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Assessment of upper-body dynamic stability during walking in patients with subacute stroke

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Abstract—The analysis of upper-body acceleration is a promising and simple technique to quantitatively assess dynamic gait stability. However, this method has rarely been used for people with stroke, probably because of some technical issues still not addressed. We evaluated the root-mean-square (RMS) and harmonic ratio of trunk accelerations for a group of 15 inpatients with subacute stroke who were able to walk (61.4 +/– 14.9 yr) and compared them with those of an age-matched group of nondisabled subjects (65.1 +/– 8.8 yr) and those of a highly functional group of young nondisabled subjects (29.0 +/– 5.0 yr). Small (<2%) but significant ($p < 0.03$) differences were found in RMS values obtained by applying the two most common computational approaches: (1) averaging among individual-stride RMS values and (2) computing the RMS value over the entire walking trial without stride partitioning. We found that the intersubject dependency of acceleration RMS values by selected walking speed was specific for each group and for each of the three body axes. The analysis of ratios between these three accelerations provided informative outcomes correlated with clinical scores and not affected by walking speed. Our findings are an important step toward transferring accelerometry from human movement analysis laboratories to clinical settings.

Key words: accelerometry, ambulation, biomechanics, dynamic balance, gait, mobility, movement analysis, rehabilitation, stroke, walking speed.

INTRODUCTION

In recent years, interest has been increasing in the use of accelerometry for quantifying movement patterns during walking in a suitable and simple manner [1]. In fact, the ability of a subject to maintain balance during walking can be properly assessed by measuring upper-body accelerations, because two-thirds of the body's weight is located at two-thirds of the body's height above the ground [2–3]. A large and growing body of literature has investigated the use of this technique in nondisabled subjects [4–7].

Abbreviations: ANOVA = analysis of variance, AP = anteroposterior, BI = Barthel Index, CC = craniocaudal, FAC = Functional Ambulatory Classification, HFG = high functional group, HR = harmonic ratio, ICC = intraclass correlation coefficient, LFG = low functional group, LL = laterolateral, MFG = medium functional group, RMI = Rivermead Mobility Index, RMS = root-mean-square, RMS_{AP} = RMS value along AP axis, RMS_{CC} = RMS value along CC axis, RMS_{LL} = RMS value along LL axis, RMS_{mean} = RMS value obtained averaging individual-stride RMS values, $RMS_{overall}$ = RMS value obtained over the entire walking trial without stride partitioning, WS = walking speed.

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Some of the potential benefits of using accelerometers to assess movement in clinical settings could include the low cost compared with more commonly used gait analysis equipment, the small dimensions and light weight (which enable subjects to walk relatively unrestricted), and no limitation of the testing environment to a laboratory [1].

Despite these possible advantages, body accelerations have been investigated in only a few samples of patients with low-back pain [8], cognitive impairments [9], or dystrophy [10]. Accelerometric signals have also been recently analyzed in patients with stroke to assess their dynamic stability during robotic gait training [11] and level walking [12]. Nevertheless, use of accelerometry in clinical settings to assess gait dynamic balance in patients with subacute stroke is still rare. This is probably because the parameters extracted by accelerometric signals still need to be suitably adapted to a clinical population, in spite of previous studies on patients with stroke [11–12]. Two of the main limiting factors might be related to the computation and normalization of these parameters, as detailed here. Notwithstanding, accelerometry can be a suitable tool for quantifying the dynamic stability and smoothness of upper-body walking patterns (reflecting motor recovery and gait capability) and, hence, for assessing walking rehabilitation outcomes in severely affected patients [12]. In fact, after stroke, balance and gait abnormalities are frequent [13]. These instabilities also relate to higher risk of falls, with consequent high economic health policy implications [14–15]. Wearable devices containing accelerometers may offer informative data about basic gait parameters and may be easily used when laboratories for human movement analysis, which contain stereophotogrammetric systems and force platforms, are not available because of limitations in space, time, and funding, as is common in daily clinical routine [16]. Furthermore, the quantitative assessment of upper-body dynamic stability during walking may help elucidate a more complete clinical picture of motor dysfunctions and compensation strategies, allowing clinicians to identify potential fall risks and detail a proper rehabilitative pathway. In fact, the measurement of upper-body accelerations can provide an informative quantification of gait dynamic stability [17].

The root-mean-square (RMS) of the acceleration signal, an indicator of acceleration dispersion, is the most commonly used parameter for assessing upper-body dynamic stability during walking [3–5,9–10,12,17]. To the best of our knowledge, two different methods have been used in previous studies to compute RMS values: (1) a mean RMS value obtained averaging individual-stride RMS values (RMS_{mean}) [3–4,10] and (2) an overall RMS value obtained over the entire walking trial without stride partitioning ($RMS_{overall}$) [5,12,17–18] (**Figure 1**). The first approach is probably the most suitable when the signal can be easily partitioned into single strides. The second one is easier to compute when signal stride partitioning is difficult, such as in severely affected patients. Despite the fact that the two approaches are not expected to be equal because of the nonlinear RMS formula (in fact, the presence of roots and squares in this formula implies that the general mean can be different from the mean of single means, see "Methods" section), far too little attention has been paid to the comparability of the results obtained with these two methods and, hence, to the usefulness of generalization.

Furthermore, acceleration RMS values are affected by walking speed (WS), because of the kinematic link between acceleration and velocity. Acceleration is the rate of change of velocity over time. It implies that an increase in upper-body acceleration could be attributed to an unsteady speed due to pathological instabilities as well as to an increase in WS. So, since increasing or decreasing gait velocity results in a corresponding increase or decrease in acceleration amplitude [1], RMS values need to be normalized between subjects and populations walking at different WSs to suitably assess only the upper-body dynamic instabilities imputable to a physical impairment [12]. Previous studies, conducted with subjects walking at different speeds, highlighted a quadratic relationship between trunk acceleration RMS and WS [17,19]. However, no reliable evidence exists that this intrasubject association can be generalized between subjects and, least of all, among populations walking at different speeds. In fact, this association between velocity and acceleration found in nondisabled subjects along all three body axes [4,17] can be altered in impaired conditions [20]. The measurement of these alterations can provide informative data about reductions of gait dynamic stability [12]. Because WS affects all three RMS values, with craniocaudal (CC) accelerations usually being less informative than anteroposterior (AP) and laterolateral (LL) ones [7,9,18], it might be possible to normalize AP and LL RMS accelerations by dividing them by the CC accelerations.

The aim of this study was to quantitatively assess gait dynamic stability by comparing upper-body accelerations of patients with subacute stroke to those of nondisabled subjects, addressing the still unsolved main issues: the

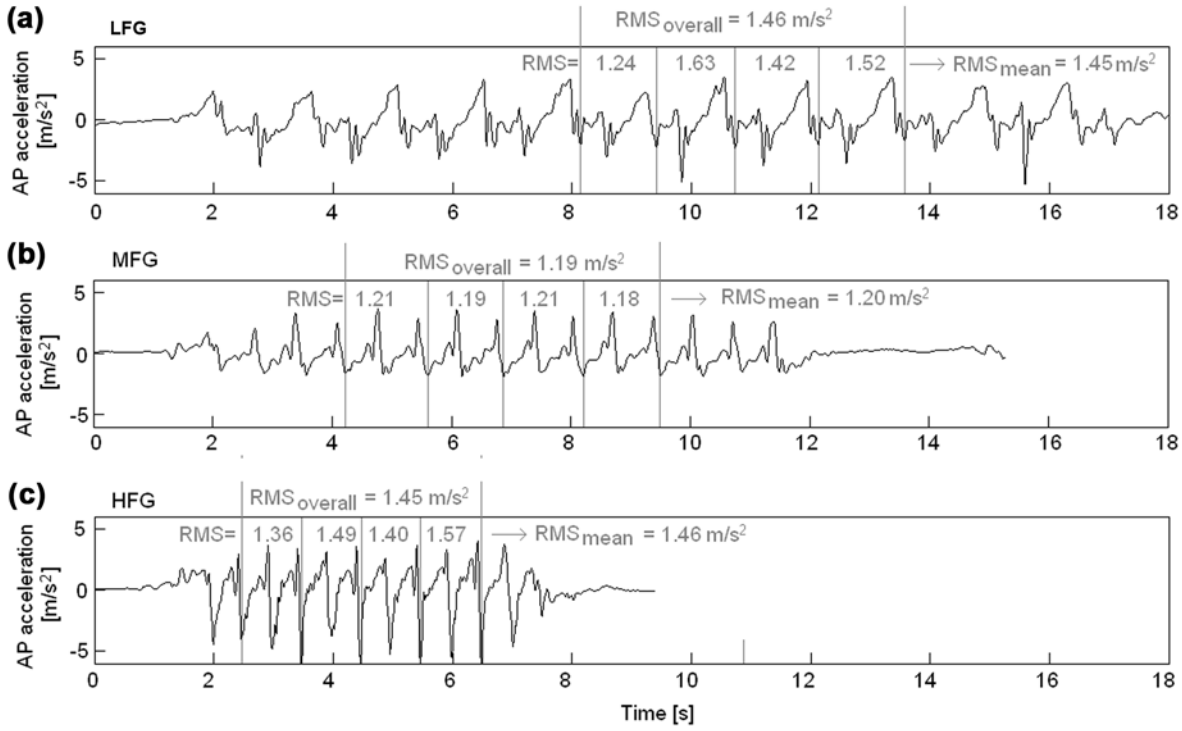

Figure 1.
Root-mean-square (RMS) computation: anteroposterior (AP) acceleration signals of three subjects, one for each group: **(a)** subject (walking with cane) from low functional group (LFG); **(b)** subject from medium functional group (MFG); **(c)** subject from high functional group (HFG). RMS values computed on four central strides identified by gray vertical lines. RMS of AP acceleration reported for each stride. RMS_{mean} = RMS value obtained averaging individual-stride RMS values, $RMS_{overall}$ = RMS value obtained over entire walking trial without stride partitioning.

methodological one, related to the RMS computation, and the biomechanical one, related to RMS normalization. Three groups of subjects with different functional levels were enrolled: nondisabled young, nondisabled elderly, and subjects with hemiparesis due to subacute stroke. The overall aim of the study was to test the hypothesis that upper-body dynamic stability can be accurately assessed by measuring and using a novel evaluation of acceleration RMS in patients with stroke.

METHODS

Participants

Three groups of 15 subjects each were enrolled in this study. The first group was composed of 15 inpatients with hemiparesis due to stroke (61.4 ± 14.9 yr, 6 women, 93.3 ± 39.3 d from stroke event), who were expected to represent a low functional group (LFG). The second was an age-matched group (p = 0.42, t-test) of 15 nondisabled adults (65.1 ± 8.8 yr, 5 women). Since previous studies revealed compromised gait dynamic balance in an elderly population [4–5], this group was expected to represent a medium functional group (MFG) in terms of gait dynamic stability. The third group was composed of 15 nondisabled young adults (29.0 ± 5.0 yr, 8 women), who were expected to represent a high functional group (HFG).

The mobility of patients was clinically assessed by means of the Rivermead Mobility Index (RMI) (mean RMI = 10.7 ± 3.4), a scale measuring mobility of subjects with stroke in relationship to many aspects of their static and dynamic balance [21]. Furthermore, the Barthel Index (BI) (mean BI = 74.7 ± 24.0) and Functional Ambulation Classification (FAC) (mean FAC = 4 ± 1) were also assessed for these patients. The BI is the most commonly used scale to assess the degree of independence of a

patient in various activities of daily living (including mobility and transfers). The FAC allows easy classification of patients with respect to their walking ability. Nobody in the LFG had maximum scores in all these three clinical scales. This determined their inclusion in the LFG. This study included inpatients with subacute stroke able to walk independently (FAC > 3 with or without the need of a cane) and also inpatients able to walk only under a physiotherapist's supervision (where slight contact might have been required at times, FAC ≤ 3). Conversely, this study excluded, after previous pilot tests, patients unable to walk autonomously and needing an influential external body-weight support during gait, provided by a therapist or a walker, and patients with lower-limb muscle clonus during walking (**Figure 2**, more details in "Results" section).

Protocol

Subjects were asked to stand still on a line marked on the floor and then to walk straight for 10 m at their self-selected speed (10 m walking test [22]) until arriving at another line on the floor in a 30 m-long rehabilitation gymnasium. Subjects wore an elastic belt with an inertial sensor device (FreeSense®, Sensorize s.r.l.; Rome, Italy; sampling frequency = 100 Hz) located on their back, corresponding to the second and third lumbar spinous processes, close to the body center of mass. This device is lightweight (93 g) and contains a triaxial accelerometer to measure accelerations along the three body axes (AP, LL, and CC). All subjects wore their commonly used shoes during the tests. In the LFG group, five patients performed the test autonomously without any aid, five patients performed the test autonomously but using a cane, and the other five patients performed the test under a physiotherapist's supervision (two of them also used a cane).

Analysis of Acceleration Signals

We analyzed the acceleration data recorded during four consecutive strides (eight steps) in the central part of the walking pathway. Each negative peak of AP accelerations was used to identify the beginning of a step (foot contact with the ground, **Figure 1**) [3]. We analyzed signals after subtracting their mean values and low-pass filtering them at 20 Hz [4]; then we computed the acceleration RMS for each body axis. This parameter is a measure of acceleration dispersion, and it coincides with the standard deviation of the accelerometric data because of signal mean subtraction. It is the most used parameter for assessing gait stability on the basis of upper-body accelerations, and it is computed with acceleration signals (a_i with $i = 1 \ldots n$) as follows [1,6]:

$$\mathrm{RMS} = \sqrt{\frac{\sum_{i}^{n} a_i^2}{n}}$$

We also computed the harmonic ratio (HR), a measure of the smoothness of movement and gait rhythmicity [1,4]. The HR is based on the premise that the unit of measurement from a continuous walking trial is a stride (two steps), and it is computed as the ratio between the sum of even/odd for AP and CC or odd/even for LL accelerometric harmonics amplitudes calculated via discrete Fourier transformation as follows [1,6]: HR = amplitudes of even harmonics/amplitudes of odd harmonics along AP and CC axes, and HR = amplitudes of odd harmonics/amplitudes of even harmonics along LL axis.

Because of its strict relationship to stride partitioning, HR is usually computed only by averaging among individual-stride HR values, analogous to $\mathrm{RMS}_{\mathrm{mean}}$. For this reason, we did not compute an HR over the entire walking trial, but we did compute this parameter over individual strides to provide the most complete clinical picture of our patients.

Methodological Issue: Mean and Overall Acceleration

The RMS of acceleration is the parameter most commonly used to summarize upper-body accelerations. We applied the two methods most commonly used to compute RMS: (1) $\mathrm{RMS}_{\mathrm{mean}}$ was computed by averaging four RMS values related to the four analyzed strides [3–4,10], and (2) $\mathrm{RMS}_{\mathrm{overall}}$ was computed over the four analyzed strides without single-stride partitioning [5,12,17–18]. The mathematical results of these two computations are not equal because of the nonlinear formula used to compute the RMS (as previously reported). This nonlinearity implies the loss of the properties of homogeneity and translativity typical of linear arithmetic, in which a mean of means coincides with the overall mean. Hence, the results of the two methods are expected to be mathematically different.

Figure 1 shows three representative cases, one for each group. Statistical significance of the differences between $\mathrm{RMS}_{\mathrm{mean}}$ and $\mathrm{RMS}_{\mathrm{overall}}$ values was analyzed in the three groups of subjects.

Biomechanical Issue: Acceleration and Speed

We then attempted to address another critical issue. Since the results of previous studies obtained with subjects asked to walk at different WSs showed a quadratic relationship between trunk acceleration RMS and WS [17,19], we divided the RMS by the square of the WS and multiplied it by the average step length to obtain a dimensionless normalized value, hypothesized to not be affected by WS [12]. This approach suffers from a serious weakness because the intrasubject correlation between the above-mentioned parameters was arbitrarily extended between subjects and even between different populations (patients and nondisabled subjects), both walking at their own self-selected speeds. Furthermore, it has also been shown that locomotion can be altered if subjects are asked to walk slower or faster than their self-selected speed [23]. Furthermore, the way WS influences upper-body movements is still poorly understood [19], and their relationship can be altered in certain disabling conditions because of physical or environmental factors [20]. Hence, we analyzed the intersubject relationships between WS and RMS in each one of the three groups of subjects expected to have different levels of gait functioning. Since WS can affect all the RMS values assessed along one of the three body axes, we also evaluated the relationship among these three RMS values, computing the two intrasubject ratios: RMS value along AP axis (RMS_{AP})/RMS value along CC axis (RMS_{CC}) and RMS value along LL axis (RMS_{LL})/RMS_{CC}.

Statistical Data Analysis

The following analyses were performed to test the hypothesis that upper-body dynamic stability can be accurately assessed by measuring and using a novel evaluation of acceleration RMS in patients with stroke. The RMS_{mean} and $RMS_{overall}$ values were compared using a paired *t*-test for each group of subjects. Test-retest reliability of the RMS_{mean} among values related to successive strides for each group of subjects and along each body axis were assessed using the two-way random intraclass correlation coefficient (ICC(2,1)). The successive analyses were performed only on the values of RMS_{mean}.

A repeated-measures analysis of variance (ANOVA) was performed, with functionality as a factor between subjects (HFG, MFG, LFG) and axis as a factor within subjects (AP, LL, CC) to assess differences in terms of RMS and HR. A one-way ANOVA was performed to assess the differences in terms of WS among groups. All the ANOVA calculations were followed by post hoc analyses performed with *t*-tests with Bonferroni correction.

Linear and quadratic regressions were performed to assess the relationships between RMS and WS. The adjusted coefficient of determination (R^2) was computed to assess the quality of these regression fits, taking into account the difference between the number of parameters of linear and quadratic fitting equations. The correlations were assessed by means of Pearson's correlation coefficient (R) when two continuous variables were compared and by means of Spearman's correlation coefficient (ρ) when a continuous variable was compared with a clinical score. From a clinical point of view, our aim was to test the hypothesis that dynamic gait stability can be accurately assessed by measuring acceleration RMS in patients with stroke. In order to test this hypothesis, we evaluated (1) whether this measure highlighted differences between patient and control groups, (2) whether it was reliable within each group, and (3) whether it provided results that correlated with commonly used clinical scores. The relationship among clinical scores and the ratios RMS_{AP}/RMS_{CC} and RMS_{LL}/RMS_{CC} were also investigated. Significance level was set at 0.05 (for 15 subjects this implies $R > 0.514$ [$R^2 > 0.264$] to achieve significant level).

RESULTS

Participants and Relevant Acceleration Signal Profiles

Figure 1 shows the typical AP acceleration signal recorded for one representative subject from each group. Both nondisabled subjects (young and elderly) showed similar patterns inside a stride; i.e., the acceleration patterns were repeated twice (and/or in multiples of two) within any stride because of the similarities between two consecutive steps. Furthermore, their RMS values were repeatable among strides. Conversely, the patient with stroke showed two different patterns inside the same stride because of functional asymmetries due to hemiparesis. As expected, subjects with a higher functional level required less time to complete the test than those with a lower functional level.

By way of example, **Figure 2** shows the signals of three severely affected patients. The first one (**Figure 2(a)**) walked with a cane and with the hand of the therapist on his back (if needed). His acceleration signal profile was clearly different from that of the patient and the nondisabled

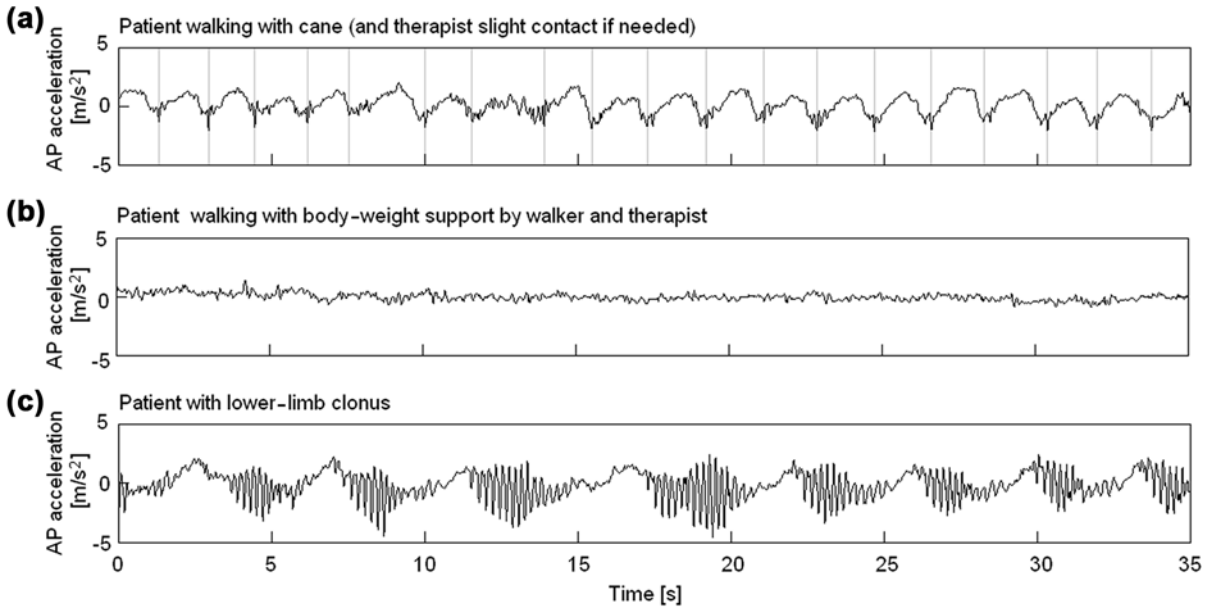

Figure 2.
Pilot cases: anteroposterior (AP) acceleration signals of three subjects collected during pilot tests. **(a)** Severely affected patient walking with cane and supervision of therapist. If needed, therapist also provided slight contact with hand on patient's back. **(b)** Patient walking with need of body-weight support provided by walker and therapist. **(c)** Patient with lower-limb clonus.

subjects reported in **Figure 1**. The smoothness of his acceleration profile was due to the stabilizing effect of the therapist contact and his slow gait speed. In fact, he performed about 20 steps in 35 s, being one of the most severely affected patients included in the study. **Figure 2(b)** shows the AP acceleration signal of a patient walking very slowly and with the need of body-weight support provided by a walker and a therapist. His trunk acceleration profile was quite blocked during walking, and determining stride partitioning from the AP signal was arduous. Finally, **Figure 2(c)** shows the AP acceleration signal of a patient with lower-limb clonus: a high frequency signal due to clonus was clearly superimposed on the slow main movement. These two last patients were excluded from this study.

Methodological Issue: Mean and Overall Acceleration

Small differences were observed in **Figure 1** between the values of $RMS_{overall}$ and those of RMS_{mean}. After analyzing the data from all subjects enrolled in the three groups, we found that the two RMS values were highly correlated with each other ($R > 0.99$) and their difference was less than 2 percent of their mean. However, these small differences were systematic and, hence, statistically significant. In fact, RMS_{mean} was higher than $RMS_{overall}$ in all the HFG and MFG subjects along the AP axis ($p < 0.001$ for both) and in 11 HFG subjects along the CC axis ($p = 0.02$, two-tailed paired *t*-test), whereas RMS_{mean} was lower than $RMS_{overall}$ in 13 of 15 LFG subjects along the LL axis ($p = 0.004$).

As shown in **Figure 1**, the RMS_{mean} could vary stride by stride in the same subject. However, the intrasubject variability was lower than the intersubject variability assessed within each group. This resulted in good reliability of acceleration RMS_{mean} values as proven by high ICC values for the HFG (ICC = 0.966, 0.889, and 0.941, along the CC, LL, and AP axes, respectively), the MFG (0.966, 0.852, and 0.944, along the CC, LL, and AP axes, respectively), and the LFG (0.934, 0.898, and 0.953, along the CC, LL, and AP axes, respectively). This good reliability supports the suitability of RMS_{mean}, computed by averaging among many RMS values, each of which was computed over a single stride. Because of the high correlation between RMS_{mean} and $RMS_{overall}$ and in order not to duplicate the results of the acceleration and speed relationship, in the following analyses we investigated only the values of RMS_{mean}.

Biomechanical Issue: Acceleration and Speed

Another important aspect shown in **Figure 1** is that the HFG subject completed the task in less time (about 8 s) and with higher trunk AP acceleration (RMS_{mean} = 1.46 m/s^2) than the older, nondisabled MFG subject (time = 12 s and RMS_{mean} = 1.20 m/s^2). As expected, the patient took the longest time to complete the task (18 s). Despite this slow WS, the patient's trunk AP acceleration (RMS_{mean} = 1.45 m/s^2) was similar to that of the younger and faster nondisabled subject. Higher accelerations were observed in all young adults in the HFG. A significant effect of functionality was observed among groups ($F_{2,42}$ = 37.28, p < 0.001). Significantly different values were recorded along different body axes within groups ($F_{2,84}$ = 134.82, p < 0.001), and the interaction between functionality and axis was also statistically significant ($F_{4,84}$ = 17.65, p < 0.001, **Figure 3**). Post hoc analyses revealed significant differences among all three groups of subjects in terms of RMS values (p < 0.005 for all the pairwise comparisons, **Figure 3**). Nevertheless, this result was affected by the association between WS and acceleration RMS values (**Table**).

Conversely, the HR, a parameter less dependent on WS and representing the smoothness of accelerations, was lower in the LFG along the AP and CC axes (p < 0.001, indicated with stars in **Figure 3**). For the HFG and MFG, the HR was not dependent on WS (p > 0.3), while a significant positive correlation was only found between HR along the AP axis and WS in the LFG (R = 0.664, p = 0.007). It was probably due to the fact that the most affected patients had both a slower and less symmetrical gait (**Table**).

Significant effects of functional level ($F_{2,42}$ = 20.66, p < 0.001), body axis ($F_{2,84}$ = 39.96, p < 0.001), and their interaction ($F_{4,84}$ = 6.50, p < 0.001) were found for HR (**Figure 2**).

Obviously, the lower acceleration RMS values found for the LFG were not due to a higher dynamic stability but to a slower gait. In fact, a significant difference was found among groups for WS ($F_{2,42}$ = 28.96, p < 0.001). Post hoc analysis confirmed that the patients' WS (0.60 ± 0.29 m/s) was significantly lower than that of the HFG (1.13 ± 0.11 m/s, LFG vs HFG: p < 0.001) and MFG (1.02 ± 0.16 m/s, LFG vs MFG: p < 0.001), whereas no significant differences were found between the two groups of nondisabled subjects (MFG vs HFG: p = 0.29). Lower RMS values were due to slower WS, as confirmed by the positive correlation found between these two parameters for the LFG (R = 0.838, p < 0.001; R = 0.646, p = 0.09; R = 0.687, p = 0.005, respectively, for RMS_{CC}, RMS_{LL}, and RMS_{AP}), the MFG (R = 0.820, p < 0.001; R = 0.702, p = 0.004; R = 0.752, p = 0.001, respectively, for RMS_{CC}, RMS_{LL}, and RMS_{AP}), and the HFG (R = 0.750, p = 0.001; R = 0.716, p = 0.003; R = 0.584, p = 0.02, respectively, for RMS_{CC}, RMS_{LL}, and RMS_{AP}). **Figure 4** shows these relationships between RMS values and WS. The best fit reported in this figure for each data set was chosen as that with the highest adjusted R^2 value. Different RMS-WS relationships were observed for different groups of subjects, despite faster patients having similar RMS values to slower elderly subjects.

The RMS, normalized by dividing by the square of WS and multiplying for step length [12], was inversely and still significantly related to WS for the LFG (R = –0.724, p = 0.002; R = –0.785, p = 0.001; R = –0.753, p = 0.001, respectively, along the CC, LL, and AP axes). Hence, the relationship between this normalized RMS and functional

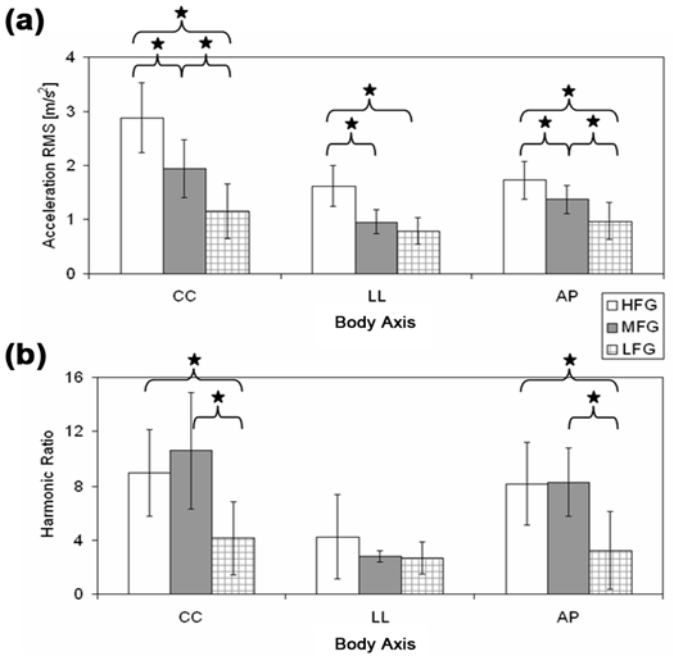

Figure 3.
Mean and standard deviation of **(a)** acceleration root-mean-square (RMS) and **(b)** harmonic ratio (parameter is dimensionless) for high functional group (HFG) (white), medium functional group (MFG) (gray), and low functional group (LFG) (grid) along craniocaudal (CC), laterolateral (LL), and anteroposterior (AP) body axes. Stars indicate statistically significant difference between two groups highlighted by post hoc analyses performed with Bonferroni correction.

Table.
Correlation results for comparison of measured parameters with walking speed (WS) and clinical scores. Pearson (r) and Spearman (ρ) correlation coefficients and relevant p-values are shown. Statistically significant correlations shown in boldface.

Parameter	WS		RMI		BI		FAC	
	r	***p*-Value**	ρ	***p*-Value**	ρ	***p*-Value**	ρ	***p*-Value**
WS	—	—	0.866	**<0.001**	0.802	**<0.001**	0.736	**0.002**
RMS_{CC}	0.838	**<0.001**	0.632	**0.01**	0.540	**0.04**	0.635	**0.01**
RMS_{LL}	0.646	**0.009**	0.440	0.10	0.418	0.12	0.406	0.13
RMS_{AP}	0.687	**0.005**	0.534	**0.04**	0.452	0.09	0.643	**0.01**
RMS_{LL} / RMS_{CC}	–0.726	**0.002**	–0.588	**0.02**	–0.499	0.06	0.625	**0.01**
RMS_{AP} / RMS_{CC}	–0.711	**0.003**	–0.596	**0.02**	–0.518	**0.048**	–0.236	0.40
HR_{CC}	0.366	0.18	0.397	0.14	0.621	**0.01**	0.498	0.06
HR_{LL}	0.262	0.35	0.430	0.11	0.396	0.14	0.356	0.19
HR_{AP}	0.664	**0.007**	0.531	0.04	0.429	0.11	0.434	0.11

BI = Barthel Index, FAC = Functional Ambulatory Category, HR_{AP} = harmonic ratio along anteroposterior axis, HR_{CC} = harmonic ratio along craniocaudal axis, HR_{LL} = harmonic ratio along laterolateral axis, RMI = Rivermead Mobility Index, RMS = root-mean-square, RMS_{AP} = RMS value along anteroposterior axis, RMS_{CC} = RMS value along craniocaudal axis, RMS_{LL} = RMS value along laterolateral axis.

status as assessed by the RMI (significant along the LL and AP axes: $p < 0.001$) could still be imputable to WS dependency. In fact, for patients, the RMI score was positively correlated with the WS and therefore also with the RMS values (**Table**). Similar positive correlations were also found when the agreement between the BI and FAC with WS and RMS values was tested (**Table**).

As evident in **Figure 4**, the RMS_{CC} seemed to be the parameter most dependent on WS in all three groups of subjects. The RMS_{AP}/RMS_{CC} and RMS_{LL}/RMS_{CC} ratios were significantly different among groups ($F_{2,42} = 9.26$, $p < 0.001$) and axes ($F_{1,42} = 48.67$, $p < 0.001$); the interaction of group and axis was also statistically significant ($F_{2,42} = 6.94$, $p = 0.002$). The values of these ratios were significantly higher for the LFG than for the HFG ($p < 0.001$) and MFG ($p = 0.02$) (**Figure 5)**. Furthermore, these values were not significantly correlated with WS for the HFG and MFG and were inversely correlated with WS for the LFG ($R = –0.711$, $p = 0.003$; $R = –0.726$, $p = 0.002$, respectively, for RMS_{AP}/RMS_{CC} and RMS_{LL}/RMS_{CC}). Finally, significant correlations were found when these two values and the three HRs were compared with the clinical scores, as detailed in the **Table**.

DISCUSSION

Patients with a severe impairment, as those with subacute stroke, have many difficulties managing the dynamic instabilities involved in locomotion [14–15]. The reduction in WS typical of these patients is associated with a reduction in their upper-body accelerations; yet there is limited understanding of how gait speed influences upper-body dynamic stability during walking [19]. Gait dynamic stability could be defined as the capacity to move the body segments in a coordinated fashion so that the body can be displaced with a proper speed (i.e., functional to the required task, such as crossing the road safely), minimizing upper-body accelerations [24]. The acceleration values are affected by the self-selected WS of each subject [4,10]. This might be the leading cause of the low clinical use of accelerometers, despite the possible advantages of these devices over conventional systems for gait analysis: low weight, low cost, wireless, wearable, informative data, and ease of use in terms of preparation and data analysis [1,25]. To avoid this problem, in some studies researchers have measured the amount of acceleration attenuation from the pelvis to the head instead of the acceleration absolute values [6,10]. However, this approach requires the use of more than one accelerometer, and it is not focused on the dynamic balance maintained by a body segment in close proximity to the body's center of mass [1,19].

The dichotomy of whether intersubject differences in acceleration patterns are related to an impairment of the neuromuscular system or simply to a difference in gait speed is an issue that needs to be clearly addressed in order to allow clinical adoption of accelerometric techniques

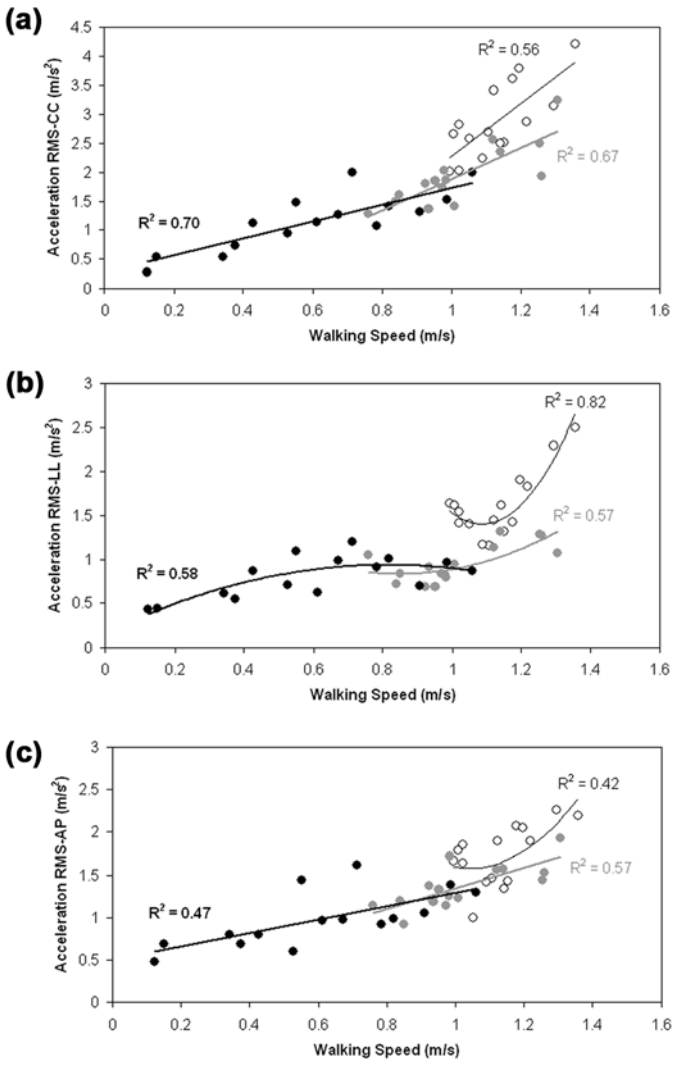

Figure 4.
Relationship between acceleration root-mean-square (RMS) and walking speed. Acceleration RMS values computed along **(a)** craniocaudal (CC), **(b)** laterolateral (LL), and **(c)** anteroposterior (AP) body axes were plotted against self-selected walking speed for high functional group subjects (empty circles), medium functional group subjects (gray circles), and low functional group subjects (black circles). Relevant best fits (chosen between linear and quadratic regression as those having higher adjusted R^2) and coefficient of determination (R^2) also reported.

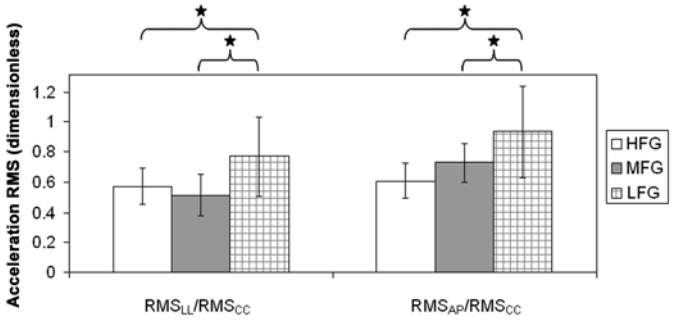

Figure 5.
Mean and standard deviation of acceleration root-mean-square (RMS) ratios between RMS values evaluated along different body axes (RMS_{LL}/RMS_{CC} on left and RMS_{AP}/RMS_{CC} on right) for high functional group (HFG) (white), medium functional group (MFG) (gray), and low functional group (LFG) (grid). Stars indicate statistically significant difference between two groups highlighted by post hoc analyses performed with Bonferroni correction. RMS_{AP} = RMS value along anteroposterior axis, RMS_{CC} = RMS value along craniocaudal axis, RMS_{LL} = RMS value along laterolateral axis.

[19]. But before facing the issue of the relationship between center of mass speed and acceleration, we investigated the effect of different computations of acceleration RMS on the resultant values. We found significant differences between the RMS values when computed along the entire walking trial ($RMS_{overall}$) versus when RMS values were computed on single strides (RMS_{mean}), the two techniques most commonly used in previous studies [1–5,7,10,12,15,17–18]. However, the small magnitude of these differences, less than 2 percent, and the optimal correlation between $RMS_{overall}$ and RMS_{mean} values seem to allow for the use of both computations. Hence, our results suggest that both methodologies can be used, but the comparison of results obtained with different computations needs particular attention. It should be carefully taken into account that the small computational differences, being systematic, can result in a statistically significant difference due only to a methodological inconsistency.

It is noteworthy that the computation over the entire walking trial avoids the partitioning of single strides that could sometimes be difficult, as shown in **Figure 2(b)**. This is probably the reason why the $RMS_{overall}$ was sometimes used when gait stability was investigated for severely affected patients [12] or for elderly patients walking on irregular pathways [5,17–18]. However, as shown in this figure, the feasibility of this acceleration technique to assess gait dynamic balance in the presence of a body-weight support stabilizing the trunk needs careful further study.

The other issue addressed in this study was the close association between acceleration RMS and WS. For

young and older nondisabled people, the values of RMS_{mean} obtained in this study and their relationship with WS were found to be in accordance with previous literature [4,6,17]. Slower WS and lower trunk accelerations were observed in patients. It has already been reported that reduced WS is a compensatory strategy to reduce upper-body accelerations and maintain balance [18]. However, despite their lower mean speed of progression, their speed fluctuations implied higher accelerations in the AP and LL axes with respect to those in CC axis, as well as reduced HRs. This finding provided quantitative information about the difficulties encountered by our patients in controlling upper-body stability.

Even though we did not compare the intra- versus intersubject RMS-WS relationship, the results of this research support the idea that it could be population specific. This relationship was found to be different among groups of subjects and along body axes, and the interaction of these last two factors also seemed to play a significant role. Indeed, this relationship was approximately quadratic along the LL and linear along the CC axes, whereas it was group-dependent along the AP axis. Moreover, the goodness of the fit (assessed by adjusted R^2) was very different among groups and axes.

It should be noted that this relationship was quite different even between the HFG and MFG, two groups with no significant differences in WS. A number of studies have found lower WS together with higher trunk accelerations in elderly people compared with young subjects [2,5,17]. This discrepancy is probably due to the fact that MFG could not be defined as representative of elderly people, but it is representative of a nondisabled group of subjects matching the age range of our patients (in fact, only 8 of 15 MFG subjects were older than 65). In our study, only a trend of reduced WS and gait stability was highlighted for MFG.

Among the three groups, the RMS_{CC} seemed to be the least informative parameter because of its relationship with WS and, in agreement with literature, its biomechanical constraints [7,9]. Obviously, this finding cannot be extrapolated to all pathologies. However, for the population of people with subacute stroke, our results suggest that the analysis of the intrasubject relationship between the three RMS values (each one evaluated along a different body axis) might provide an informative interpretation of acceleration data to assess gait stability. In fact, the ratios, obtained by dividing RMS_{AP} and RMS_{LL} by RMS_{CC} were found to be not correlated with WS in nondisabled subjects (both young and older ones) and negatively correlated with WS and clinical scores in patients. This means that AP and LL accelerations (with respect to the CC accelerations) were higher in the most severely affected patients. Moreover, lateral and frontal trunk bending have been already described as higher in people with increased fall risk [7,18]. The statistically significant correlations found in this study between AP and LL accelerations and clinical scores support the validity of measuring upper-body accelerations to assess gait stability [5–7,10]. In fact, low and rhythmic upper-body accelerations can facilitate the control of equilibrium. They reduce the perturbations needing to be controlled, facilitate the processing of the vestibular system signals, stabilize the optic flow, and reduce energy expenditure [2–3,10]. Conversely, our patients, despite their slow gait (and hence reduced CC accelerations), showed high AP and LL instabilities that put them at increased risk of falling.

The main limitation of our study is the small sample size with respect to the many features of stroke. Larger samples could also include patients who need external body-weight support to walk. For them, stride partitioning based on the identification of foot contact by the accelerometric signal could be arduous. It might lead to computation of the RMS accelerations over the entire walking trial. In this case, future researchers should carefully take into account our findings. Moreover, our study has explored two main issues, but many others still need to be investigated to obtain suitable results from the analysis of accelerations in people with stroke. Nevertheless, our findings might prove to be helpful in applying accelerometry to assess gait stability in different diseases. It would be interesting to assess the effects of WS selection on the dynamic balance of people at risk of falling. Another important aspect that could be investigated is the stabilizing effect of the use of a cane or a therapist's light touch during walking. It has been shown that the use of a cane or the assistance of a therapist has immediate positive effects on WS and gait symmetry in people with stroke [26–27]. It is conceivable that these aids can also reduce upper-body accelerations and increase gait HR. However, the small sample size of our patient group and the fact that we did not test the patients with and without aids did not allow us to quantify these effects in the present study. Finally, researchers should investigate whether the measurement of upper-body accelerations can be a prognostic factor of patients' risk of falling.

In light of our results, it seems clear that rehabilitation could take advantage of the use of accelerometry as an objective tool to assess patients' locomotor ability and evaluate progressive motor outcomes. However, accelerometric data still need to be summarized in a few informative parameters that are easily interpreted by clinicians with respect to normative ranges.

CONCLUSIONS

Our findings support the hypothesis of this study that upper-body dynamic stability can be accurately assessed by measuring upper-body accelerations in patients with stroke. In fact, for patients with subacute stroke, the present study confirms the suitability of accelerometric techniques to provide informative results about their upper-body dynamic stability during walking. By addressing methodological and biomechanical issues to interpret accelerometric data in a proper manner, our work will contribute to the transfer of this technique from research laboratories to clinical practice.

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REFERENCES

1. Kavanagh JJ, Menz HB. Accelerometry: a technique for quantifying movement patterns during walking. Gait Posture. 2008;28(1):1–15. [PMID:18178436] http://dx.doi.org/10.1016/j.gaitpost.2007.10.010
2. Winter DA. Human balance and posture control during standing and walking. Gait Posture. 1995;3(4):193–214. http://dx.doi.org/10.1016/0966-6362(96)82849-9
3. Kavanagh JJ, Barrett RS, Morrison S. Upper body accelerations during walking in healthy young and elderly men. Gait Posture. 2004;20(3):291–98. [PMID:15531176] http://dx.doi.org/10.1016/j.gaitpost.2003.10.004
4. Mazzà C, Iosa M, Pecoraro F, Cappozzo A. Control of the upper body accelerations in young and elderly women during level walking. J Neuroeng Rehabil. 2008;5:30. [PMID:19014631] http://dx.doi.org/10.1186/1743-0003-5-30
5. Marigold DS, Patla AE. Age-related changes in gait for multi-surface terrain. Gait Posture. 2008;27(4):689–96. [PMID:17962018] http://dx.doi.org/10.1016/j.gaitpost.2007.09.005
6. Mazzà C, Iosa M, Picerno P, Cappozzo A. Gender differences in the control of the upper body accelerations during level walking. Gait Posture. 2009;29(2):300–3. [PMID:19013799] http://dx.doi.org/10.1016/j.gaitpost.2008.09.013
7. Mazzà C, Zok M, Cappozzo A. Head stabilization in children of both genders during level walking. Gait Posture. 2010;31(4):429–32. [PMID:20163963] http://dx.doi.org/10.1016/j.gaitpost.2010.01.012
8. Lamoth CJ, Meijer OG, Wuisman PI, van Dieën JH, Levin MF, Beek PJ. Pelvis-thorax coordination in the transverse plane during walking in persons with nonspecific low back pain. Spine. 2002;27(4):E92–99. [PMID:11840116] http://dx.doi.org/10.1097/00007632-200202150-00016
9. Lamoth CJ, van Deudekom FJ, van Campen JP, Appels BA, de Vries OJ, Pijnappels M. Gait stability and variability measures show effects of impaired cognition and dual tasking in frail people. J Neuroeng Rehabil. 2011;8:2. [PMID:21241487]
10. Iosa M, Mazzà C, Pecoraro F, Aprile I, Ricci E, Cappozzo A. Control of the upper body movements during level walking in patients with facioscapulohumeral dystrophy. Gait Posture. 2010;31(1):68–72. [PMID:19782569] http://dx.doi.org/10.1016/j.gaitpost.2009.08.247
11. Iosa M, Morone G, Bragoni M, De Angelis D, Venturiero V, Coiro P, Pratesi L, Paolucci S. Driving electromechanically assisted Gait Trainer for people with stroke. J Rehabil Res Dev. 2011;48(2):135–46. [PMID:21480088] http://dx.doi.org/10.1682/JRRD.2010.04.0069

12. Mizuike C, Ohgi S, Morita S. Analysis of stroke patient walking dynamics using a tri-axial accelerometer. Gait Posture. 2009;30(1):60–64. [PMID:19349181] http://dx.doi.org/10.1016/j.gaitpost.2009.02.017
13. de Oliveira CB, de Medeiros IR, Frota NA, Greters ME, Conforto AB. Balance control in hemiparetic stroke patients: main tools for evaluation. J Rehabil Res Dev. 2008;45(8):1215–26. [PMID:19235121] http://dx.doi.org/10.1682/JRRD.2007.09.0150
14. Harris JE, Eng JJ, Marigold DS, Tokuno CD, Louis CL. Relationship of balance and mobility to fall incidence in people with chronic stroke. Phys Ther. 2005;85(2):150–58. [PMID:15679466]
15. Belgen B, Beninato M, Sullivan PE, Narielwalla K. The association of balance capacity and falls self-efficacy with history of falling in community-dwelling people with chronic stroke. Arch Phys Med Rehabil. 2006;87(4):554–61. [PMID:16571397] http://dx.doi.org/10.1016/j.apmr.2005.12.027
16. Brandes M, Zijlstra W, Heikens S, van Lummel R, Rosenbaum D. Accelerometry based assessment of gait parameters in children. Gait Posture. 2006;24(4):482–86. [PMID:16427287] http://dx.doi.org/10.1016/j.gaitpost.2005.12.006
17. Menz HB, Lord SR, Fitzpatrick RC. Acceleration patterns of the head and pelvis when walking on level and irregular surfaces. Gait Posture. 2003;18(1):35–46. [PMID:12855299] http://dx.doi.org/10.1016/S0966-6362(02)00159-5
18. Menz HB, Lord SR, Fitzpatrick RC. Age-related differences in walking stability. Age Ageing. 2003;32(2):137–42. [PMID:12615555] http://dx.doi.org/10.1093/ageing/32.2.137
19. Kavanagh JJ. Lower trunk motion and speed-dependence during walking. J Neuroeng Rehabil. 2009;6:9. [PMID:19356256] http://dx.doi.org/10.1186/1743-0003-6-9
20. Moe-Nilssen R. A new method for evaluating motor control in gait under real-life environmental conditions. Part 2: Gait analysis. Clin Biomech (Bristol, Avon). 1998;13(4–5): 328–35. [PMID:11415804] http://dx.doi.org/10.1016/S0268-0033(98)00090-4
21. Thieme H, Ritschel C, Zange C. Reliability and validity of the functional gait assessment (German version) in subacute stroke patients. Arch Phys Med Rehabil. 2009; 90(9):1565–70. [PMID:19735785] http://dx.doi.org/10.1016/j.apmr.2009.03.007
22. Bohannon RW. Walking after stroke: Comfortable versus maximum safe speed. Int J Rehabil Res. 1992;15(3):246–48. [PMID:1428391] http://dx.doi.org/10.1097/00004356-199209000-00009
23. Philbeck JW, Woods AJ, Arthur J, Todd J. Progressive locomotor recalibration during blind walking. Percept Psychophys. 2008;70(8):1459–70. [PMID:19064490] http://dx.doi.org/10.3758/PP.70.8.1459
24. Cappozzo A. Low frequency self-generated vibration during ambulation in normal men. J Biomech. 1982;15(8): 599–609. [PMID:7142226] http://dx.doi.org/10.1016/0021-9290(82)90071-9
25. Donati M, Camomilla V, Vannozzi G, Cappozzo A. Anatomical frame identification and reconstruction for repeatable lower limb joint kinematics estimates. J Biomech. 2008; 41(10):2219–26. [PMID:18550066] http://dx.doi.org/10.1016/j.jbiomech.2008.04.018
26. Beauchamp MK, Skrela M, Southmayd D, Trick J, Kessel MV, Brunton K, Inness E, McIlroy WE. Immediate effects of cane use on gait symmetry in individuals with subacute stroke. Physiother Can. 2009;61(3):154–60. [PMID:20514177] http://dx.doi.org/10.3138/physio.61.3.154
27. Hesse S, Jahnke MT, Schaffrin A, Lucke D, Reiter F, Konrad M. Immediate effects of therapeutic facilitation on the gait of hemiparetic patients as compared with walking with and without a cane. Electroencephalogr Clin Neurophysiol. 1998;109(6):515–22. [PMID:10030684] http://dx.doi.org/10.1016/S1388-2457(98)00033-9

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