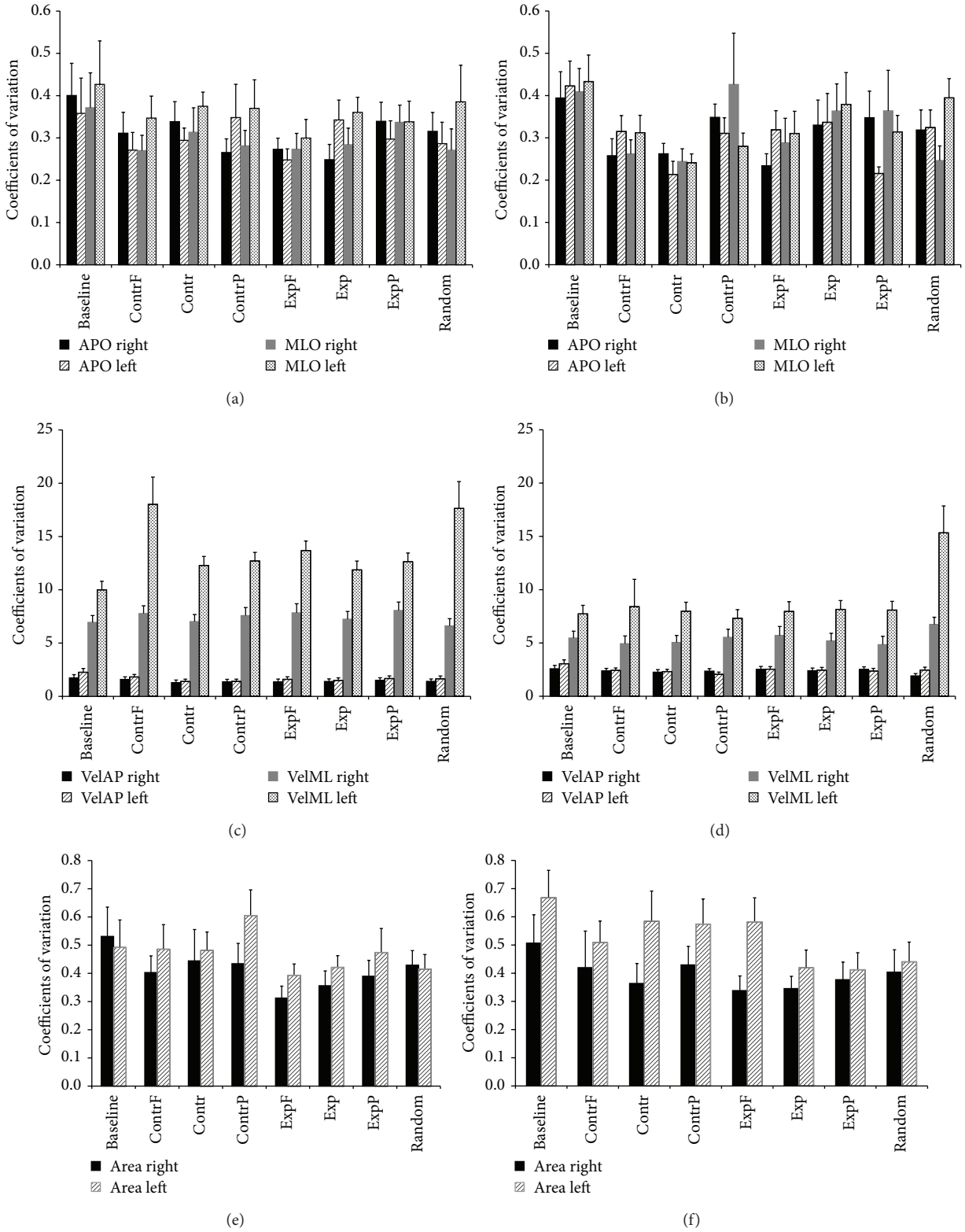

Figure 5: Coefficients of variations of COP parameters across the right and left feet in men and women. (a) Women anteroposterior range of oscillation (APO) and mediolateral range of oscillation (MLO). (b) Male APO and MLO. (c) Women anteroposterior velocity (VelAP) and mediolateral velocity (VelML). (d) Men VelAP and VelML. (e) Female sway Area. (f) Male sway Area.

meaning a continuous load/unload balance between the two feet. The lack of correlations between the behavior of the two feet in the majority of the stimuli and parameters is another indication of gender and limb differences, further emphasized by the different variability observed in the mediolateral velocity of left and right foot in women. These findings seem to suggest that foot asymmetry induces inter-leg coordination dynamics based on postural demands during optic flow stimulation and during increasing difficulty to maintain correct body balance. This might reveal the use of multiple timescale processes within each leg to produce a stable and flexible postural strategy.

Gender differences in brain asymmetry are well documented and may explain the different postural strategies exhibited by men and women. The brain of adult women is, from the functional point of view, less asymmetrical than that of men [26, 27]. A recent study showed a larger left > right asymmetry in women in anterior brain regions and a larger right > left asymmetry in men orbitofrontal, inferior parietal, and inferior occipital cortices [28]. The brain asymmetry is also evident in motor function, as it is known that the gray matter density in the corticospinal tract shows a hemispheric asymmetry related to hand preference, and the maturation of the corticospinal tract during adolescence differs between men and women due to the influence of testosterone [29]. It seems that the leftward asymmetry of the corticospinal tract may reflect an early established asymmetry in the corticomotoneuronal fibres. The present results, together with those of previous findings [30], suggest that men and women adapt differently to cortical and corticospinal asymmetry leading to different behaviors of the right and left limb.

5. Conclusions

This study provides new evidence on the postural strategy used by men and women in the control of stance under visual optic flow stimulation. The feet asymmetry observed during optic flow stimulation causes specific inter-leg coordination dynamics necessary to maintain the control of posture. This might suggest that the postural control system uses various mechanisms within each leg to produce the most appropriate postural response to interact with the extrapersonal environment.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Relation between the Sensory and Anthropometric Variables in the Quiet Standing Postural Control: Is the Inverted Pendulum Important for the Static Balance Control?

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The aim of this study was to evaluate the relation between the sensory and anthropometric variables in the quiet standing. *Methods*. One hundred individuals (50 men, 50 women; 20–40 years old) participated in this study. For all participants, the body composition (fat tissue, lean mass, bone mineral content, and bone mineral density) and body mass, height, trunk-head length, lower limb length, and upper limb length were measured. The center of pressure was measured during the quiet standing posture, the eyes opened and closed with a force platform. Correlation and regression analysis were run to analyze the relation among body composition, anthropometric data, and postural sway. *Results*. The correlation analysis showed low relation between postural sway and anthropometric variables. The multiple linear regression analyses showed that the height explained 12% of the mediolateral displacement and 11% of the center of pressure area. The length of the trunk head explained 6% of displacement in the anteroposterior postural sway. During eyes closed condition, the support basis and height explained 18% of mediolateral postural sway. *Conclusion*. The postural control depends on body composition and dimension. This relation is mediated by the sensory information. The height was the anthropometric variable that most influenced the postural sway.

1. Introduction

During the quiet upright position, both internal and external factors affect the postural sway. To deal with any type of physical or physiological constraint or perturbation, sensory information is necessary for the postural control. The sensory information about posture and kinesthesia feeds the postural control to build a postural frame of reference. For the sagittal plane, the quiet upright posture can be modeled as an inverted pendulum [1] which sways mostly around the ankle joint. This condition reveals the ankle strategy [2]. On the other hand, for the frontal plane, the same orthostatic posture can be modeled as a double inverted pendulum, revealing the hip strategy [2]. The postural control uses the combination of both strategies to keep the postural sway inside the basis of support. The postural sway is the response to control the position of the center of mass [1]. Those postural strategies suggest that close loop control is important to control balance.

To control the inverted pendulum during standing, afferent sensory information is used to tune the gain of postural responses [3] and to deal with the body dimensions constraints. Then, sensory information and anthropometric variables are important factors for the postural control [3]. The variables of the inverted pendulum model that affects the postural control are its length, mass, and joint stiffness [1, 4]. Body mass changes the postural stability in prepubescent children and adolescents [5, 6], young adults [7–11], and old adults [12, 13].

The changes in sensory information about the postural frame of reference are potentially dangerous for postural control. The postural sway parameters increase during standing

Table 1: Characteristics of the study population (anthropometric and posturographic).

Variables	General group mean (SD) $N = 100$
Anthropometrics	
Height (cm)	168.8 (9.5)
Body mass (kg)	69.9 (14.3)
BMI (kg/m^2)	24.3 (3.6)
Trunk-cephalic length (cm)	89.9 (4.4)
% fat	30.2 (10.1)
Lean mass (kg)	46.9 (11.8)
Bone mineral composition (kg)	2.77 (0.55)
Waist-hip ratio (cm)	81.7 (7.6)
Posturographic measurements ($\log_{10}$)	
Eyes open	
ML COP RMS (cm)	−0.685 (0.154)
AP COP RMS (cm)	−0.421 (0.128)
Sway velocity (cm/s)	−0.130 (0.097)
Displacement area (cm^2)	0.140 (0.243)
Eyes closed	
ML COP RMS (cm)	−0.612 (0.161)
AP COP RMS (cm)	−0.332 (0.148)
Sway velocity (cm/s)	0.008 (0.110)
COP area (cm^2)	0.306 (0.259)

ML COP RMS: medial lateral center of pressure root mean square; AP COP RMS: anteroposterior center of pressure root mean square; BMI: body mass index; SD: standard deviation.

when the eyes are closed [14]. Chiari et al. [14] have shown that biomechanical factors (body size and body composition) are strongly related to postural sway under eyes closed condition. Therefore, the inverted pendulum model can explain what happens during standing. How important are the mechanical properties of the body for the postural control when no visual information is available? Is the inverted pendulum model able to explain how the postural control regulates the body sway?

In order to answer those questions, the aim of this study was to evaluate the relation between the sensory and anthropometric variables in the quiet standing postural control. The hypothesis of this study is that sensory information constrains the importance of the inverted pendulum variables for the postural control.

2. Method

2.1. Participants. Fifty young men and fifty young women were the participants. Their characteristics are described in Table 1. They gave their written informed consent to participate in this study and this protocol was approved by the Local Ethical Committee (number 1256/06). The inclusion criteria were as follows: no history of injury or lower limbs and trunk surgery; to be irregularly physically active over the last six months according to the International Physical Activity Questionnaire; no disease or functional impairment of the sensory system; and no current use of medications that might alter postural control. The exclusion criteria were inability to carry out the balance tests.

2.2. Procedures. The anthropometric and body composition variables were measured. Those measurements were always performed by the same person, who was trained previously to make such measurements. The anthropometric measurements were made according to the ISAK (International Society for the Advancement of Kinanthropometry) standards [15]. The participants' height, weight, body mass index (BMI), trunk-encephalic length, and the waist-hip ratio were recorded. The bone densitometry (LUNAR-DPX, Madison, USA), using dual energy X-ray absorptiometry (DEXA), was used to measure body composition: the percentage of fat, bone, and body lean mass.

A portable force platform (AccuSway Plus, AMTI, MA, USA) was applied to measure the ground reaction forces and moments of force during the quiet standing posture task. The sampling frequency of the forces (F) and moments of forces (M) was 100 Hz. The center of pressure was calculated according to the following equations:

$$\mathrm{COP}_{\mathrm{AP}} = \frac{M_{\mathrm{AP}}}{F_{\mathrm{v}}}$$
$$\mathrm{COP}_{\mathrm{ML}} = \frac{M_{\mathrm{ML}}}{F_{\mathrm{v}}}, \quad (1)$$

where the indexes of center of pressure (COP), M, and F indicate the anteroposterior (AP), mediolateral (ML), and vertical (v) directions. The raw COP signal was filtered (10 Hz low-pass 4th order Butterworth filter) and demeaned to eliminate the foot position bias.

The participants should stand as quiet as possible on the force platform with their arms alongside the body with their eyes opened or closed for 60 seconds. Every participant repeated the task three times for each vision condition. Their feet position was marked with a template drawn on the force plate surface.

The COP variables are the root mean square (RMS) of the COP for the AP and ML directions, mean velocity, and its sway area (COP area).

2.3. Statistical Analysis. To check the data distribution, the Kolmogorov-Smirnov test was run. The logarithmic transformation was applied when any variable was not normally distributed. Two analyses were applied on the postural sway (dependent) and body composition and anthropometrical (independent) variables: the analysis of correlation and the multiple linear regression model (MLRM) analysis.

The MLRM analysis was applied when the variables presented $P \leq 0.20$ in the correlation analysis. Those variables were ranked from lowest to highest P value. Then, the MLRM using stepwise forward selection was run and the variables were added to model one by one, according to their ranking. The variables for which $P \leq 0.05$ were kept in the model.

Table 2: Correlation between balance and the anthropometric variables in the group, with eyes open.

Variables	ML COP RMS (cm) r (P)	AP COP RMS (cm) r (P)	Sway velocity (cm/s) r (P)	COP area r (P)
Height (cm)	0.40 (0.004)*	0.28 (0.005)*	0.33 (0.001)*	0.35 ($P \leq 0.001$)*
Mass (kg)	0.23 (0.09)	0.24 (0.01)*	0.15 (0.11)	0.23 (0.02)*
BMI (kg/m^2)	0.03 (0.79)	0.12 (21)	−0.05 (0.57)	0.06 (0.54)
Trunk-cephalic length (cm)	0.18 (0.20)	0.27 (0.006)*	0.24 (0.01)*	0.24 (0.01)*
% fat	0.03 (0.82)	−0.03 (0.73)	−0.26 (0.009)*	−0.14 (0.15)
Lean mass (g)	0.37 (0.007)*	0.21 (0.03)*	0.28 (0.004)*	0.27 (0.006)*
Bone mineral composition (g)	0.29 (0.03)*	0.22 (0.02)*	0.19 (0.05)*	0.24 (0.01)*
Waist-hip ratio (cm)	0.12 (0.39)	0.07 (0.43)	0.18 (0.06)	0.17 (0.08)

Pearson's coefficient (r); *$P \leq 0.05$, ML COP RMS: medial lateral center of pressure root mean square; AP COP RMS: anteroposterior center of pressure root mean square; BMI: body mass index.

Table 3: Correlation between balance and the anthropometric variables in the whole group, with eyes closed.

Variables	ML COP RMS (cm) r (P)	AP COP RMS (cm) r (P)	Sway velocity (cm/s) r (P)	COP area r (P)
Height (cm)	0.35 ($P \leq 0.001$)*	0.05 (0.56)	0.31 (0.001)*	0.25 (0.01)*
Mass (kg)	0.22 (0.02)*	0.08 (0.42)	0.17 (0.08)	0.18 (0.07)
BMI (kg/m^2)	0.04 (0.69)	0.06 (0.53)	−0.02 (0.84)	0.06 (0.50)
Trunk-cephalic length (cm)	0.21 (0.03)*	0.05 (0.60)	0.16 (0.09)	0.16 (0.10)
% fat	−0.12 (0.22)	−0.002 (0.98)	−0.18 (0.06)	−0.05 (0.59)
Lean mass (g)	0.24 (0.01)*	0.06 (0.54)	0.25 (0.01)*	0.17 (0.08)
Bone mineral composition (g)	0.22 (0.02)*	−0.002 (0.98)	0.16 (0.09)	0.13 (0.17)
Waist-hip ratio (cm)	0.25 (0.01)*	0.02 (0.81)	0.18 (0.06)	0.16 (0.09)

Pearson's coefficient (r); *$P \leq 0.05$, ML COP RMS: medial lateral center of pressure root mean square; AP COP RMS: anteroposterior center of pressure root mean square; BMI: body mass index.

3. Results

The anthropometric and posturographic data of the participants are described in Table 1.

3.1. Correlation Analysis. The correlation coefficients of the postural sway and the anthropometric and body composition variables according to visual conditions are presented in Tables 2 and 3. For the opened eyes condition, the height, waist-hip ration, trunk-cephalic length, and bone mineral composition were correlated to AP and ML COP RMS and COP area. For the closed eyes condition, the height was correlated to AP and ML COP RMS and COP area.

3.2. Regression Analysis. The MLRM analysis with the postural sway and the anthropometric and body composition variables for the visual conditions is described in Table 4. For the opened eyes condition, the height explained 12% of the ML COP RMS, 10% of the sway velocity, and 11% of the COP area; and the trunk-cephalic length explained 6% of the AP COP RMS. For the closed eyes condition, the height explained 18% of the ML COP RMS; and the trunk-cephalic length explained 10% of the sway velocity and 5% of the COP area.

4. Discussion

The aim of this study was to analyze the influence of sensory and anthropometric variables in the postural sway. The main result suggests that the visual information changes the relation of the anthropometric variables and the postural sway. When the eyes were closed, only the mediolateral postural sway could be explained by body size. This result supports the hypothesis that sensory information constrains the importance of the inverted pendulum variables for the postural sway.

Modeling the standing posture as an inverted pendulum is a strategy to reduce the number of biomechanical variables that could affect the body (center of mass) or postural (center of pressure) sways. The parameters of the inverted pendulum model are body stiffness and the inertia about the ankle [1]. Less than 20% of the postural sway (anteroposterior or mediolateral directions) could be explained by any anthropometric parameter. It suggests that the physical parameters of the body (size, mass distribution, and inertial properties of the body segments) can partly explain the behavior of the postural sway. Nevertheless, the assumption that postural sway should be normalized [14] by any body dimension must be carefully adopted.

Table 4: Linear regression analysis on postural balance and the anthropometric variables for the whole group, with eyes opened and eyes closed.

Group condition	Variables	Height (cm) β (P)	Trunk-cephalic length (cm) β (P)	Lean mass (kg) β (P)	Waist-hip ratio (cm) β (P)	r^2 adjusted
Eyes opened	ML COP RMS (cm)	+0.006 (<0.001)	—	—	—	0.12
	AP COP RMS (cm)		+0.008 (0.006)	—	—	0.06
	Sway velocity (cm/s)	+0.003 (0.001)	—			0.10
	COP area (cm^2)	+0.009 (<0.001)	—	—	—	0.11
Eyes closed	ML COP RMS (cm)	+0.007 (<0.001)	—	—	—	0.18
	AP COP RMS (cm)	—	—	—	—	
	Sway velocity (cm/s)	—	+0.004 (<0.001)	—	—	0.10
	COP area (cm^2)	—	+0.007 (0.01)	—	—	0.05

r^2: r adjusted; $^*P \leq 0.05$, β: beta value; ML COP RMS: medial lateral center of pressure root mean square; AP COP RMS: anteroposterior center of pressure root mean square.

The correlation between postural sway and body size and mass distribution was more common when the eyes were opened. The postural sway changes without the visual information [16, 17]. In general, the postural sway increases when the eyes are closed [14]. When the eyes are closed, the difference between the position of the center of mass and center of pressure increases [2] and the muscle activation at the ankle also must increase [3]. Therefore, probably the stiffness at the ankle and other joints may increase in an attempt to decrease the chance of falling.

The postural sway also reflects the ankle and hip strategies to maintain the standing position [2]. Considering the inverted pendulum model [1], anterior-posterior postural sway is related to the ankle strategy [2] and the mediolateral is related to the hip strategy when the feet are parallel [2]. Adopting this model to explain the postural sway, the motion of a pendulum depends on its length, mass, and stiffness. For the postural control, it means that the ankle strategy is affected by the body mass, height, and ankle stiffness. Our results show that trunk-cephalic length explains a small portion (6%) of the anterior-posterior postural sway and height and postural sway are positively correlated. On the other hand, the taller the participant is, the worse the balance will be [14, 18–20]. Berger et al. [21] stated that ankle displacements and the response of the gastrocnemius muscle increase with taller subjects. Allard et al. [22] and Lee and Lin [23] reported that ectomorph individuals presented greater postural sway than endomorph and mesomorph individuals because they have a higher center of mass.

The trunk-cephalic length, or the head-trunk length, was positively correlated to the mediolateral postural sway. For the hip strategy, under the inverted pendulum model, the mediolateral postural sway depends on the body mass, the head-trunk length, and the hip and lower back joints stiffness [2]. According to the regression analysis, the importance of the trunk-cephalic length for the postural sway decreased when the participants closed their eyes; otherwise, the height has increased its importance for mediolateral sway postural with closed eyes.

The waist-hip ratio was positively related to the mediolateral postural sway. Menegoni et al. [10] suggested that the waist-hip ratio leads to bad postural control. It is possible that the fat mass concentration in the chest and abdomen (android shape) increases the load on the hips, explaining the larger ML COP. Therefore, we showed that the lean mass was positively correlated to the postural control. Those results suggest that lower lean body mass and higher waist-hip ration can be risk factors for the postural control.

The absence of visual information changes the importance of body composition and dimensions. The regression analysis showed that, under the closed eyes condition, only the anthropometric variables explained the postural sway. When the visual information is suppressed, a greater importance is required from the somatosensory and vestibular systems for the postural control. The afferent information is important to set the muscle activity and tonus in an adequate level. And we just showed that the body lean mass is related to postural sway. Winters and Snow [18] correlated the bone mineral density with the anthropometric variables and found an interrelation between them; but they reported that it did not influence the postural balance.

How important are the mechanical properties of the body for the postural control when no visual information is available? If the visual information is absent, the influence of body composition on the balance postural vanishes, while the importance of the body dimensions increases. Is the inverted pendulum model able to explain how the postural control regulates the body sway? Our results suggest that the ankle and hip strategies have opposite behaviors in relation to vision and the inverted pendulum. Nevertheless, the lengths of the single and the double inverted pendulum are important for the postural sway. Attempts are proposed to understand how the nervous system controls the postural sway during standing [1, 2]. If the postural control fails during an unstable condition, the person may trip when it walks over an obstacle, falls down, and may have an injury. The higher postural sway is related to higher risk to fall down in the elderly [24].

5. Conclusion

The postural control depends on body composition and dimension. This relation is mediated by the sensory information.

The height was the anthropometric variable that most influenced the postural sway.

Conflict of Interests

The authors wish to confirm that there is no known conflict of interests associated with this publication and there has been no significant financial support for this work that could have influenced its outcome.

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Review Article

A Decade of Progress Using Virtual Reality for Poststroke Lower Extremity Rehabilitation: Systematic Review of the Intervention Methods

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Objective. To develop a systematic review of the literature, to describe the different virtual reality (VR) interventions and interactive videogames applied to the lower extremity (LE) of stroke patients, and to analyse the results according to the most frequently used outcome measures. *Material and Methods*. An electronic search of randomized trials between January 2004 and January 2014 in different databases (*Medline, Cinahl, Web of Science, PEDro,* and *Cochrane*) was carried out. Several terms (*virtual reality, feedback, stroke, hemiplegia, brain injury, cerebrovascular accident, lower limb, leg*, and *gait*) were combined, and finally 11 articles were included according to the established inclusion and exclusion criteria. *Results*. The reviewed trials showed a high heterogeneity in terms of study design and assessment tools, which makes it difficult to compare and analyze the different types of interventions. However, most of them found a significant improvement on gait speed, balance and motor function, due to VR intervention. *Conclusions*. Although evidence is limited, it suggests that VR intervention (more than 10 sessions) in stroke patients may have a positive impact on balance, and gait recovery. Better results were obtained when a multimodal approach, combining VR and conventional physiotherapy, was used. Flexible software seems to adapt better to patients' requirements, allowing more specific and individual treatments.

1. Introduction

Virtual reality (VR) is an innovative technology that describes a scenario generated by a computer (virtual world), in which the users can interact. This allows creating multisensorial stimuli that transfer the complexity of the physical world to another controlled environment, in which it is possible to modify and control a great number of physical variables [1]. The computational approach means an important analysis for the motor system in the neuroscience field, which offers the opportunity to unify experimental data in a theoretical framework. This approach allows the patient to make more precise and effective movements through a sensorimotor feedback [2]. With regard to stroke patients, the loss or

impaired ability to walk is one of the most devastating consequences, and gait recovery has been recognized as a primary objective in stroke rehabilitation [3].

For the past decade, there has been a significant progress in the use of VR systems for the recovery of the plegic lower extremity (LE) after stroke, [4–10] with positive results, not only in gait but also in variables such as balance. Apart from specific VR, adaptation of interactive videogames to the stroke patient's rehabilitation provides an interesting and useful approach. Due to the heterogeneity of the different trials in the literature, it is difficult to reach a conclusion concerning the most important aspects to take into account to create an efficient VR system for poststroke lower extremity rehabilitation. The objective of this trial was to make a systematic review of the literature, to update and describe the different VR interventions or interactive videogames that have been used for LE recovery in stroke patients, and to analyze the previous findings according to the most frequently used variables.

2. Material and Methods

2.1. Identification of Trials. A search of articles published between January 2004 and January 2014 was carried out in different electronic databases (*Medline, Cinahl, Web of Science, PEDro,* and *Cochrane*) by two independent reviewers. This term combination (virtual reality OR feedback) AND (stroke OR hemiplegia OR brain injury OR cerebrovascular accident) AND (lower limb OR leg OR gait) was used to find those sources considered to be relevant.

Reviewers also performed a manual bibliographic search of full texts and reviews, in order to identify additional relevant studies, including congress contributions and cited references.

2.2. Eligibility Criteria. Randomized controlled trials were only included if the study design compared pre- and postintervention values. Likewise, the treatment had to be specifically referred to VR techniques or interactive videogames used for the LE recovery in stroke patients rehabilitation, either compared to an alternative intervention or not. The studies were considered for review purposes only if patients had a single stroke episode, with no restrictions of mean age of length or recovery. On the contrary, case series studies, single case clinical reports, and review studies were not included for assessment in the present systematic review. Likewise, studies with a sample of hemiplegic subjects after a medical diagnosis different from ischemic or hemorrhagic stroke were excluded.

After a first selection process, only those trials, in which proper assessment scales (with validity and reliability) were used, were finally included.

The *PEDro* scale was used to analyse the methodological quality of each trial by two independent reviewers [11].

Figure 1 shows the flow chart diagram of the study selection process.

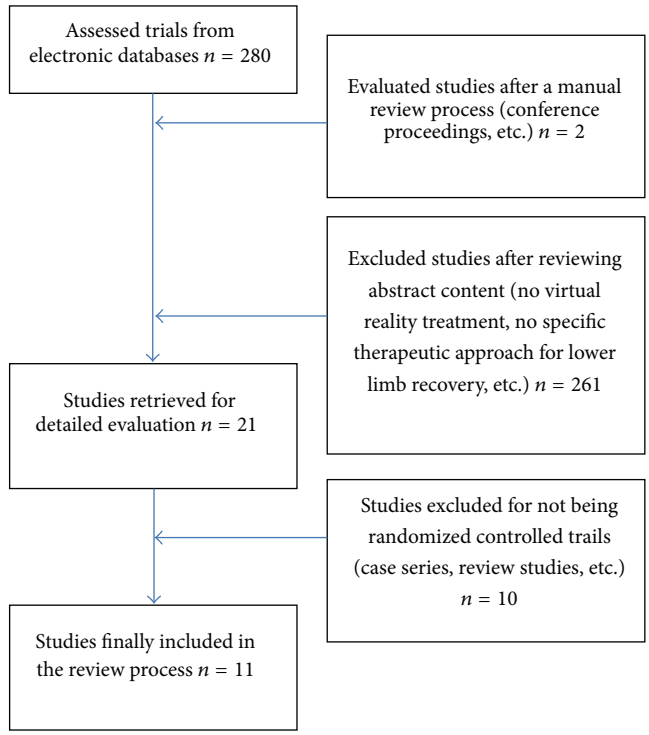

Figure 1: Flow chart of the articles selection.

3. Results

After the selection process, 11 randomized controlled trials (RCT) were included for assessment purposes [3, 12–21], accounting for a total of 9 different projects since two groups of authors [14, 16, 18, 21] published study data from the same trial in different articles. All the included trials account for a total of 183 participants (Table 1). To the best of authors' knowledge, there are no RCT on this matter before 2004. Most trials have shown moderate to high quality, with total scores ranging from 6 to 10 in the *PEDro* scale. Table 1 lists the specific scores of each study. Table 1 also includes the number of patients in each study, as well as aspects related to the mean age of the study sample, the time since the stroke onset (taken six months as a reference to differentiate subacute poststroke patients from chronic poststroke subjects), the treatment approach used in the trial, the number and average of treatment sessions, and the main outcome measures.

The present findings show that sample sizes were generally quite small in all studies, with a mean sample of around 20 subjects, equally distributed into the intervention or the control group. The mean age of the participants ranged between 47 and 66 years. The youngest study samples were those from Gil-Gómez et al. [17] and Park et al. [20] with a mean age below 50 years (Table 1).

All patients can be considered to be in a chronic poststroke stage (more than 6 months after the stroke episode). However, even though they were all in a chronic stage, there is a great heterogeneity between studies with regard to the time of recovery after the stroke and the baseline study assessment, which makes it difficult to compare between trials. While

Table 1: Synthesis of results.

Author year *PEDro*	Group/sample	Age range years	Time since stroke onset	Type of VR	Type of feedback	Sessions	Outcome measures
Jaffe et al. (2004) [12] *PEDro*: 4/10	ROT: 10 OVR: 10	60 mean	Chronic 3-4 years	Immersive: VR scripting language	Visual, vibrotactile, auditory	6 ses. 1 h/d, 3 d/w, 2 w	Stride gait analysis system, obstacle test, 6MWT
You et al. (2005) [13] *PEDro*: 5/10	VR: 5 C: 5	54 mean	Chronic 1-2 years	Semi-immersive: IREX VR system	KR/KP	20 ses. 1 h/d, 5 d/w, 4 w	FAC, MMAS, fMRI
Mirelman et al. (2009) [14] *PEDro*: 5/10	R + VR: 9 R: 9	61 mean	Chronic 3-4 years	Semi-immersive: Rutgers Ankle	KR/KP	12 ses. 1 h/d, 3 d/w, 4 w	Gait speed, 6MWT, PAM (gait spatiotemporal parameters)
Yang et al. (2008) [3] *PEDro*: 6/10	TT: 9 VRTT: 11	58 mean	Chronic 6 years	Immersive: visual screen, 3D Fastrak Polhemus	Auditory, visual	9 ses. 20 m/d, 3 d/w, 3 w	Walking speed, WAQ, ABC scale
Kim et al. (2009) [15] *PEDro*: 6/10	CP + VR: 12 CP: 12	51 mean	Chronic 2-3 years	Semi-immersive: IREX VR system	KR/KP	16 ses. CP: 40 m/d, 4 d/w, 4 w VR: 30 m	BPM, BBS, 10MWT, MMAS, GAITRite
Mirelman et al. (2010) [16] PEDro: 3/10	R + VR: 9 R: 9	61 mean	Chronic 3-4 years	Semi-immersive: Rutgers Ankle	KR/KP	12 ses. 1 h/d, 3 d/w, 4 w	Gait analysis (kinematic and kinetic parameters)
Gil-Gómez et al. (2011) [17] *PEDro*: 6/10	WBB: 9 CR: 8	47 mean	Chronic 1-2 years	Nonimmersive: WBB easy balance virtual rehabilitation	Auditory, visual	20 ses. 1 h/d (3–5 d/w)	BBS, BBA, ART, TST, ST, 1MWT, 10MWT, TUG, 30SST
Fritz et al. (2013) [18] *PEDro*: 6/10	Game: 15 C: 13	66 mean	Chronic 3-4 years	Nonimmersive and semi-immersive: WBB and play station	Auditory, visual	20 ses. 1 h/d, 4 d/w	FMS, BBS, TUG, 6MWT, Dynamic Gait Index, 3MWT, Stroke Impact Scale
Cho and Lee (2013) [19] *PEDro*: 7/10	CP + TRWVR: 15 CP + TT: 15	64 mean	Chronic 1-2 years	Semi-immersive: treadmill + screenshot of real-world video recording	Auditory, visual	18 ses. 30 m/d, 3 d/w, 6 w	BBS, TUG, gait performance (GAITRite)
Park et al. (2013) [20] PEDro: 4/10	VR + CP: 8 CP: 8	47 mean	Chronic 11 years	Immersive VR-based postural control program (head-mounted display)	Visual	12 ses. 30 min/d, 3 d/w, 4 w	10MWT, GAITRite (gait parameters)
Cho and Lee (2014) [21] *PEDro*: 7/10	CP + TRWVR: 15 CP + TT: 15	64 mean	Chronic 1-2 years	Semi-immersive: treadmill + screenshot of real-world video recording	Auditory, visual	18 ses. 30 m/d, 3 d/w, 6 w	BBS, TUG, Postural Sway Platform System, GAITRite

3MWT: 3-Minute Walking Test; 6MWT: 6-Minute Walking Test; 10MWT: 10-Meter Walking Test; 30SST: 30-Second Sit to Stand Test; ABC scale: activities-specific balance confidence; ART: Anterior Reach Test; BBA: Brunel Balance Assessment; BBS: Berg Balance Scale; BPM: Balance Performance Monitor; C: control (no intervention); CP: conventional physiotherapy; FAC: Functional Ambulatory Scale; FMS: Fugl-Meyer Scale; KP: knowledge of performance; KR: knowledge of results; MMAS: Modified Motor Assessment Scale; OVR: virtual obstacle training; PAM: Patient Activity Monitor, Ossur (walking activity); R: robot; ROT: real obstacles training; ST: Steeping Test; TRWVR: treadmill training based real-world video recording; TST: Timed Stair Test; TT: treadmill training; TUG: Timed Up Go; VRTT: virtual reality-based treadmill training; WAQ: Walking Ability Questionnaire; WBB: *Wii Balance*.

studies from You et al. [13], Gil-Gómez et al. [17], Cho and Lee [19], and Cho and Lee [21] included patients with a 1- to 2-year mean period of time after the stroke, Park et al. [20] evaluated individuals with more than 10 years of poststroke recovery.

The total number and week average of treatment sessions has been also taken into account when comparing the different studies. The number of sessions varies between 6 and 20, with a mean average of 15 sessions.

According to the eligibility criteria, the study design of all trials was based on a comparison of postintervention and preintervention scores after the VR treatment. In most cases, the VR treatment lasted one hour per day. However, the treatment duration was inferior in some trials, between 20 minutes and one hour, with a mean frequency of 3 times per week. The VR intervention approach was compared to either conventional physiotherapy, an alternative intervention, or no treatment. In some trials, the impact of a multimodal approach combining VR and conventional physiotherapy was assessed.

As previously mentioned, there are different types of virtual environments, according to their level of immersion:

(i) nonimmersive VR, in which the computer generates environments that are projected either on a screen or on a wall in front of the patient;

(ii) semi-immersive VR or augmented reality, which overlays virtual images to real images increasing the informative content of the real ones;

(iii) immersive VR, in which the viewer is a part of the environment. One example is head mounted display (HMD), a device with a helmet that provides images within a computer, as a unique visual stimulation [1].

All the assessed trials used different types of virtual environments. Immersive systems are purported to be more effective because they provide a more intense feeling of reality; however, they may provoke "cybersickness" (symptoms such as vomiting, dizziness) in some participants [22]. For this reason, most authors prefer to choose semi-immersive systems.

The visual/auditory feedback was used in most of the studies, along with the knowledge of results/performance (KR/KP) approach. On the other hand, the vibrotactile feedback was not frequently used.

We will briefly describe the VR systems that were used in each trial.

(i) Jaffe et al. [12] compared two training groups: real and virtual. In the first one, real obstacles were used whereas in the virtual group a head-mounted device was used to observe the simultaneous registration of the legs' real movement, introducing virtual stationary images of obstacles and getting a patient's feedback. The virtual obstacle training generated greater improvements in gait velocity compared with real training during the fast walk test and the self-selected walk test. Overall, subjects showed clinically meaningful changes in gait velocity, stride length, walking endurance, and obstacle clearance capacity as a result of either training method.

(ii) You et al. [13] compared a control group (no intervention) with a group that used the IREX VR system, which allows interaction (within cybergloves, etc.) with virtual objects in environments that can be individualized. This allows optimizing motor relearning. Three specific exercises were included: going up/down stairs, diving among sharks, and snowboarding. VR seems to induce cortical reorganization from aberrant ipsilateral to contralateral primary sensorimotor cortex activation. In this study, motor function was significantly improved after VR.

(iii) Mirelman et al. [14, 16] compared a training treatment based on a robotized system of VR (Rutgers Ankle) with another intervention only based on a robot. The Rutgers Ankle system consists of a Stewart platform with six-degree feedback strength of foot freedom and a screen that allows the patient to train the LE. While the patient is sitting and simulates driving a boat or a plane, parameters are individualized by the physiotherapist. The LE training that combined a robotic device and VR improved walking ability in the laboratory and the community (velocity and distance walked) and showed a higher impact than the robot training alone. The main observed effects after training included improved motor control at the ankle, which enabled the patient to do other functional improvements.

(iv) Yang et al. [3] compared training on a treadmill with training in a VR system composed of a treadmill, a screen with a high vision field, and a 3D capture system (Fastrak Polhemus) for leg movements. Virtual scenarios represented a field, some obstacles, and a task to cross a street. Gait speed was increased in each session. VR group improved significantly more than treadmill group in walking speed and community walking time in a short term and in Walking Ability Questionnaire score at a follow-up period.

(v) Kim et al. [15] measured the additional effect of the VR system to a conventional physiotherapy approach. The same system as previously described by You et al. [13] was used. The VR group improved BBS scores, balance, and dynamic balance angles (ability to control weight shifting) compared to the patients that underwent only physiotherapy.

(vi) Gil-Gómez et al. [17] compared an intervention program with the *Nintendo Wii Balance Board* (WBB) with eBaViR to a conventional physiotherapy treatment in patients with brain damage. Although 6 of the 17 patients had hemiplegia not secondary to a stroke, this trial was also included for the treatment approach that it shows since flexible software for *Wii*, specific for rehabilitation, was designed. Patients using WBB had a significant improvement in static balance (BBS and ART), compared to patients who underwent traditional therapy. With regard to dynamic balance, there were no differences between study groups.

(vii) Fritz et al. [18] compared an experimental group that used *Nintendo Wii* (*Wii Sports* and *Wii Fit*) and *Play Station* (*Eye Toy Play 2* and *Kinetic*) with a control group that underwent no intervention. No statistically significant differences in the comparison between or within groups were found, either in the short term or in the follow-up process.

(viii) Cho and Lee [19, 21] also combined VR and conventional physiotherapy. One group of patients underwent a training treatment on a treadmill and another group used the *TRWVR system* (*treadmill training based real-world video recording*), which uses different virtual environments registered in the real world (paths, obstacles, etc.). The TRWVR group showed better results in walking balance (BBS and TUG), dynamic balance gait, and spatiotemporal parameters (velocity and cadence) than the treadmill group.

(ix) Park et al. [20] compared a VR approach with a conventional physiotherapy treatment. They used a programme for improving postural control and gait ability through a visual feedback, comparing the reference scenario of the movement and the real movement. In the gait parameters, subjects in the VR group showed a significant improvement, except for cadence immediately after training and at the follow-up when compared to the conventional physiotherapy group. In the comparisons between groups, the VR group also showed significantly greater improvement only in stride length compared with the control group. On the contrary, no significant differences were found in other gait parameters.

4. Discussion

The validity and reliability of the outcome measures used in the different trials are crucial to determine the quality of the findings. The most frequently used outcome measures were gait speed, balance, and improvement of the motor function.

4.1. Gait Speed. Gait speed was used as an outcome assessment tool in 8 of the 9 selected trials. On the other hand, You et al. [13] evaluated motor function, functionality, and cortical changes by means of nuclear magnetic resonance.

Gait speed is considered to be a significant, sensitive, and reliable tool to evaluate the impairment severity and the community's functional ability to walk [9].

The *6-Minute Walking Test* has been the most frequently used test to measure gait speed. Systems of movement analysis to measure space-time parameters and the *10-Meter Walking Test* were also used.

Most of the trials evaluated spontaneous gait speed. However, some authors (Jaffe et al. [12] and Fritz et al. [18]), who observed a significant increase of maximum speed in the VR group but no significant changes in spontaneous speed, considered both spontaneous and maximum speeds. A plausible reason may be the small number of treatment sessions [12] and the lack of physiotherapist's guidance [18] during the intervention.

In the rest of the trials, there was a significant difference in the spontaneous gait speed in the VR group compared to the control group, except for Gil-Gómez et al. [17]. A possible reason is that the intervention aimed to improve balance, but exercises were not oriented for improving speed. Park et al. [20] designed an intervention program focused on improving posture. In addition, poststroke time of recovery in both groups was quite long and, therefore, we may assume less neuronal plasticity in the study sample.

4.2. Balance. Balance was used as an outcome measure in 4 of the 9 selected trials. Mirelman et al. [14, 16] only used the *Berg Balance Scale* (BBS) in the baseline assessment but not in the postintervention evaluation. Jaffe et al. [12] also employed a balance test to describe the initial stage. Although some authors used additional scales, the BBS was used in all of them.

Three out of 4 trials that compared postintervention balance scores with baseline assessment concluded significant improvement results in the VR group for balance as measured by means of the BBS. Fritz et al. [18] did not find any significant differences in their first pilot study. Since they used two different videogames, it is difficult to state the specific influence of each system on the results. However, as stated by them, a physiotherapist's guidance is necessary to facilitate and orientate the patient about the most suitable motor strategies in order to find better and clinically significant results [6, 18].

Gil-Gómez et al. [17] observed significant differences in balance, by comparing an intervention group with the *Wii Balance Board* (WBB) and a control group with a conventional physiotherapy approach. It must be pointed out that, in the WBB group, the software was flexible, and the physiotherapist was constantly adjusting the degree of difficulty and other parameters. This issue may demonstrate the importance of physiotherapist's intervention in this type of systems, since results are poorer when the patient is not guided [18].

4.3. Motor Function of the Plegic LE. Improvement of motor function was evaluated in 3 trials. You et al. [13] and Kim et al. [15] used the MMAS (*Modified Motor Assessment Scale*). On the contrary, Fritz et al. [18] used the *Fugl-Meyer* (FM) scale, in which it is specific for LE assessment. Only the first two studies obtained significantly better results in the VR group.

Other authors performed measurements of space-time and kinematic-kinetic parameters within systems of gait analysis [12, 15, 16, 20, 21]. They all obtained a significant improvement in the VR group, except for Park et al. [20]. A plausible reason is that the time of recovery after the stroke was too long in the latter study.

In recent years, the use of VR systems for the functional recovery of the gait poststroke has increased. Despite the fact that some systematic reviews have been published in the issue, most of them are focused on the motor recovery of the upper limb [23]. Therefore, it is important to build stronger evidence about the impact of VR treatment on LE recovery, because only 4 articles [24, 25] on this issue have been formerly reviewed.

As a main finding, it seems that patients should receive a minimum number of ten sessions for the intervention to be effective. In most cases, better results were obtained when a multimodal approach combining VR and conventional physiotherapy was used. However, a more homogeneous methodology is crucial and future research is needed to elucidate the possible effect of each individual system in gait and LE recovery after stroke. It is also important to carry out a proper baseline assessment to establish which system

is the most appropriate for handling the different disorders (balance, functionality, gait speed, etc.). An early treatment after stroke is also important [26], because intervention seems to be less effective when the recovery time after stroke is too long. Neural plasticity may be a factor to understand this aspect.

5. Conclusions

The number of trials is small and the evidence is limited. Nevertheless, the present findings seem to suggest that VR intervention has a positive impact on balance and gait after stroke.

The use of commercial videogames in rehabilitation seems to be useful. However, a proper physiotherapist's guidance is needed to facilitate and orientate the patient on the most suitable motor strategies. It is also necessary to develop more flexible software to individualize treatments and adapt the intervention to the patients requirements.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Normative Data for an Instrumental Assessment of the Upper-Limb Functionality

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Upper-limb movement analysis is important to monitor objectively rehabilitation interventions, contributing to improving the overall treatments outcomes. Simple, fast, easy-to-use, and applicable methods are required to allow routinely functional evaluation of patients with different pathologies and clinical conditions. This paper describes the Reaching and Hand-to-Mouth Evaluation Method, a fast procedure to assess the upper-limb motor control and functional ability, providing a set of normative data from 42 healthy subjects of different ages, evaluated for both the dominant and the nondominant limb motor performance. Sixteen of them were reevaluated after two weeks to perform test-retest reliability analysis. Data were clustered into three subgroups of different ages to test the method sensitivity to motor control differences. Experimental data show notable test-retest reliability in all tasks. Data from older and younger subjects show significant differences in the measures related to the ability for coordination thus showing the high sensitivity of the method to motor control differences. The presented method, provided with control data from healthy subjects, appears to be a suitable and reliable tool for the upper-limb functional assessment in the clinical environment.

1. Introduction

Kinematic analysis plays a fundamental role in the decision making process in the clinical and surgical treatment of the lower limb (LL) [1–6]. Indeed, it is a fact that instrumental gait analysis has widely been used in clinics for almost twenty years for assessing, planning, and monitoring the results of therapies in the rehabilitation of neurological patients [7–9]. Similarly, it is recognized that the success of upper-limb (UL) rehabilitation treatments largely depends on the possibility of defining the patient's functional pathological profile through objective quantification of motor performances [10–13]. Although poly-EMG is used routinely to plan UL treatments [14–19], kinematic analysis is yet much less commonly applied. The transfer of knowledge and experience gained in the LL movement analysis to the UL domain is actually hindered by several sources of complexities, related to both the kinematic model and the evaluation protocol to be used.

Regarding the model, being the UL, a kinematically redundant multijoint system [20] performing spatial movements, sophisticated models for skeleton-markers matching should be used for movement analysis. (It is provided with multiple degrees of freedom so that different strategies can be used to perform the same goal-directed movement.) Unfortunately, a complex model does not comply with clinical requirements such as limiting the tracking setup overhead and avoiding time-expensive calibration procedures requiring a full, yet in patients rarely available, UL range of motion. Consequently, notwithstanding different 3D models and computational methods have been proposed over the last two decades [21–28], how to practically apply these models

into the clinical practice is still an open problem and their impact on clinical procedures is therefore still limited. Only few recent clinical studies are in fact, to the best of the authors' knowledge, positively based on kinematic analysis [28–30].

Regarding the evaluation protocol, the complexity derives from both the very nature of the arm, which is inherently multitasking, and the nature of the performed movements, that are far from being restricted to cyclic periodical uniform patterns like the gait [31]. Consequently, appropriate methods and evaluation protocols have to be defined and standardized in order to define a set of normative data to be used as reference. A recent review of clinical studies, based on the kinematic characterization of only reaching movements, highlighted how the scattering of the many diverse methodological approaches is a risky factor for preventing any comparative analysis of published results [32]. Nonetheless, the major problem of standardization to date is represented by the lack of protocols for routinely clinical application. No consistent guidelines and routines are in fact provided in the clinical literature about the type (e.g., single or multijoint [33]) or the application mode (e.g., single shot movements or continuous cyclic gestures) of UL movements to be preferably investigated. Even more importantly, there is a remarkable lack of clarity and standardization in procedures about following either a functional or a segmental approach. The lack of methodological works about shared and standard protocols for data acquisition (e.g., conveniently usable sampling procedures according to models) and analysis of UL movements (e.g., methods to effectively evaluate and compare acquired data) can be also reckoned as the main cause of the to-date unavailability of normative data. Peculiarly, no quantitative description on UL joint-level kinematics during the performance of a selected set of common tasks is available in the literature. This lack is particularly important for the treatment and the recovery assessment of functional tasks involving shoulder and elbow (coordinated) movements, essential for the usage of the hand and highly important in major therapies for functional ADL capabilities recovery. In particular, the shoulder flexion coupled to the elbow extension (reaching movement, RM) and the elbow flexion (hand-to-mouth movement, HtMM), both against gravity, complementarily define a subset of motor abilities to be effectively assessed in structured rehabilitation practice.

Such motivations are the basis of the Reaching and Hand-to-Mouth Evaluation Method, a procedure used to assess the UL functionality in neurological patients. It has been applied for fifteen years to patients affected by different pathologies (stroke, TBI, and Parkinson and Friedreich's disease) to evaluate their residual functionality. It was also used to test the effects of the Bilateral Subthalamic Deep Brain Stimulation [34]. Moreover, the method in its final form, the one presented and tested in this work, was effectively applied to verify the efficacy of the constraint induced movement therapy (CIMT) [35, 36].

This work describes deeply the model and the assessment protocol of the Reaching and Hand-to-Mouth Evaluation Method and presents a dataset of normative data to be used as reference in the UL functional evaluation.

2. Material and Methods

2.1. Participants. Forty-two healthy (neurologically and orthopedically intact) subjects, distributed uniformly between 18 and 80 years old, were included in the study. They all were unaware of the purpose of the study and had to give written informed consent before inclusion in the study. Ethical approval of the evaluation protocol was granted by the local ethical committee at Como Valduce Hospital.

2.2. Study Design. Prior to testing, all subjects were questioned and clinically evaluated for the presence of neurological or orthopaedic signs. They were also tested to define the hand dominance and the degree of dominance using the Edinburg Inventory Test [37]. Subjects were then evaluated through kinematic analysis using the Reaching and Hand-to-Mouth Evaluation Method, systematically formulated in this work and briefly outlined in [36]. Sixteen subjects were reevaluated two weeks after the first testing session in order to estimate the test-retest reliability of the presented method of analysis. The method sensitivity was tested comparing dominant/nondominant limb results and clustering all subjects into three groups, namely, young (18–35 years), middle-aged (36–50 years), and elderly subjects (51–80 years).

2.3. Functional Tasks and Behavioural Testing. The tasks to be tested were defined with the purpose of assessing the residual functional capability of the patient at the shoulder and elbow levels. More specifically, the objective was to assess the ability of the patient to (1) extend the elbow while flexing the shoulder and (2) flex the elbow, both movements preformed against gravity. These are key movements from a functional point of view because they allow, respectively, (1) reaching for objects placed in front of the subject up to the shoulder height and (2) taking objects towards the body and face. As previously introduced, they have been named the reaching movement (RM) against gravity and the hand-to-mouth movement (HtMM). Such patterns, in fact, allow the study of shoulder and elbow compound movements used in ADLs, such as reaching for objects and eating. The tasks were designed to be performed in a sitting position to apply the proposed method even to all those patients who may not assume the standing position. Such considerations contributed to establishing the evaluation protocol hereafter described.

2.4. Protocol Description. The subjects sat on a chair, adjustable for height, with the feet resting on the floor and the knees and hips bent at 90 degrees. In the rest position, both hands were lying on the thighs, and the arms were positioned with flexed elbow and slightly extended shoulder. Each subject, starting from rest position, was asked to carry out the two movements without moving his/her back away from the backrest. The two selected movements were performed in the following ways:

(i) RM: each subject was asked to move one hand toward a target located in front of the subject at shoulder

height, at a distance slightly longer than that of the fully extended UL (Figure 1(b) and Figure 3);

(ii) HtMM: each subject was asked to move one hand to the mouth and touch it with the palm; no special indication was given about how to move the arm and forearm, with the explicitly asked exception of not to move the head toward the hand (Figure 1(c) and Figure 4).

Both movements were performed at a speed freely chosen by the subject. At the "go" command, each subject repeatedly performed the movements, without pausing, until the tester operator issued a "stop" command. Two trials of at least 12 repetitions were acquired for each test and each movement. (The trial was repeated immediately if the subject did not complete every single movement (reach the target and return to starting position) or if he/she did not respect the instruction not to move trunk and head.) Each subject was asked to perform the movements starting randomly with either the dominant or the nondominant hand.

2.5. The Experimental Setup and Kinematic Model. Data were collected with a 3-dimensional (3D) optoelectronic motion tracking system (8 TVc 100 Hz; Elite B|T|S, Italy). However, the protocol was tested positively also using 6 TVc only. Markers were placed using the setup shown in Figure 1; positions and derivatives were filtered using a low-pass, second-order Butterworth filter (cut-off frequency 5 Hz). In order to limit the overall setup time and facing the stringent requirements of the clinical practices, only 5 markers were used to track the arm kinematics. They were applied to $M1$ spinous process of D5, $M2$ spinous process of C7, $M3$ acromion, $M4$ lateral epicondyle of the elbow, and $M5$ styloid process of the ulna (Figure 1(a)). A sixth marker for the RM, namely, the target marker $M6 \equiv T$, was placed in front of the subject at exceeding reaching distance (Figure 1(b)). For the HtMM, the virtual target marker position was estimated on $M2$ coordinates (spinous process of C7). In other words, the target was supposed to be placed 10 cm above and 5 cm in front of $M2$, that is, virtually inside the mouth (Figure 1(c)). Neglecting the pronation/supination the UL can be modelled as a 4-degree-of-freedom kinematic serial chain. This is surely a valid assumption for the RM; it may not be for the HtMM and, therefore, the possible introduced error on the elbow angle due to pronatoin/supination was estimated. The results are reported in Section 3.5. (The error analysis is not presented in the paper for the sake of simplicity, however, the description of the procedure can be asked to the corresponding author). Let us denote by $\{b\}$ the reference frame (Figure 3) centred in the shoulder and by X, Y, and Z the coordinate axes defining the principal body planes, that is, XY-sagittal, YZ-frontal, and XZ-transverse. The centres of rotation of the shoulder S, the elbow E, and the wrist W are

$$S \equiv [x_S, y_S, z_S]^T_{\{b\}},$$
$$E \equiv [x_E, y_E, z_E]^T_{\{b\}},$$
$$W \equiv [x_W, y_W, z_W]^T_{\{b\}}. \tag{1}$$

Let $\mathbf{u}_a \equiv \mathbf{u}_a(E, S)$ and $\mathbf{u}_f \equiv \mathbf{u}_f(W, E)$ be the unit vectors of the arm and forearm, respectively, computed as

$$\mathbf{u}_a = \frac{E - S}{\|E - S\|}, \qquad \mathbf{u}_f = \frac{W - E}{\|W - E\|}. \tag{2}$$

The elbow angle (EA) and the angle of arm flexion (AAF) can be calculated as

$$\mathrm{EA} = \frac{180}{\pi}\mathrm{acos}\left(\mathbf{u}_a \cdot \mathbf{u}_f\right),$$
$$\mathrm{AAF} = 90 + \frac{180}{\pi}\mathrm{atan}\frac{\mathbf{u}_a \cdot \mathbf{u}_y}{\mathbf{u}_a \cdot \mathbf{u}_x}. \tag{3}$$

From marker placement description kinematics definition (Figures 1 and 2), the following assumption is valid:

$$S \approx M3, \qquad E \approx M4, \qquad W \approx M5. \tag{4}$$

Therefore, the flexion-extension angle of the elbow EA is calculated from markers $\{M3, M4, M5\}$ positions, while the arm direction is identified with markers $\{M3, M4\}$, yielding AAF.

Anyhow, the introduced approximation between the anatomical joint centres and markers centres affects the joint angles accuracy. An analysis of the introduced inaccuracies was done and the results are presented in Section 3.5.

2.6. Dependent Measures. Among all the possible measurable and calculable variables, a meaningful set of measures was defined to assess the upper-limb functionality. The measures were selected aiming at answering the following three questions: (1) how fast, (2) how extent, and (3) how well controlled are the two performed gestures? The identified set of indexes chosen to answer these questions is made up of movement duration (MD_M), elbow angle (EA_M), mean elbow angular velocity (EAV_M), acceleration coefficient of periodicity (ACP_M), and normalized jerk (NJ_M) for the HtMM; movement duration (MD_R), elbow angle (EA_R), angle of arm flexion (EAF_R), mean target approaching velocity (TAV_R), acceleration coefficient of periodicity (ACP_R), and normalized jerk (NJ_R) for the RM (Table 1).

In fact, the velocity (assessed through MD_M, EAV_M, MD_R, and TAV_R), the ROM (assessed through EA_M, EA_R, and EAF_R), and the level of motor control (assessed through ACP_M, NJ_M, ACP_R, and NJ_R) are important issues to be tested because they are strictly correlated to the possibility of performing ADLs tasks. It is worth mentioning that, by evaluating these two gestures only, an indirect measure of the strength of the elbow and shoulder flexors is also obtained. In other words an answer is given to the following questions: (1) are patients able to flex the elbow against gravity? (2) Are they able to flex the shoulder against gravity?

2.6.1. Motor Control. While the RM and HtMM velocities and ROM may be directly measured through the use of

Table 1: Metrics.

Measure	Task quantities RM	Task quantities HtMM	Description
Time	MD_R	MD_M	Movement duration
Position	EA_R	EA_M	Elbow angle at end movement EA > 0 *flexion* | EA = 0 *fully extended* | EA < 0 *hyperextension*
	AAF_R		Angle of arm flexion at end of movement AAF > *0 flexion* | AAF = 0 *rest position* | AAF < 0 extension
Velocity		EAV_M	Mean elbow angular velocity
	TAV_R		Mean target approaching velocity normalize on UL length
Repeatability	ACP_R	ACP_M	Coefficient of periodicity calculated on WTD acceleration
Smoothness	NJ_R	NJ_M	Normalized jerk calculated on WTD WTD: Wrist Target Distance (see Figure 1(b) and Figure 1(c))

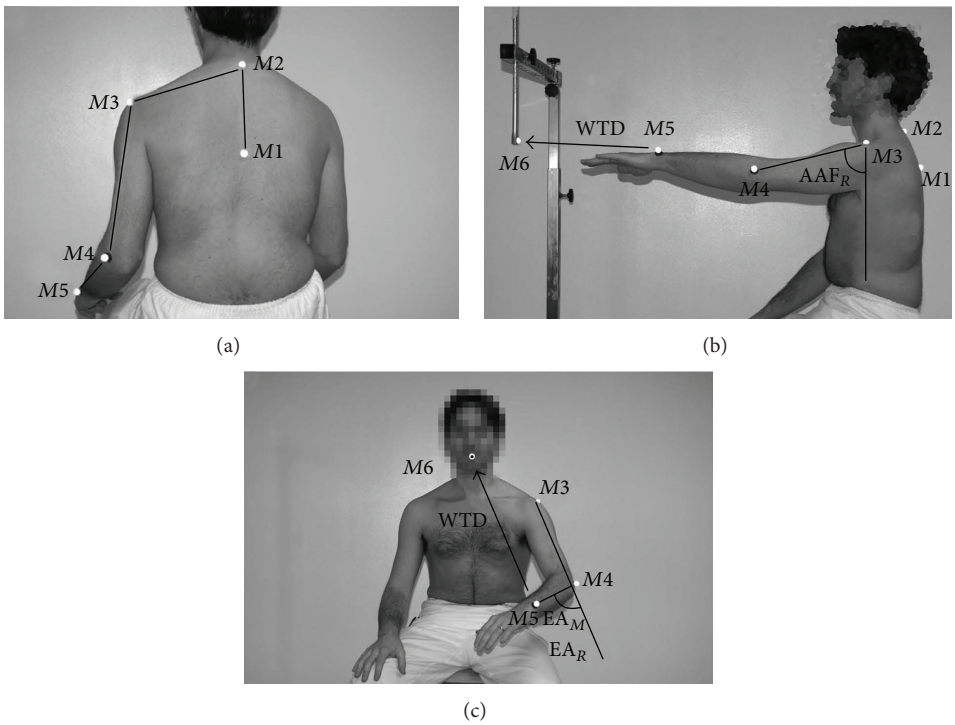

Figure 1: Marker placement. Five hemispherical markers with a diameter of 15 mm are attached to the spinous process of D5 ($M1$), the spinous process of C7 ($M2$), the acromion ($M3$), the lateral epicondyle of the elbow ($M4$), and the styloid process of the ulna ($M5$) (a). $M6$, the target marker, is, for the RM, attached to a tailor-made support adjustable for height (b), whereas it is estimated for the HtMM and placed 10 cm above and 5 cm in front of $M2$, that is, virtually inside the mouth (c). The calculated variables are also indicated.

the selected variables listed above, the choice of using ACP and NJ to assess the amount of motor control requires a deeper explanation. In general terms, it is worth recalling that positioning the hand in the space, thus allowing reaching for objects and manipulation, is the main UL function. The position of the hand in the space during gestures/tasks is the result of coordinated shoulder and elbow joints movements; the wrist accounts for the orientation in space of the hand. For this reason, quantities derived from the position of the wrist (around which the hand rotates) may be used as indirect measures for coordination and motor control ability. Therefore, in the present work, two measures, namely, ACP and NJ, are used to assess movement repeatability and smoothness, respectively. In fact, as explained hereafter, ACP and NJ are calculated on the wrist-to-target distance (WTD), namely, the distance between $M5$ on the wrist and the target marker $M6$.

The repeatability among individual repetitions of reaching as well as hand-to-mouth movement is evaluated by means of the singular value decomposition pattern analysis (SVDPA) [38], a data-driven approach which allows identifying repetitive patterns within quasiperiodic events. The result of the processing is a number between 0 and 1, referred to as the coefficient of periodicity (CP), the periodicity of

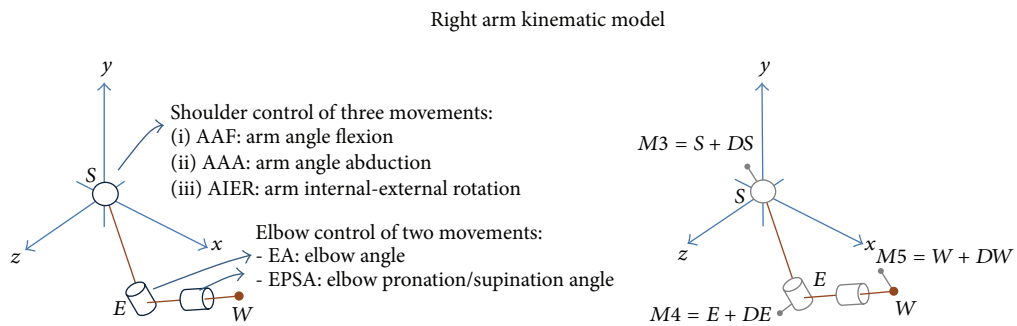

Figure 2: Kinematic model. Upper extremity model with 5 degrees of freedom: arm angle flexion (AAF), arm angle abduction (AAA), arm internal-external rotation (AIER), elbow angle (EA), and elbow pronation/supination angle (EPSA). As dependent measures, only AAF and EA are considered, although also AAA and AIER can be estimated using the marker placement reported in Figure 1. The coordinate axes are those defining the principal body planes, that is, XY-sagittal, YZ-frontal, and XZ-transverse. By contrast, the displacement between the markers and joint centres (DS, DE, and DW) and EPSA cannot be estimated with the model used. Therefore, an analysis of the accuracy was performed and the results are reported in Section 3.5.

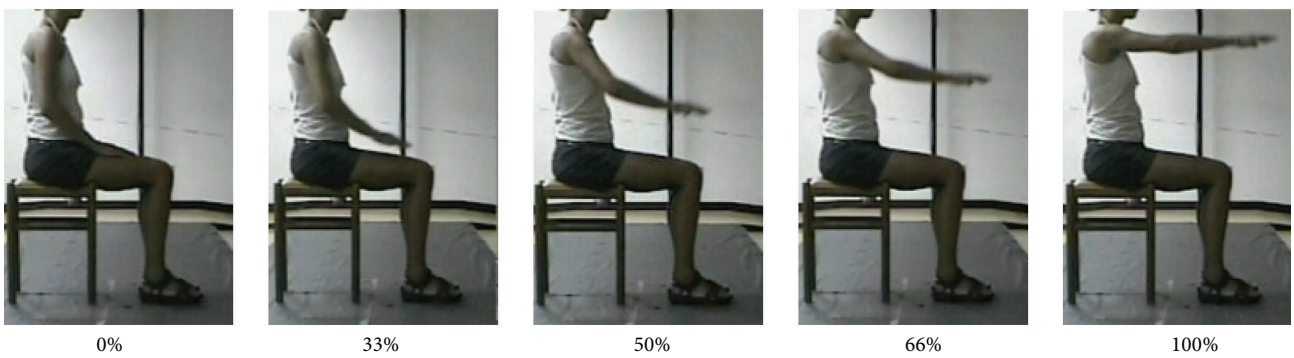

Figure 3: Reaching movement. Frames sequence of the RM performed by a healthy subject.

movement value. The value corresponding to strictly periodic movements, that is, those which are repeated identically over time, is 1; as the movement loses periodicity, the CP gradually decreases. The SVDPA was applied to the target approaching acceleration $\ddot{\mathrm{WTD}}$, which is intrinsically more sensitive than WTD to pattern changes [36]. The acceleration coefficient of periodicity (ACP), thus obtained, is a measure of the consistency of the acceleration profiles across movements along each trial.

The smoothness is evaluated by jerk analysis (i.e., the third derivative) of WTD. It has been shown that the time-integrated squared jerk decreases with increased smoothness of movement; it is, therefore, often used as a measure of the quality of selective motor control [39]. Since the time-integrated squared jerk depends sensitively on the duration and size of the movement, the authors adopted the normalized jerk (NJ) index [40] proposed by Teulings et al.:

$$\mathrm{NJ} = \sqrt{\frac{1}{2} K_J \int_{T_{\mathrm{start}}}^{T_{\mathrm{end}}} \dddot{\mathrm{WTD}}^2 (t)\, dt} \tag{5}$$

with

$$K_J = \frac{\mathrm{MD}^5}{\left\| \mathrm{WTD}\left(T_{\mathrm{end}}\right) - \mathrm{WTD}\left(T_{\mathrm{start}}\right) \right\|^2}, \tag{6}$$

where T_{start} and T_{end} denote times of the beginning and end of movements, respectively. K_J is a normalization factor formulated using the duration of movement MD, that is, the execution time ($T_{\mathrm{start}} - T_{\mathrm{end}}$), and the minimum relative travel length of movement execution. The normalized jerk (NJ) is thus a dimensionless number that is comparable among movements of different durations and lengths. From a mathematical point of view, it is independent of the amount of movement and it may be applied even when the task is not completed due to reduced ROM.

The NJ, inversely proportional to the movement smoothness, is an indirect measure of the ability for coordination.

2.7. Numerical Evaluation. For each of the two trials of the testing session, the selected kinematic quantities were calculated on the 10 central repetitions and only on the forward phase of the movement, that is, reaching the target. For each subject, at each testing session, the mean of the values obtained in 20 movements (10 for each trial) was calculated for every single quantity. For the ACP, which is an index of the repeatability of different movements within the same trial, the calculated values represent the mean of the two trials.

All elaborations were made in Matlab® 6.0.

Figure 4: Hand-to-mouth movement. Frames sequence of the HtMM performed by a healthy subject.

2.8. Statistics. The test-retest reliability of the kinematic variables was estimated by the size of the Pearson product-moment correlation coefficients between data from the first and second testing sessions performed by the subgroup of subjects who were tested twice.

Further, the following nonparametric statistics were used:

(i) the Wilcoxon matched-pairs signed-rank test for comparison of data relative to similar subjects' groups,

(ii) the Mann-Whitney U test for comparison of data from different groups.

The alpha-error significance level was 0.05.

Although data were verified to be normally distributed (Kolmogorov-Smirnov test), the authors decided to use nonparametric tests for comparison because of the small size of the samples; in fact, normality tests can easily fail when the sample is tiny and even in the case of normally distributed data nonparametric tests may still be used (despite losing some evaluation power, around 95%).

3. Results

3.1. Protocol Application. Even if the preparation of the subject and the acquisition procedure were not actually timed, an average total time for the bilateral administration of the whole protocol can be estimated. The application of the 8 markers (D5, C7, and 3 markers per limb, namely, shoulder, elbow, and wrist) takes around 5 to 10 minutes while a single gesture acquisition is very short and takes less than 30 seconds. The total number of evaluated gestures is 8 (2 RM + 2 HtMM per limb) and, therefore, the acquisition time is around 4 minutes. Considering also the time for checking the quality of the acquisition and downtime, the total duration for administering the whole evaluation protocol is less than 30 minutes, thus complying with clinical requirements. In fact, in the authors' experience, in the case of patients, the single gesture acquisition time can increase but it never goes beyond 1 minute; moreover, in the case of severe impaired patients, the number of repetitions may be reduced and each gesture (RM or HtMM) may be evaluated only once; the downtime also can increase as the patient may have to rest between two trials.

3.2. General Movement Description. Figures 3 and 4 show the frames sequences relative to the RM and HtMM, respectively. Figures 5 and 6 show the kinematic patterns of five consecutive RM and HtMM. Mean movements kinematics is also shown in the right panels, highlighting a good movement repeatability along repetitions. Analysing the kinematic patterns, some considerations may be done on the nature of such movements. The RM (Figure 5) is composed of coordinated movements of the shoulder and elbow. It is characterized by a first short preparatory phase in which there is a slight elbow and shoulder flexion which allow lifting the hand from the thigh. The proper movement starts immediately after the detachment of the hand and is mainly characterized by the shoulder flexion, which in turn continues throughout the whole movement duration in a smooth and gradual way. By contrast, the elbow extension mainly takes place in the second half of the movement, once the shoulder has reached approximately 45 degrees of flexion. The resulting wrist-target approaching velocity profile is quite smooth and nearly bell shaped. The HtMM (Figure 6) is instead mainly composed of a coordinated movement of elbow flexion and elbow supination, the first allowing the reaching of the mouth with the hand and the second adding the possibility of touching the mouth with the palm. The elbow supination is not tracked with the present model and, therefore, is not represented in Figure 6. Regarding the shoulder, the movements are limited to some little flexion and abduction to facilitate the reaching of the mouth (although, unnaturally, the HtMM may be actually performed by solely flexing the elbow with the arm fixed to the chest in totally adducted position). Also, for this movement, the resulting wrist-target approaching velocity profile is quite smooth and nearly bell shaped. A common characteristic of the RM and HtMM is the high repeatability across the single repetitions apparently shown in the left-hand panels of Figures 5 and 6 and further put in evidence by the low standard deviations in the graphs showing the average traces (Figures 5 and 6, right-hand panels). Repeatability, beyond linear and angular displacements, extends also to velocity and acceleration profiles.

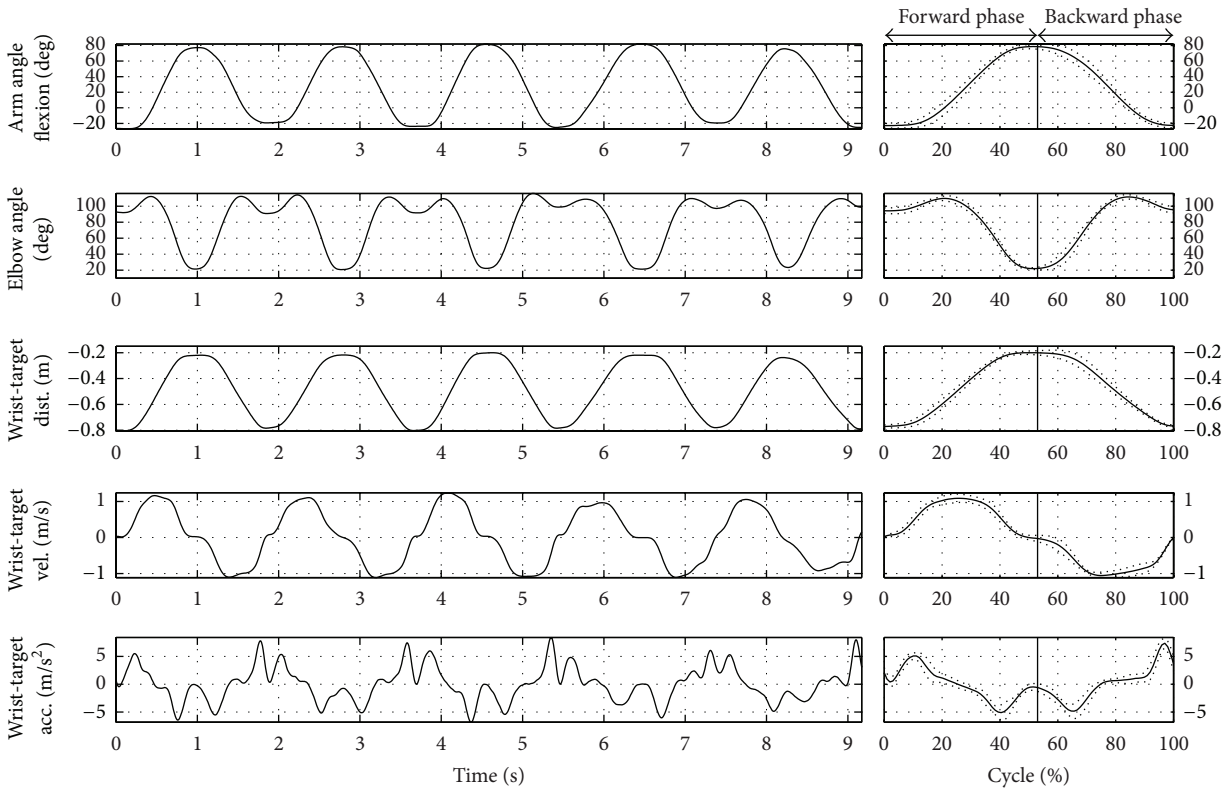

Figure 5: The reaching kinematics. Left-hand panel: wrist-target kinematics and joint angles trends relative to five RM repetitions. Right-hand panel: mean curves (normalized on the cycle percentage) and related standard deviations.

3.3. Normative Data. The voluntary group included 42 subjects (22 men, 3 left-handed) who were clustered by age into three subgroups: young (18–35 years old), middle-aged (36–50 years old), and elderly subjects (51–80 years old). Table 2 reports the composition of the whole group and subgroups. The dominant and nondominant arm reference normative data are presented in Table 3 where the I_R and I_M refer to the generic index I evaluated for the RM and HtMM, respectively.

No statistically significant difference in any variable was found among the three groups neither for the dominant nor for the nondominant arm. In fact, mean values are stable across groups and standard deviations are low. In addition, in none of the three groups, a statistically significant difference between dominant and nondominant arm was detected.

In Figures 7 and 8, the single subjects' NJ values for the RM and HtMM, respectively, are plotted against the movement time execution (MD) for both the dominant and nondominant arm. Plotted values represent the average results of the two trials performed by each subject; a linear regression curve relative to the entire group's data (right-hand side) and to the three subgroups' data (left-hand side) is also represented.

Figure 9 shows ACP data against MD of the whole group, that is, unclustered, since no difference in behaviour is observed among the three age subgroups.

3.4. Test Retest. An additional subgroup of 16 subjects (age 38 ± 15 years, 6 men, 2 left-handed) was reevaluated two weeks after the first testing session to test the method for repeatability.

Comparing the first and the second sessions, no statistically significant difference was found, neither for the dominant arm nor for the nondominant one. The test-retest reliability was, in fact, consistent, with r ranging between 0.66 and 0.90 ($1E-7 < P < 1E-3$) for almost all variables (Table 4). The angle variable AAF_R showed low correlation ($r < 0.55$, $P = 0.02$) in the case of the dominant arm. No correlation was found in HtMM repeatability (ACP_M) both for the dominant and nondominant arm ($0.20 < r < 0.26$, $P > 0.05$) and in RM repeatability (ACP_R) for the dominant arm ($r = 0.36$, $P > 0.05$).

3.5. Accuracy. The maximum possible error for the angles values at end movement (i.e., the angle values reported in Table 3) was, under unfavourable and pessimistic conditions, estimated to be 7 degrees for AAF_R and 6.5 degrees for EA_R in the RM and 4.8 degrees for EA_M in the HtMM.

The accuracy of ACP and NJ is very high as it depends only on the accuracy of the system as they are calculated from markers $M5$ and $M6$ coordinates.

4. Discussion

In this work, the Reaching and Hand-to-Mouth Evaluation Method and a set of normative data are presented. A discussion on the sensitivity, reliability, and applicability of

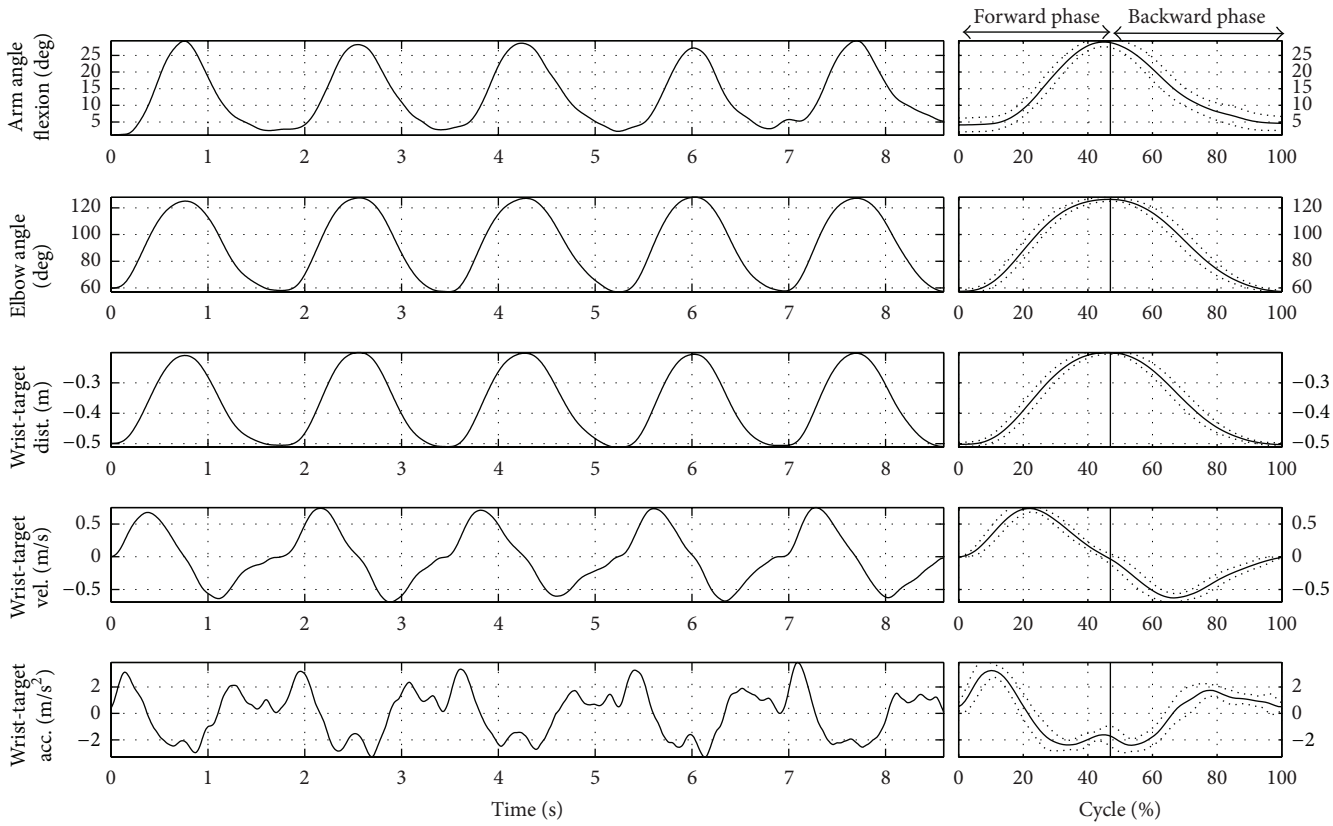

Figure 6: The hand-to-mouth kinematics. Left-hand panel: wrist-target kinematics and joint angles trends relative to five HtMM repetitions. Right-hand panel: mean curves (normalized on the cycle percentage) and related standard deviations.

Table 2: Healthy subjects and subgroups' composition.

	Healthy subjects (WG)	Young subgroup (YG)	Middle-aged subgroup (MG)	Old subgroup (OG)	Test-retest subgroup (—)
Number of subjects	42	13	14	15	16
Number of women	20	7	6	7	9
Number of left-handed	3	1	1	1	1
Age of women (years)	43 ± 14	27 ± 5	46 ± 4	59 ± 4	36 ± 13
Age of men (years)	44 ± 18	22 ± 3	43 ± 3	60 ± 9	41 ± 18
Age (years)	45 ± 16	24 ± 4	44 ± 4	60 ± 7	38 ± 15

Values are group means ± standard deviation.

the method is hereafter reported. Its limitations and future works will be also discussed.

4.1. Sensitivity. The sensitivity of the method was tested both by comparing dominant and nondominant arm results and by comparing data from the three groups of different ages. The discussion on sensitivity is based mainly on the evaluation of NJ over MD, ACP over MD.

Results show no statistically significant differences between dominant and nondominant arm performances. The authors infer that motor skills (i.e., capabilities) required by the examined tasks are not demanding enough to highlight differences in performance between the dominant and nondominant arm. Tasks requiring more sensory feedback, greater force production, and finer nervous control like in the use of the hand are probably better suited for this purpose [41, 42].

No statistical difference can be observed even between the average data from the three groups of different ages. Differences between groups' smoothness are highlighted when NJ is plotted against MD (Figures 7 and 8). In all groups, movement smoothness decreases (greater NJ) with increasing time of execution and, interestingly, the gradient of the regression curve differs among groups. Decreasing smoothness with decreasing movement-execution velocity was found also by Levy-Tzedek and colleagues [43] who studied the relation between speed and accuracy in a task requiring the comodulation of speed and position throughout

Table 3: Kinematic results.

		Dominant arm				Nondominant arm			
		Young group	Middle-aged group	Old group	Whole group	Young group	Middle-aged group	Old group	Whole group
HtMM	MD_M (s)	0.95 ± 0.17	0.92 ± 0.15	0.89 ± 0.13	0.92 ± 0.15	0.94 ± 0.16	0.90 ± 0.14	0.86 ± 0.15	0.90 ± 0.15
	EA_M (°)	136 ± 3	136 ± 5	135 ± 4	136 ± 4	136 ± 4	136 ± 4	135 ± 4	136 ± 4
	AVE_M (°/s)	59 ± 11	55 ± 11	63 ± 14	59 ± 12	58 ± 13	58 ± 12	67 ± 16	61 ± 14
	ACP_M (—)	0.93 ± 0.03	0.93 ± 0.04	0.92 ± 0.06	0.92 ± 0.05	0.91 ± 0.06	0.93 ± 0.04	0.92 ± 0.08	0.92 ± 0.06
	NJ_M (—)	19 ± 5	22 ± 7	22 ± 8	21 ± 7	19 ± 4	24 ± 8	24 ± 9	22 ± 7
RM	MD_R (s)	1.07 ± 0.16	1.03 ± 0.13	1.01 ± 0.12	1.04 ± 0.14	1.10 ± 0.21	1.03 ± 0.13	0.99 ± 0.11	1.04 ± 0.16
	AAF_R (°)	79 ± 6	80 ± 6	82 ± 5	80 ± 5	80 ± 7	80 ± 5	83 ± 3	81 ± 5
	EA_R (°)	26 ± 4	23 ± 5	25 ± 6	25 ± 6	27 ± 6	24 ± 7	23 ± 6	25 ± 6
	TAV_R (s^{-1})	1.02 ± 0.17	1.09 ± 0.17	1.07 ± 0.18	1.06 ± 0.17	1.01 ± 0.18	1.04 ± 0.16	1.09 ± 0.15	1.05 ± 0.16
	ACP_R (—)	0.93 ± 0.04	0.95 ± 0.02	0.95 ± 0.04	0.94 ± 0.03	0.94 ± 0.03	0.94 ± 0.03	0.94 ± 0.03	0.94 ± 0.03
	NJ_R (—)	25 ± 7	26 ± 7	25 ± 6	25 ± 6	27 ± 9	27 ± 5	25 ± 6	26 ± 7

Hand-to-mouth: MD_M: movement duration, EA_M: angle at elbow at end movement, AVE_M: mean angular velocity at elbow, ACP_M: acceleration coefficient of periodicity, and NJ_M: normalized jerk; reaching: MD_R: movement duration; AAF_R: angle of arm flexion at end movement, EA_R: angle at elbow at end movement, TAV_R: mean value of target approaching velocity, ACP_R: acceleration coefficient of periodicity, and NJ_R: normalized jerk.

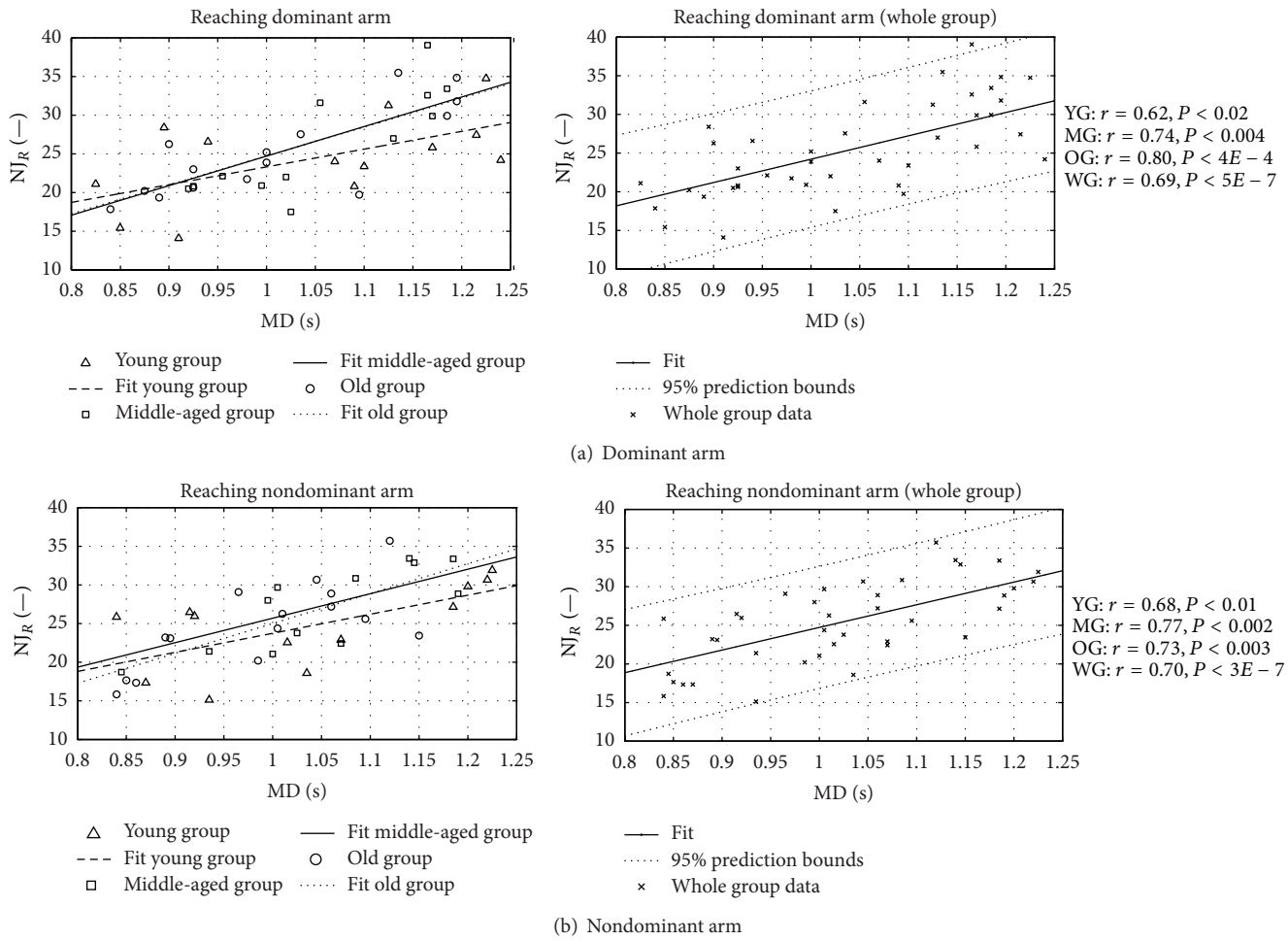

Figure 7: Reaching normalized jerk. Normalized jerk versus movement duration in RM. Pearson's correlation value and statistical significance level r and P are shown for young, middle-aged, old, and whole groups (YG, MG, OG, and WG, resp.).

Table 4: Test retest.

		Dominant arm				Nondominant arm			
		1st session	2nd session	r	P	1st session	2nd session	r	P
HtMM	MD$_M$ (s)	1.00 ± 0.14	0.93 ± 0.13	0.81	7E − 5	0.99 ± 0.12	0.92 ± 0.14	0.74	4E − 4
	EA$_M$ (°)	137 ± 4	136 ± 4	0.71	9E − 4	137 ± 3	136 ± 3	0.71	1E − 3
	AVE$_M$ (°/s)	58 ± 11	63 ± 13	0.79	1E − 4	57 ± 10	60 ± 14	0.79	1E − 4
	ACP$_M$ (—)	0.92 ± 0.03	0.92 ± 0.03	0.20	ns	0.91 ± 0.04	0.92 ± 0.03	0.26	ns
	NJ$_M$ (—)	23 ± 6	21 ± 5	0.75	4E − 4	24 ± 7	22 ± 6	0.82	6E − 5
RM	MD$_R$ (s)	1.10 ± 0.13	1.05 ± 0.18	0.73	7E − 4	1.13 ± 0.17	1.08 ± 0.17	0.90	8E − 7
	AAF$_R$ (°)	82 ± 6	80 ± 5	0.54	2E − 2	83 ± 5	82 ± 5	0.70	1E − 3
	EA$_R$ (°)	26 ± 4	27 ± 5	0.75	4E − 4	26 ± 6	27 ± 6	0.88	4E − 6
	TAV$_R$ (s^{-1})	0.99 ± 0.15	1.05 ± 0.18	0.85	1E − 5	0.97 ± 0.14	1.02 ± 0.16	0.75	3E − 4
	ACP$_R$ (—)	0.94 ± 0.03	0.94 ± 0.03	0.36	ns	0.93 ± 0.02	0.94 ± 0.03	0.66	2E − 3
	NJ$_R$ (—)	28 ± 6	26 ± 8	0.83	3E − 5	29 ± 8	27 ± 8	0.82	5E − 5

the task. Interestingly enough, considering that the smoothness is an indirect measure for the capacity for coordination [36, 40, 44, 45], the results reported in the present study indicate lower motor control ability in old subjects compared to the young ones, as expected, especially at low movement speeds (greater NJ). It is worth underlining that Pearson's product correlation coefficients are more than adequate and that even if the number of subjects per group is limited, the level of statistical significance for the correlation is high in all groups.

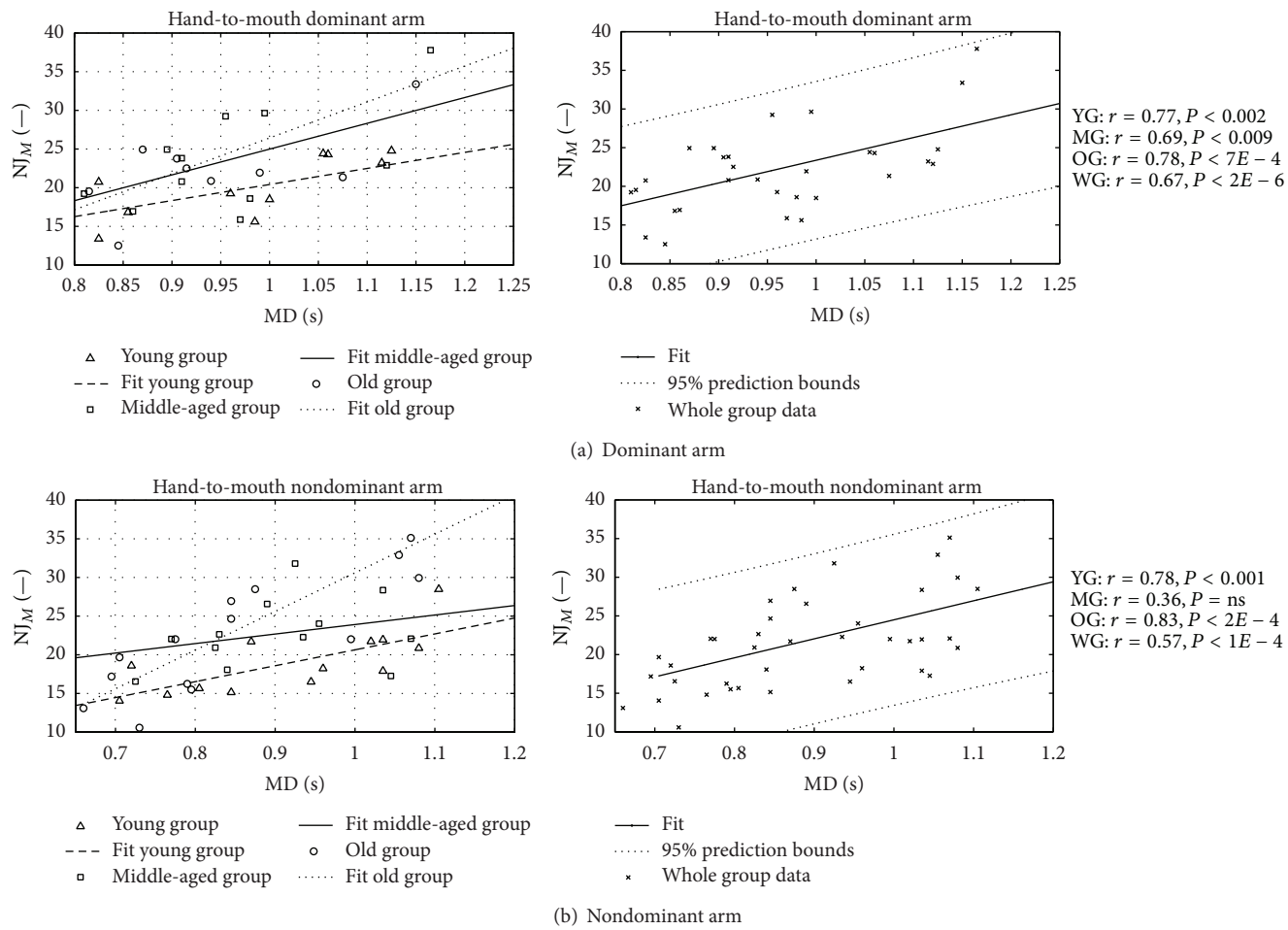

Figure 8: Hand-to-mouth normalized jerk. Normalized jerk versus movement duration in HtMM. Pearson's correlation value and statistical significance level r and P are shown for young, middle-aged, old, and whole groups (YG, MG, OG, and WG, resp.).

The ACP index, correspondingly, shows an overall reduced ability in performing repeatable movements at low velocity (see Figure 9) but, for ACP, no difference in trend was found among age groups. The experimental data confirm that NJ and ACP, both indirect measures of the subject's motor ability, are likely related to different mechanisms and levels of motor control: NJ, quantifying the smoothness of each single movement, is related to the capacity for interjoint coordination [44, 45]; ACP, measuring periodicity, is an assessment of the similarity of pattern production along single repeated movements. The ACP may thus be considered as a measure of consistency of the motion planning and control across repetitions.

4.2. Reliability. The test-retest reliability ranges from adequate to more than adequate for most variables ($0.54 < r < 0.90$ and $2E - 2 < P < 8E - 7$), and only three parameters (ACP_M dominant and nondominant arm and ACP_R dominant arm) show no correlation ($0.20 < r < 0.36$ and $P > 0.05$). Given the stability of the mean values across sessions and the very low intrasession standard deviations of the three variables ACP, NJ, and MD, the small size of the sample probably accounts for the low correlations observed. ACP, especially in the case of HtMM, is not enough sensitive to spot differences in the performance of single healthy subjects. This is not a limitation and ACP still may be considered as a valuable clinical measure because, by contrast, it has been demonstrated that it is enough sensitive to spot differences in repeatability even in high-functioning stroke patients [36].

4.3. Applicability. The strength of the method is in its simplicity that regards both the model and the protocol used. It is fast and user friendly, from the point of view of both the patient and the operator. Its applicability is wide and, considering the amount of information obtainable with a set of two movements only, it may be considered an important tool for routinely UL functional evaluation. The applicability of the method has been verified, in more than one decade, using it successfully in the evaluation of patients affected by different pathologies [34, 36, 46]. The authors did not have the opportunity to test the method on pathological children but, in their opinion, it has all the characteristics for being applied also in paediatric clinical practice. The evaluation of

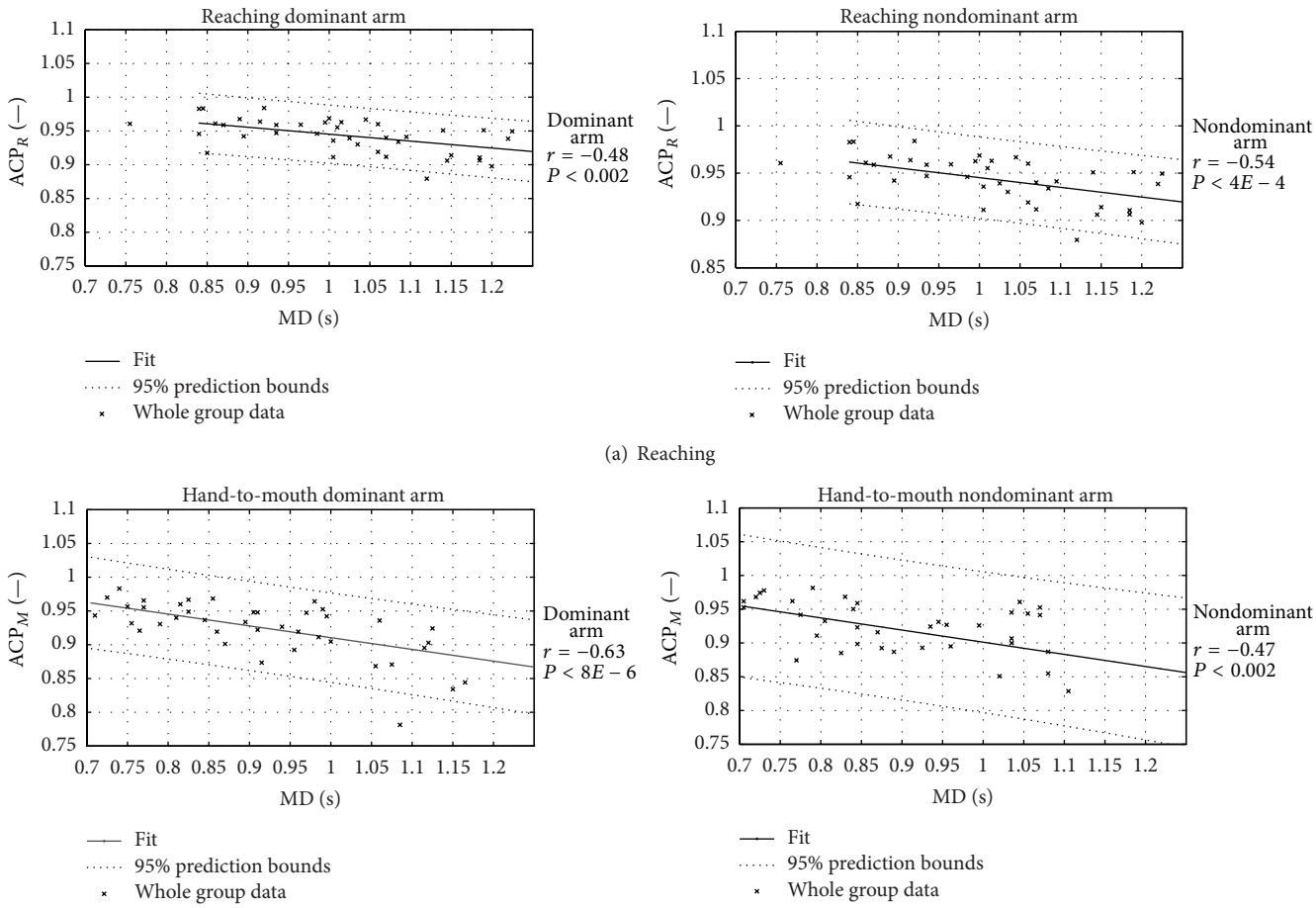

Figure 9: Acceleration coefficient of periodicity. Acceleration coefficient of periodicity versus movement duration in RM (upper panel (a)) and HtMM (lower panel (b)) is shown for dominant arm (left-hand panels) and nondominant arm (right-hand panels). Pearson's correlation value and level of statistical significance are indicated with r and P.

the RM has been tested on a group of healthy children and young boys aged 7 to 16 years: no problems in administering the protocol arose and the method appeared to be sufficiently sensitive to differentiate between the motor behaviours of subjects younger and older than 11 years [47].

4.4. Limitations and Future Works. The first limitation of the method lies in the model used which approximates the actual position of joint centres and, therefore, the evaluated articulation angles. To better estimate the joint centres position, the number of markers should be increased and some calibration procedure introduced [27, 48]. However, this would lead to greater complexity and longer procedural times that are not compatible with clinical practice. In other words, the overall applicability of the method could seriously be affected. For all these reasons, the authors decided to formulate an as simple as possible method and, concurrently, to estimate how the introduced approximations affect the overall accuracy. It is worth underlining that the assessment of the level of motor control is not affected by the introduced approximations as it is estimated using ACP and NJ that are calculated on the position of $M5$ and $M6$ only.

A second limitation is related to the average overall cost of a marker-based tracking acquisition system, typically affordable only by clinical and research centres specialized in movement analysis. The data-analysis protocol is however independent of a specific tracking system and other acquisition solutions, such as inertial-based sensors or off-the-shelf markerless tracking systems for consumer market (e.g., Microsoft Kinect), may be investigated. These new technological solutions would moreover possibly help in overcoming some limitations: they could face the tracking of the pronation and supination, and the hand pose, which, for the sake of simplicity, are neglected by the method illustrated in this work. Of course, reliability and sensitivity of such modified data acquisition setup should be investigated again and results should be compared with the control data presented with this work.

Finally, a limitation of the present work is that the dynamics is neglected. As a matter of fact, the kinematic

data acquired would be sufficient to estimate, under some assumptions and through inverse dynamics, joint powers and energetics of both the RM and the HtMM. First analyses of the shoulder torque and of the efficacy of a newly elaborated *effort index* have already been done on the RM with promising results [49, 50]. Once the efficacy of the dynamic model, together with dynamic parameters and the effort index, will finally be tested on patients, the data of the presented work will be enriched including dynamics.

5. Conclusions

This work led to the creation of a reliable database of normative data of the Reaching and Hand-to-Mouth Evaluation Method. Its simplicity and brevity make the whole procedure widely applicable to the UL functional assessment. The method appears to be an adequate tool to be used for routine UL functional assessment in clinics equipped with a movement analysis laboratory. Future works will focus on testing more affordable and user-friendly acquisition solutions in order to further extend the use of the method to smaller rehabilitation centres and clinical ambulatories.

Conflict of Interests

The authors declare that they have no competing interests.

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Research Article

Relative Contribution of Arms and Legs in 30 s Fully Tethered Front Crawl Swimming

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The relative contribution of arm stroke and leg kicking to maximal fully tethered front crawl swimming performance remains to be solved. Twenty-three national level young swimmers (12 male and 11 female) randomly performed 3 bouts of 30 s fully tethered swimming (using the whole body, only the arm stroke, and only the leg kicking). A load-cell system permitted the continuous measurement of the exerted forces, and swimming velocity was calculated from the time taken to complete a 50 m front crawl swim. As expected, with no restrictions swimmers were able to exert higher forces than that using only their arm stroke or leg kicking. Estimated relative contributions of arm stroke and leg kicking were 70.3% versus 29.7% for males and 66.6% versus 33.4% for females, with 15.6% and 13.1% force deficits, respectively. To obtain higher velocities, male swimmers are highly dependent on the maximum forces they can exert with the arm stroke ($r = 0.77$, $P < 0.01$), whereas female swimmers swimming velocity is more related to whole-body mean forces ($r = 0.81$, $P < 0.01$). The obtained results point that leg kicking plays an important role over short duration high intensity bouts and that the used methodology may be useful to identify strength and/or coordination flaws.

1. Introduction

The main goal in competitive swimming is to take the shortest time possible to complete an established distance. However, the factors that influence that performance may differ according to the distance to be swum. For instance, in long-distance competitive events swimming technique and race tactics are major factors for success [1]. On the other hand, in sprint events muscular power is crucial [2] since very high speeds are targeted. Therefore, in this latter category muscle force production must be very high to overcome the water resistance and drag [3, 4].

The required force that each swimmer has to apply may be exerted by arms and/or legs, and its assessment may be of great interest for training prescription. Although there have been substantial advances in technology available for swimmers evaluation, the direct measurement of force application during swimming performance remains difficult and challenging [5]. For instance, the use of force plates, usually referenced as the MAD-system, requires the swimmers not to user their legs [6], thus not being able to examine the relative contribution of leg kicking to free swimming. Keeping that in mind, it seems reasonable to use tethered swimming to assess the exerted forces in water, as it has proven to be a valid and reliable methodology [7, 8], with muscular activity [9], oxygen consumption [10], and stroke and physiological responses [11] similar to those seen in free swimming. In fact, tethering a swimmer to a load-cell allows the assessment of each individual force-time curve [12]. This individual evaluation may highlight the ability to effectively use muscular force production in the water, which is more important (and not necessarily related to) compared to assessing strength [13].

On the one hand, both arms and legs are considered the main contributors for force exertion in the water [14, 15], thus

being of major importance for performance enhancement over short distance events. Within these segments, their contribution seems to differ, with the leg kicking being commonly considered a factor of secondary importance for front crawl propulsion [16]. It has been stated that its participation enables the achievement of higher velocities by an average value of ~10% [17], but its contribution to overall swimming remains uncertain [14]. In fact, Swaine and colleagues [18] reported a higher importance of leg kicking than that previously assumed, highlighting its contribution to overall propulsion. Their experiments were carried out involving a new swimming training machine to examine the power produced by each limb, which brought new insights into this issue. However, experiments were performed in a dry-land situation being necessary to compare their data with results obtained by experiments carried out in the water.

On the other hand, it has been suggested that combining arms and legs with an appropriate coordination may generate a small amount of additional force exerted in water [18]. Indeed, since early 70s, that front crawl is known to involve highly skilled coordination of the leg kick and arm stroke to enhance utmost forward propulsion [19]. Nevertheless, empirical studies proving this concept are scarce. We do believe that the measurement of forces exerted in the water, restricting the swimmers to use only the arm stroke or the leg kicking, may provide new insights into this issue. Furthermore, it may discern dissimilarities according to gender, as the importance of force and/or technique may vary due to anthropometric characteristics of the swimmer [20]. Indeed, it is uncertain if gender differences in forces measured in tethered swimming during adolescence are similar to the ones described during adulthood [21]. It is known that throughout and up to the end of puberty, males become taller, heavier, and gain more muscle strength than females [22]. Hence, it might be useful to verify if these musculature gains lead to higher capability to effectively apply force in the water.

Research has failed to measure the capability of force exertion in the water using solely arm stroke or leg kicking. Thus, the main purpose of this study was to examine the relative contributions of arm stroke and leg kicking to fully tethered maximal swimming. Likewise, it was intended to examine possible differences in the relative contributions, according to gender, and how these contributions relate to swimming velocity.

2. Material and Methods

2.1. Subjects. The study included 23 young swimmers with a minimum of 5 years of experience in competitive swimming; main physical and performance characteristics of the subjects are described in Table 1, according to gender. All participants were sprint or middle-distance specialists, participating in national level competitions on a regular basis. Participants were informed about the purpose of the study and any known risks, and parents and coaches gave their consent for inclusion. All procedures followed the 1975 Declaration of Helsinki concerning human research and were approved by the Ethics Committee of the hosting research centre. None of the swimmers suffered from illness or from restrictions that hindered their performances during the experiments. Tests were performed during the competitive period of the spring training cycle to ensure that the participants were in a prime training period.

Table 1: Main physical and performance characteristics of the subjects, according to gender.

	Males ($n = 12$)	Females ($n = 11$)
Age (years)	15.2 ± 0.9	15.7 ± 1.4
Height (m)	1.73 ± 0.07*	1.61 ± 0.06
Upper limb length (cm)	64.8 ± 2.1†	58.1 ± 3.6
Lower limb length (cm)	82.5 ± 3.6†	75.8 ± 4.4
Body mass (kg)	61.8 ± 7.1†	55.7 ± 5.8
Body fat (%)	11.7 ± 3.1†	23.7 ± 3.6
Personal best 100 m freestyle (s)	59.5 ± 2.0†	67.1 ± 5.9

Values are mean ± SD; ∗ and † significantly higher than females ($P < 0.05$ and $P < 0.01$, resp.).

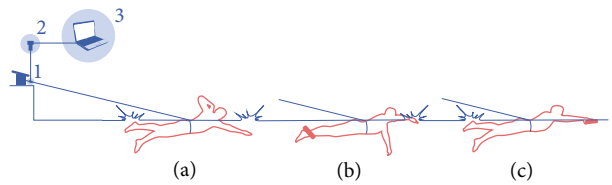

Figure 1: Apparatus used for the fully tethered whole body (a), arm stroke (b), and leg kicking (c) swimming tests: 1 = load-cell; 2 = ergometer data acquisition system; 3 = personal computer.

2.2. Experimental Design. The swimmers completed a 1000 m standardized warm up (400 m swim, 100 m pull, 100 m kick, 4 × 50 m at increasing speed, and 200 m easy swim) before performing 3 × 30 s maximal intensity fully tethered front crawl swimming. The tests were performed in a randomized order and were separated by a minimum of 20 min active recovery. For 1 test, no constrains were applied so that participants could use their whole body (cf. Figure 1(a)). For the other 2 tests, floating devices (pull buoys) were used to restrict the movements of legs or arms (cf. Figures 1(b) and 1(c), resp.). Visual inspection by the scientific personnel was made to verify if the swimmers were able to keep their streamlines for each condition; if not, the test was repeated. In addition, the participants had their ankles fastened together to prevent them from performing leg kicking when assessing forces for the arm stroke test (Figure 1(b)). On the subsequent day, swimmers performed 2 maximal 50 m front crawl swimming bouts with an in-water start, to diminish the effect of starts and glide. All experimental testing was performed in a 50 m indoor swimming pool with a water temperature range of 25.5–27°C.

2.3. Experimental Measurements. A 3.5 m inextensible steel cable was used to attach the participants to the starting

block. Between the cable and the starting block a load-cell was used to measure forces exerted by the swimmers. Equipment provided a recording rate of 100 Hz and a maximum capacity of 4905 N. The load-cell was connected to a Globus Ergometer data acquisition system (Globus, Italy), which exported the data in ASCII format to a PC. Preceding the starting signal, swimmers adopted a horizontal position with the cable fully extended and performed for 5 s at low intensity. Data collection only started after the completion of the first maximum cycle to avoid the inertial effect of the cable extension, usually observed during the first upper limb action. The participants were told to follow the breathing pattern they would normally apply during a 50 m freestyle event and were verbally encouraged throughout the tests to maintain maximal effort during the 30 s. The end of the tests was set through an acoustic signal emitted by the scientific personnel.

2.4. Data Analysis. ASCII data files were exported to signal processing software (AcqKnowledge v.3.7, Biopac Systems, Santa Barbara, USA) and filtered through a 4.4–4.8-Hz cut-off low-pass fourth-order Butterworth filter. The cut-off value was chosen according to residual analysis (residual error versus cut-off frequency). As the force vector in the tethered system presented a small angle (5.7°), data was corrected computing the horizontal component of force. Individual force-time curves were assessed to estimate maximum and mean values. Swimming velocity was calculated from the 50 m free swim best time.

2.5. Statistical Analysis. Normality and homoscedasticity assumptions were checked by Shapiro-Wilk and Levene tests, respectively. The significance of differences between genders was evaluated with an independent samples t-test. Repeated measures analyses (ANOVA, with Bonferroni post hoc test) and Pearson product-moment correlation coefficients (r) were calculated among the tests, controlling the swimmers gender. All statistical procedures were performed using SPSS 20.0 (Chicago, IL, USA). The level of statistical significance was set at 95% ($P < 0.05$).

3. Results

Illustrative force-time curves for the 3 tests are shown in Figure 2. Different profiles can be perceived as the restriction of legs or arms affected swimmers capability to exert force in the water. When the swimmer increases the force exertion, an upward trace arises and, on the other hand, when the force exertion decreases a downward trace occurs.

In Table 2, the maximum and mean forces obtained for each condition are displayed, according to gender. With no restrictions, swimmers were able to exert higher forces than using only their arm stroke or leg kicking ($P < 0.001$). The sum of arm stroke and leg kicking mean forces was higher than the whole-body mean force for all subjects, except for two female swimmers. Estimated relative contributions of arm stroke and leg kicking were 70.3% versus 29.7% for males and 66.6% versus 33.4% for females. Considering the sum of arm stroke and leg kicking mean forces as 100%, the forces exerted using the whole body were 84.4 ± 6.8% for males and 86.9 ± 9.9% for females. This corresponded to a force deficit of 15.6% and 13.1%, respectively (cf. Figure 3).

In the males group, the ones that obtained higher maximum forces were not the ones that produced higher mean forces ($r = 0.46$, $P > 0.05$), whereas in the females group there was a significant association between the maximum and mean forces ($r = 0.76$, $P < 0.01$). For both genders, forces exerted using the whole body were positively related to the forces obtained through arm stroke ($r = 0.7$, $P < 0.05$) and leg kicking ($r = 0.8$, $P < 0.05$).

Table 2: Data collected from the 30 s fully tethered front crawl swimming tests, according to gender.

	Males ($n = 12$)	Females ($n = 11$)
Maximum force (N)		
Whole body	325.4 ± 27.8†	222.3 ± 61.8
Arm stroke	243.7 ± 27.7†	168.5 ± 36.2
Leg kicking	100.1 ± 28.2†	72.0 ± 9.4
Mean force (N)		
Whole body	98.8 ± 13.7†	74.0 ± 12.4
Arm stroke	82.5 ± 12.0†	56.9 ± 8.7
Leg kicking	35.1 ± 7.6†	28.4 ± 4.6

Values are mean ± SD; †significantly higher than the females ($P < 0.01$).

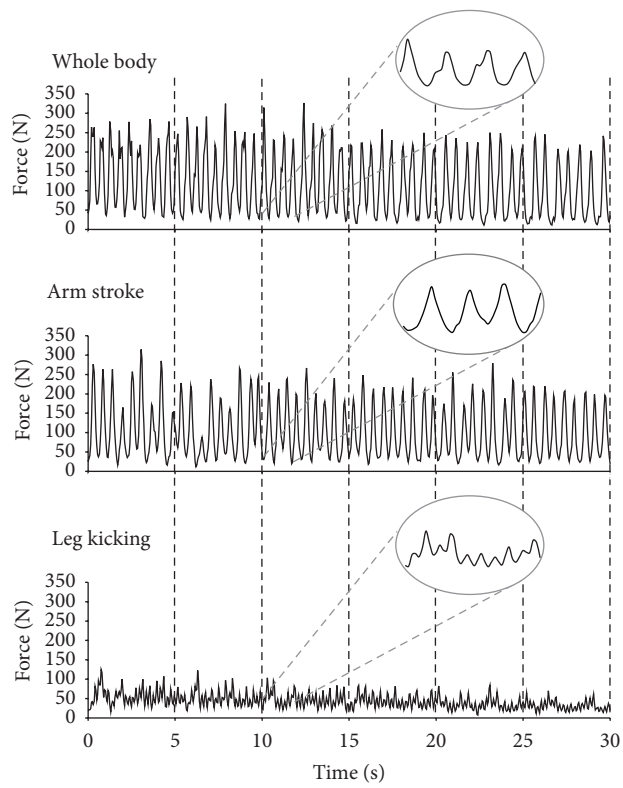

Figure 2: Illustrative typical force profiles in a 30 s tethered whole body, arm stroke, and leg kicking swimming tests.

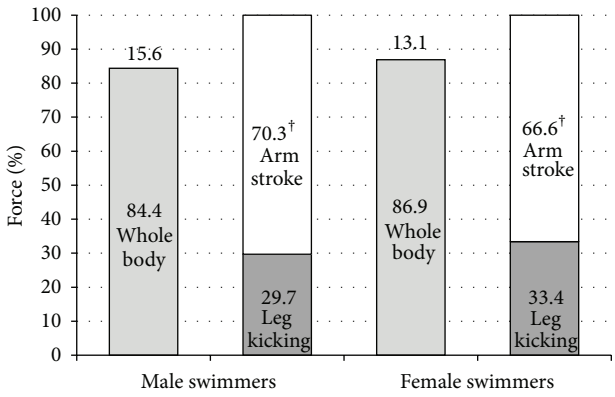

Figure 3: Relative contribution (%) of arms and legs in fully tethered front crawl swimming, according to gender. $^{\dagger}P < 0.01$ compared with the whole-body conditions. Numbers in the columns represent mean values.

Male swimmers performed higher swimming velocities than their female counterparts (1.71 ± 0.05 versus 1.53 ± 0.11 m/s, $P < 0.01$). The velocities presented higher correlations with the arm stroke maximum force for males ($r = 0.77$, $P < 0.01$) and with the whole-body mean force for females ($r = 0.81$, $P < 0.01$).

4. Discussion

To the best of our knowledge, few studies have aimed to analyze the importance of leg kicking in front crawl swimming [16, 17]. Thus, this study aimed to examine the relative contribution of arm stroke and leg kicking to force production in front crawl fully tethered swimming. The major findings of this study were that (i) leg kicking played a high relative contribution (~31%) and (ii) male swimmers achieved higher swimmer velocities due to the high force exertion with their arm stroke, whereas female swimmers swimming velocities were more related to whole-body mean forces. Additionally, the obtained results permitted the estimation of a force deficit, which may be a useful procedure to identify lack of strength and/or coordination.

For all swimmers tested, maximum and mean forces were higher using their whole body, followed by using only the arm stroke and followed by the leg kicking. Even though these are pioneering results, they are in accordance with the expectations, as propulsive capacity decreases in each situation. Previous studies pointed out that the arm stroke generates 90% of the total propulsive thrust in sprint freestyle [16, 17]. Using the 20 m swimming speeds with no constraints and using only the arm stroke, Deschodt and colleagues [17] indirectly reported that 10% higher speed was achieved when using the leg kicking. These authors did not normalize the data to a leg kicking condition and yet their percentage was lower than the one obtained in the present study, with lower level swimmers. On the other hand, a recent study [18] reported a 62.7 ± 5.1% and 37.3 ± 4.1% mean contribution of arms and legs, respectively, using a dry-land novel machine. Current higher percentages undoubtedly put in question the statements that swimming propulsion is almost entirely due to arms and trunk [23], at least over short duration and high intensity bouts. A relative contribution of ~31% (cf. Figure 3) may reinforce the suggestion that a much greater proportion of the force exerted in water to increase swimming speed may be attributable to the legs than previously thought [14]. The small bias between the experiment of Swaine and colleagues [18] and the present study may be due to the environment conditions, as the former was performed in dry land.

In front crawl swimming, the upper and lower limbs perform alternated movements to produce propulsive actions. Except for 2 female swimmers, the sum of forces exerted by arm stroke and leg kicking was higher than using the whole body. These results suggest that combining upper and lower limbs with an appropriate coordination may generate a small amount of additional force exerted in water. In doing so, a powerful leg kick may be almost as important as a powerful arm stroke in swimming, even though the leg kick contributes much less to propulsion [14]. The used methodology allowed identifying swimmers that were not able to achieve as much force with no constrains, as with the sum of arms and legs separately. These differences can be considered as a force deficit (which was similar among genders), providing the diagnosis of coordination flaws. In fact, low values may represent situations where strength development of arms and legs might not lead to a gain in performance, as the necessary coordination would be deficient [17]. Interestingly, Ogita et al. [24] reported that the total energy production during swimming was lower than the sum of energy production during separately measured arm stroke and leg kicking swimming. It seems, therefore, that the potentials of both the anaerobic and aerobic energy releasing processes in the muscle groups involved in arm and leg actions cannot be fully reached during free swimming [24].

Regarding the gender comparison, male swimmers obtained both higher forces and velocity than female swimmers. These gender biases may be related to anthropometrical differences (cf. Table 1), but similarities between relative contribution and force deficit should be examined. First, in order to avoid errors due to the stationary swimming, current data was obtained on competitive swimmers already familiarized with fully tethered swimming methodology, as recommended [8]. Second, it is well described in the literature that several factors influence swimming performance. For instance, both head [25] and thumb [26] positioning have proven to be influencing factors. To overcome the limitation of not controlling every variable, we assumed that swimmers would maintain similar positioning among tests. Third, apart from being taller, heavier, and with longer limbs, as is common in postpubescent stages [27, 28], the higher forces that male swimmers obtained may induce that they had higher muscle strength levels than females, which is in accordance with previous findings that point to a strength differential after puberty [28, 29]. Furthermore, our results suggest that those swimmers with higher strength levels are also those with higher swimming speed, being partially related to a greater capacity to apply propulsive force to water. Nevertheless, the relative contributions of arm stroke

and leg kicking were parallel among genders. Thus, the used procedures can be a novel approach to assess the forces exerted in each condition, in order to evaluate the relative contributions of arm stroke and leg kicking.

Relationships between front crawl tethered swimming and swimming performance have been previously studied [30–32], mostly with heterogeneous samples indifferently analyzing male and female swimmers [12, 16]. Studies showed that the stroke force that swimmers could generate was moderately to highly related to swimming speed in sprint distance efforts. However, coupling results of heterogeneous samples can discredit the results [33], and the analyses per gender might clarify performance differences in adolescent swimmers. Indeed, proper scrutiny is to be recommended, since associations of maximum or mean forces differ. It was confirmed that maximum force is a better estimator for swimming performance in male swimmers, while mean values are more appropriate for female swimmers. Differences were also noticeable for arm stroke and leg kicking experiments. In fact, the musculature of the upper body seems highly correlated with sprint performance in male swimmers. On the contrary, in female participants, whole body plays a major role over short-distance swimming performance. As remarked earlier, differences in musculature and strength become notorious at these ages [27] and should be considered when prescribing strength training.

5. Conclusions

Swimming coaches are aware that the evaluation of their swimmers should be specific to the nature of the sport. Therefore, it is essential that the chosen apparatus strongly replicates the movement patterns (if possible with the same musculature demands) employed in real training and competition situations. In summary, both arm stroke and leg kicking were a two-independent factor that took a major role in sprint distances, so that leg kicking should not be neglected in sprint swimmers training. In male swimmers, upper limbs musculature was able to reach very high values of exerted forces strongly influencing swimming performance. For female swimmers, the ability to keep force production during the 30 s, that is, mean forces, was more related to swimming performance. The used methodology may be useful to diagnose the lack of strength or coordination.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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