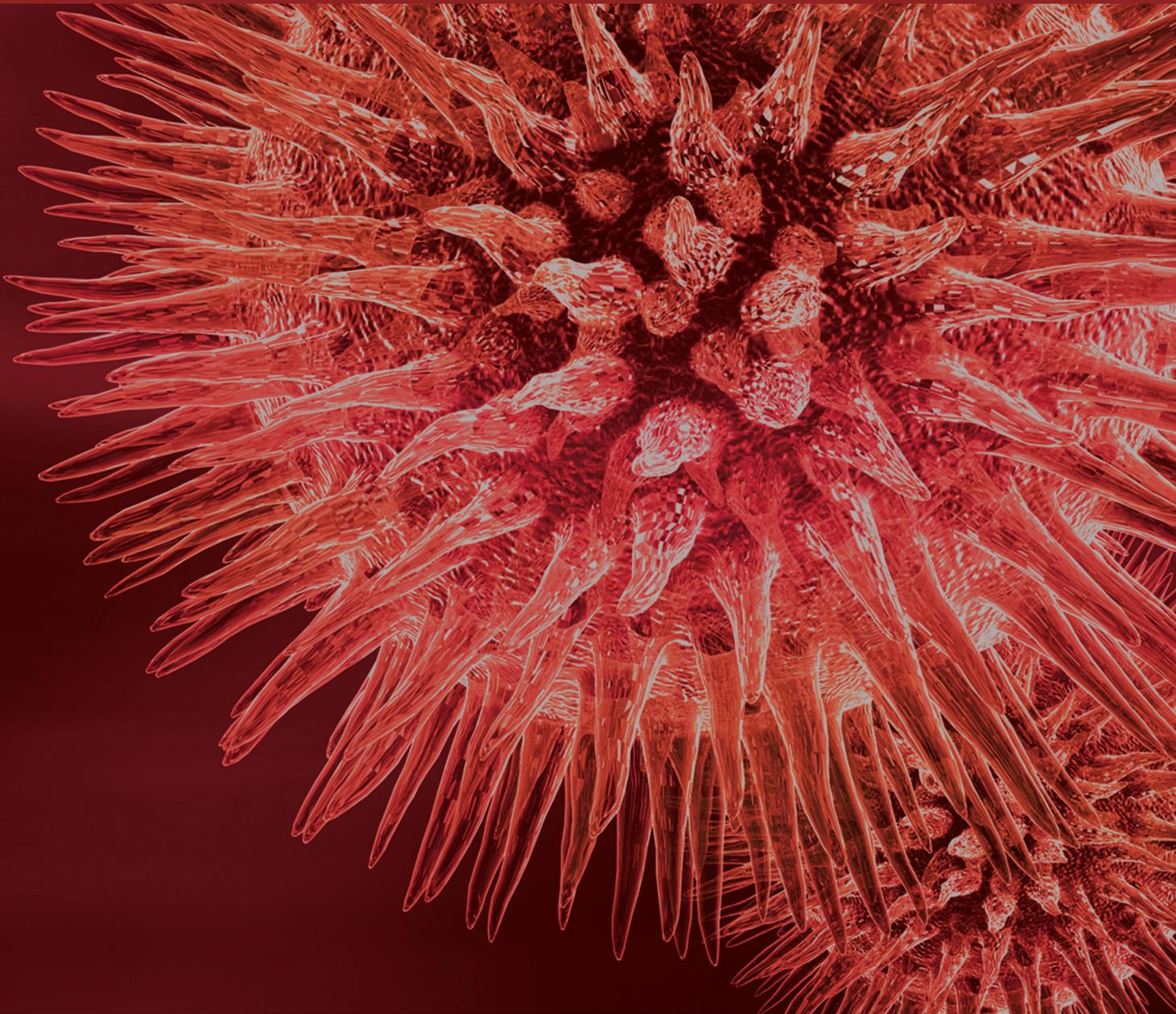


# Technological Advances in Instrumental Assessment in Rehabilitation

Guest Editors: Giorgio Ferriero, Stefano Carda, Sasa Moslavac, and Alessia Rabini





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BioMed Research International

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## Editorial

# Technological Advances in Instrumental Assessment in Rehabilitation

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In rehabilitation research, interest in instrumental assessment is rapidly growing, particularly in the last decade. A large number of tools for instrumental assessment are now available, evaluating different aspects of the single patient or patient groups. Most of these assessment tools are disease-specific and common to other medical disciplines, for example, goniometers, and clinical tests or scales that monitor patient impairment [1]. Technological advances now make it possible to perform an in-depth evaluation of patients, analyzing their abilities across a wide range of performances. In rehabilitation, high-technology assessment tools mainly concern diagnostic devices—used to obtain outcome measurements of variables of interest—or specific equipment that is necessary to apply the tests.

In fact, in a period of increasing application of measures in clinical practice, quality control, and audit procedures, assessment has become a key process in the drive to replace the empirical approach with a scientific methodology, fundamental both to the practice of evidence-based medicine and to the strengthening of the quality of research [1]. Assessment is mainly based on a measurement process characterized by the assignment of numerical values or categories to show (according to predefined rules) the quantity of certain characteristics, functions, or behaviors.

The possibility of having an objective measurement represents a fundamental advantage in several ways; for example,

it provides a scientific basis for interprofessional communication, it documents the effectiveness of treatments, and it attests their scientific credibility. Therefore, researchers are motivated to develop new instrumental assessment tools or improve old ones, demonstrating their good psychometric properties and limits. On the other hand, clinicians, who are going to use a measuring instrument, are invited to base their choice on the presence of the psychometric characteristics necessary for the specific purpose and context (preferring instruments for which the application has already been tested under conditions similar to those of interest).

Numerous scientific studies have described the main criteria for selecting an outcome measure [2, 3] and/or evaluating in detail its main psychometric properties and practices [4]. In general, the basic criterion for the choice of an instrumental assessment tool is the presence (as demonstrated through scientific publications) of adequate levels of reliability (the degree to which a measurement is free from error and, hence, the observed score gives a “true” picture), validity (degree of accuracy with which a tool measures what it is intended to measure), and responsiveness (the ability of an instrument to identify modifications or significant differences from the clinical point of view). The first two criteria are necessary for discriminative purposes (differences between subjects or groups) and predictive purposes (classification of subjects in predefined classes for prognostic purposes), while

for evaluation purposes (i.e., to detect changes over time within subjects, as in the case of analysis of effectiveness of therapeutic interventions) a good level of responsiveness is also needed. Other requirements that are extremely important to consider when selecting an outcome measure are the appropriateness (degree to which the instrument responds to the questions that the specific evaluation intends to study) and accuracy (the degree to which the measuring instrument is able to capture real differences) [2, 5].

In this special issue, we invited researchers to contribute with original research articles as well as reviews investigating the benefits of instrumental assessment or to propose new technological modalities for instrumental assessment in rehabilitation.

Our aim is to stimulate researchers to publish their research in the field of technological assessment in PMR. A wide array of topics is discussed in this special issue, related to areas such as strength assessment, posture, balance and gait analysis, functional assessment tools, and cognitive and robotic assessment. Robotic devices and passive instrumented orthoses have been proposed to assess upper limb patients affected by stroke. New software for computers has been shown to improve the cognitive assessment of neurological patients, facilitating the creation of large databases and opening up new opportunities for home-based rehabilitation. Novel technological devices and assessment protocols have been demonstrated to be reliable in the evaluation of basic motor performances, in postural control, and in gait analysis.

We are edified by the large number of papers submitted and by their high scientific level.

Finally, we wish to thank not only the authors but also the expert reviewers who, with their valuable work, have made possible the publication of this special issue.

Giorgio Ferriero  
Stefano Carda  
Sasa Moslavac  
Alessia Rabini

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## Research Article

# Ankylosing Spondylitis and Posture Control: The Role of Visual Input

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**Objectives.** To assess the motor control during quiet stance in patients with established ankylosing spondylitis (AS) and to evaluate the effect of visual input on the maintenance of a quiet posture. **Methods.** 12 male AS patients (mean age  $50.1 \pm 13.2$  years) and 12 matched healthy subjects performed 2 sessions of 3 trials in quiet stance, with eyes open (EO) and with eyes closed (EC) on a baropodometric platform. The oscillation of the centre of feet pressure (CoP) was acquired. Indices of stability and balance control were assessed by the sway path (SP) of the CoP, the frequency bandwidth (FB1) that includes the 80% of the area under the amplitude spectrum, the mean amplitude of the peaks (MP) of the sway density curve (SDC), and the mean distance (MD) between 2 peaks of the SDC. **Results.** In severe AS patients, the MD between two peaks of the SDC and the SP of the center of feet pressure were significantly higher than controls during both EO and EC conditions. The MP was significantly reduced just on EC. **Conclusions.** Ankylosing spondylitis exerts negative effect on postural stability, not compensable by visual inputs. Our findings may be useful in the rehabilitative management of the increased risk of falling in AS.

## 1. Introduction

Ankylosing spondylitis (AS) is a chronic inflammatory joint disease, enclosed in the group of spondyloarthritis, most commonly affecting the axial skeleton, usually associated with the presence of HLA-B27 [1]. Sacroiliitis and spine stiffness, due to consolidation of the articulating surfaces and inflammation, represent major clinical features of the disease [2]. Usually postural changes may be observed in early disease, becoming more marked over time. With few exceptions,

AS also leads to a rigid thoracolumbar kyphotic deformity. Consequently, patients standing in a stooped position exhibit difficulties in looking up [3] and in performing daily living activities. Moreover, in this clinical setting, poor posture may induce impairment of balance and higher risk of falls [4]. On one hand, the pathophysiological mechanism of disability in AS represents a very intriguing issue to be addressed since it provides new therapeutic opportunities in the management of these patients. In our knowledge, in literature little is reported about balance impairment in AS [5, 6]. Some reports

showed that AS patients exhibit a poorer balance as compared to normal subjects [4] and a reduced ability to balance themselves after position changes [7]. Postural balance is a complex function involving many neuromuscular processes [8, 9]. It is controlled by sensory input, central processing, and neuromuscular responses, including the vestibular, visual, and proprioceptive systems. Balance control is essential in all postural conditions, both static and dynamic, requiring integrity of neuromuscular system and an adequate muscle strength. Moreover, previous studies demonstrated the key role of visual inputs in achieving a steady posture control in normal subjects [10, 11], especially in the elderly [12, 13] and in subjects affected by neuromuscular diseases [14–17]. Furthermore, rehabilitation programs based on the recovery of visuomotor integration have been found useful in improving equilibrium [18–21]. In this regard, due to the severity of the disease, AS patients could display an impaired neuromuscular control system and balance ability, especially in absence of the visual input inflow. Therefore, they may exhibit a real dependence on the visual inputs to keep a correct posture control in static condition. Accordingly, our study is aimed to assess the standing upright control and to evaluate how the visual inflow affects the achievement of posture balance in AS patients. The quantitative assessment of posture control impairment both in open and closed eyes condition could be useful in understanding the alterations of the complex system that integrates this sensory information in AS patients.

## 2. Methods

**2.1. Patients.** From December 2011 to May 2012, 12 consecutive male subjects (mean age  $50.1 \pm 13.2$  years, mean disease duration  $20.1 \pm 13.2$  years) were diagnosed with AS according to the modified New York criteria [22] referring to the Rheumatology and Rehabilitation Research Unit of the “Salvatore Maugeri” Foundation (Telese Terme, Italy), and 12 matched healthy controls (12 males, mean age  $43.5 \pm 4.7$  years) entered the study. All patients were treated with standard dosages of an anti-TNF- $\alpha$  agent (Infliximab, 5–8 mg/kg each 6–8 weeks) for at least 12 months [23]. In order to avoid any modifications of motor-control performance, AS subjects did not receive any rehabilitative treatment during the study. The healthy group consisted of 12 subjects enrolled among hospital personnel and matched AS patients as to demographical and anthropometric features. All patients were classified as affected by “severe AS” according to the criteria by Murray et al. [5]. In order to minimize the effect of the morning stiffness, all clinical, functional, and instrumental examinations of posture control were carried out at least 3 hours after awakening. Patients underwent a morphological examination of the spine and an instrumental assessment of posture by specific tools. Moreover, a complete clinical evaluation was performed by a trained staff. Exclusion criteria were age  $>70$  years, known balance or vestibular disorders, concomitant severe cardiovascular, neurological, or psychiatric disease, diabetes, and severe visual or auditory impairments (reduced visual acuity was accepted if adequately corrected). Patients

with attested orthopedic diseases affecting spine (fractures, spinal disc herniation, spinal surgery, etc.) or lower limbs (prosthesis) were also excluded. Patients and controls were not treated with drugs affecting the central or peripheral nervous system. The study was approved by the local ethics committee and informed consent was obtained from patients and controls.

**2.2. Clinical Assessment.** The assessment of AS was obtained through appropriate physical examination. Functional status and measures of disease activity were obtained by established criteria. The Bath Ankylosing Spondylitis Functional Index (BASFI) [24] was performed to determine the degree of function limitation. Disease activity was measured by the Bath Ankylosing Spondylitis Disease Activity Index (BASDAI) [25]. The five clinical measurements (chest expansion, modified Schober’ test, occiput to wall, cervical rotation, and lateral spinal flexion) providing the Bath AS Metrology Index (BASMI) [26] were also accounted.

**2.3. Instrumental Assessment of Posture, Task, and Procedures.** Posture control in upright stance was quantized by a baropodometric platform (FDM-S, Zebris, Germany) [27, 28]. This quantitative posturography measures the forces exerted on the ground during quiet stance obtaining an index named centre of feet pressure (CoP). The latter represents the resulting body sway and the point location of forces used to keep the body mass center projection within the platform [29]. The CoP trajectory is a key output of a complex system [30, 31] that integrates several sensory inputs (visual, somatosensory, vestibular), providing information on nervous and musculoskeletal systems’ ability to generate an adequate postural balance [32]. Accordingly, posture control disorders can be detected by changes of CoP spatiotemporal features [33].

Patient and controls were naïve to these instrumental evaluations. The participants were asked to stand quiet on the baropodometric platform as still as possible in their usual posture with arms relaxed on body sides in two visual conditions: (1) eyes open (EO), looking at a visual target adjusted for eyes’ height at a 40 cm-distance, and (2) eyes closed (EC). They stood barefoot on the platform with feet spaced 17 cm apart [34]. The room was illuminated with diffuse light and background noise was very low. Trials in which sharp directional sounds unexpectedly occurred were eliminated. Aforementioned precautions were required to simulate “natural” position during evaluations. Subjects and controls were instructed to keep the gaze fixed on a cross target embedded by two vertical lines [35] and stand still for at least 50 seconds. Each subject performed a series of 6 consecutive trials [36], 3 in EO and 3 in EC condition. The visual conditions were randomly assigned and a 1-minute rest was given every two trials to avoid participant discomfort or pain in the soles. Feet position was marked on the platform to assure consistency across trials.

**2.4. Detection and Analysis of the Center of Feet Pressure (CoP) by Baropodometric Platform.** The acquisition time

TABLE 1: Demographic, anthropometric, and clinical features of study population.

	AS subjects	Control subjects	<i>P</i>
Height (cm) mean $\pm$ sd	172.4 $\pm$ 9.6	166.3 $\pm$ 9.9	0.14
Weight (Kg) mean $\pm$ sd	84.4 $\pm$ 20.5	72.7 $\pm$ 16.8	0.16
BMI (Kg/m <sup>2</sup> ) mean $\pm$ sd	30.3 $\pm$ 6.7	26.2 $\pm$ 5.4	0.12
Age (years) mean $\pm$ sd	50.1 $\pm$ 13.2	43.5 $\pm$ 4.7	0.11
Disease duration (years) mean $\pm$ sd	20.1 $\pm$ 13.2	/	
BASFI mean $\pm$ sd	41.6 $\pm$ 22.5	/	
BASDAI mean $\pm$ sd	45.9 $\pm$ 29.9	/	
BASMI mean $\pm$ sd	6.6 $\pm$ 1.5	/	
Male gender	100%	100%	1.000

was 50 seconds. We discarded the first and last 10 seconds signals to avoid any transient periods and to analyze exclusively a stationary posture. The CoP displacement was computed off-line calculating 4 parameters (2 global and 2 structural) as recommended in clinical practice [37]. The 2 global posturographic parameters express the “extent” of the CoP oscillations in the time and frequency domains, while structural posturographic parameters examine the CoP sway patterns related to the motor control activity (posturographic motor commands). For each trial in EO and EC conditions, we computed the 2 following “global” parameters: (i) the sway path (SP) of the CoP, integrating the instantaneous velocity of the CoP over the total acquisition time. SP measures the mean velocity of CoP oscillations; therefore, its increasing addresses a reduced posture stability. (ii) The frequency bandwidth (FB1) includes the 80% of the area under the amplitude spectrum [37], for both anteroposterior (A-P) and mediolateral (M-L) directions. FB1 measures the amount of quick transient CoP displacements, separately for the frontal and sagittal anatomic planes. The FB1 increasing represents an enhancement in the effectiveness of the posturographic motor commands which account for a faster control of CoP oscillations. The 2 “structural” parameters were calculated from the sway density curve (SDC). The SDC is constructed by counting the number of consecutive points of the CoP trajectory that, for each time instant, fall inside a test circle of a radius of 2.5 mm [37]. Therefore, SDC presents a regular alternation of peaks and valley. Peaks correspond to time instants in which the CoP control is relatively stable, while valleys correspond to instants in which the CoP control rapidly shifts from one stable point to the next one. Following are the 2 calculated structural parameters: (iii) the mean amplitude of the peaks (MP) of the sway density curve. MP is directly related to the degree of stability obtained by posturographic motor commands; therefore, MP increasing represents an increased quantity of CoP trajectory stable points; (iv) the mean distance (MD) between two consecutive peaks of the sway density curve [37]. Its reduction shows a faster and more efficacy release of posturographic motor commands.

2.5. *Statistical Analysis.* Continuous data were expressed as mean  $\pm$  standard deviation; categorical variables were

expressed as %. For every group we executed a paired Student’s *t*-test between EO and EC data to evaluate significant differences between visual conditions, while for all the other statistical analysis we adopted an unpaired Student’s *t*-test. The Pearson’s *r* was used to perform correlation between continuous variables. The chi-square test was performed to compare categorical data. When the minimum expected value was  $< 5$ , Fisher’s exact test was used. All the results are given as two-tailed values with statistical significance for *P* values  $< 0.05$ . For each subject and control, we calculated the mean value of the measured variables. For all statistics, the significance level was  $P < 0.05$ . Average values  $\pm$  standard deviation (SD) over all trials for all subjects and measured variables were also computed.

### 3. Results

The demographic, anthropometric, and clinical features of study population are shown in Table 1. There were no significant differences between the groups according to age, gender, height, weight, or body mass index (BMI). Table 1 also reports the results of physical evaluations performed for each patient before instrumental posture assessment.

Figure 1 shows representative plots of CoP oscillation obtained from a healthy control and an AS patient during EO and EC conditions. For both groups, the support postural base is larger along anteroposterior (A-P) direction than mediolateral (M-L) direction. Accordingly, the extent of CoP oscillations is more marked along A-P direction than M-L direction [38]. The figure also shows increased oscillation amplitude for both groups switching from EO to EC especially along A-P direction. In particular, the figure qualitatively reports the motor postural control impairment in AS. Of interest, we found differences in terms of CoP peak to peak amplitude, oscillation frequency, and the area covered by the CoP oscillations between controls and AS patients during both visual conditions.

3.1. *Balance Behavior between Eyes-Open (EO) and Eyes-Closed (EC) Trials.* As shown in Figure 2 the mean values of CoP parameters were significantly different switching from EC to EO condition. In details, MP, MD, and SP values were

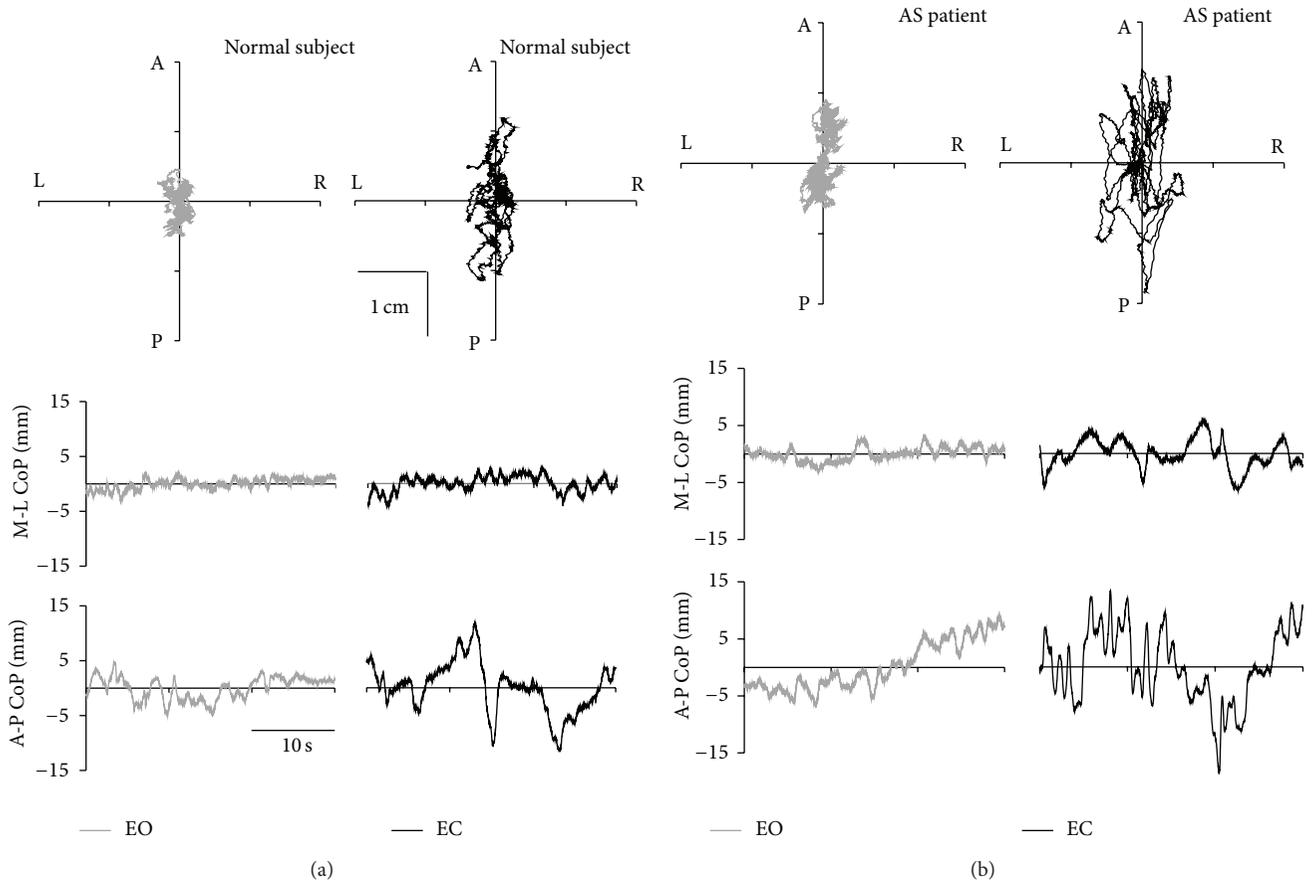


FIGURE 1: Statokinesigrams (upper 2D plots of the planar oscillation of the CoP) and stabilograms (lower traces plotted against time) of a normal subject (left half) and an AS patient (right half) plotted during eyes open (EO, gray lines) and eyes closed (EC, black lines). The stabilograms are the projection of the statokinesigram along the mediolateral (M-L, 2nd row plots) and anteroposterior (A-P, 3rd row plots) directions of the subject. A = anterior, P = posterior, L = left, R = right.

found to be significantly different for both control and AS groups ( $P < 0.005$ ), while FBI along AP direction a significant difference was found only in the control group ( $P < 0.005$ ). These findings reveal worsening of upright stance control from EO to EC condition. Therefore, in order to highlight the differences between the groups, we also evaluated the delta ( $\Delta$ ) between EO and EC for each parameter (Figure 3). We found a statistically significant difference only for  $\Delta$ -MD and  $\Delta$ -SP values between groups ( $P < 0.05$ , Figure 3). Moreover, the figure shows no difference in CoP oscillation frequency ( $\Delta$ -FBI) and in stabilizing duration time ( $\Delta$ -MP) between the groups.

In Figure 4 we report the comparison between subjects and controls according to the two visual conditions. MD and SP values were found to be significantly different between the groups ( $P < 0.05$ ), while MP was found to be different only in EC condition ( $P < 0.05$ ).

Finally, Pearson's correlation, performed to investigate the relationship between posture control indices and AS clinical outcomes, showed that BASMI was significantly related to MP values ( $r = -0.57$ ;  $P = 0.042$ ), while BASFI or BASDAI did not show a significant correlation with one of the four posturographic parameters assessed ( $P > 0.05$ ).

#### 4. Discussion

To generate goal-directed movements, such as reaching with the arm at stationary or moving objects, the brain must integrate external inputs (e.g., visual, auditory) relative to target position with intrinsic signals (proprioceptive, vestibular, motor) related to body, head, and eye positions.

The posterior parietal cortex, which is part of the visual dorsal stream, is "reciprocally" connected with motor areas of the frontal lobe representing an important sensorimotor interface for movements and posture control. To circumvent sensory feedback delays, current motor control theories postulate the existence of "forward" models, combining sensory inputs with motor commands [9]. On one hand, given the posture deterioration in AS, our study represents the first assessment of the posture stability in these patients with and without the support of visual inputs, showing that the latter is not able to effectively improve posture stability in the setting. As qualitatively and quantitatively shown in Figures 1 and 2, the availability of visual information affects postural control in both groups. Accordingly, Figure 1 shows that in healthy and AS subjects during EC, except for FBI along M-L direction, all posturographic indices were significantly

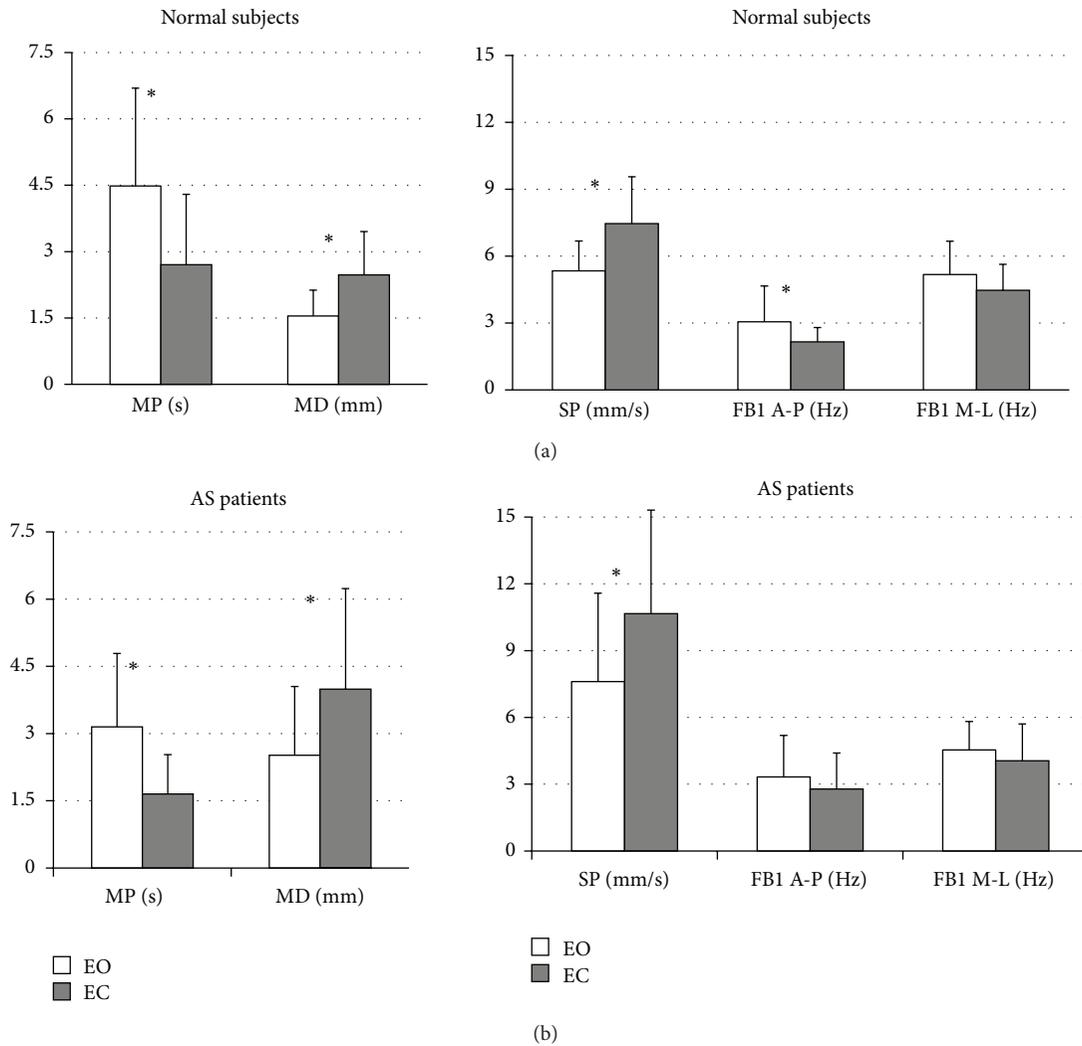


FIGURE 2: Structural parameters (MP and MD, left side) and global parameters (SP, FBI along A-P and M-L directions, right side), for normal subjects (upper panels) and AS patients (lower panels) during eyes open (EO, white columns) and eyes closed (EC, black columns). Data are expressed as mean  $\pm$  standard deviation. The asterisks represent significant difference ( $P < 0.05$ ).

different as compared to EO, evidencing a worse stability in that condition. On the other hand, only in AS patients we found no difference in FBI along A-P direction during EO versus EC. These findings, together with the significant increase in SP mean during EC, suggest that in normal subjects the control of equilibrium during EC was achieved with greater oscillations and lower velocity along A-P direction. In AS an opposite effect was found. In details, without the support of visual input, AS patients increase amplitude and velocity of CoP oscillation obtaining a still high frequency in CoP displacement. This finding is supported by the significant difference in the SP index, between the study groups (Figure 3). Moreover, the significant difference between the groups of the  $\Delta$  mean SP (calculated as the difference between EO and EC) confirmed the higher velocity of the CoP oscillation along the anteroposterior axis for AS patients when their balance was not supported by visual input. This finding shows an excessive reliance on visual

information in AS patients and a consecutive increased posture instability during EC as compared to normal subjects. Of interest, Figure 3 shows another difference between the groups. In particular, considering both MD and MP results, we may suggest that AS patients in EC condition were able to perform larger but still quicker posturographic motor commands as compared to normal subjects. This statement is further confirmed by results reported in Figure 4, showing only in EC condition a significant difference for both MP and MD values in the groups. Since AS patients have no central nervous system diseases, such modification of motor control should not derive from the deterioration of the visual sensory pathway but might be a consequence of a likely impairment of the motor controller, that is, the inability to generate quick and precise motor commands. This is evidenced by a higher amplitude of CoP oscillations (reported as higher SP values, for both EO and EC, Figure 4) and by higher MD values, especially in EC (Figure 4). Moreover, the significant

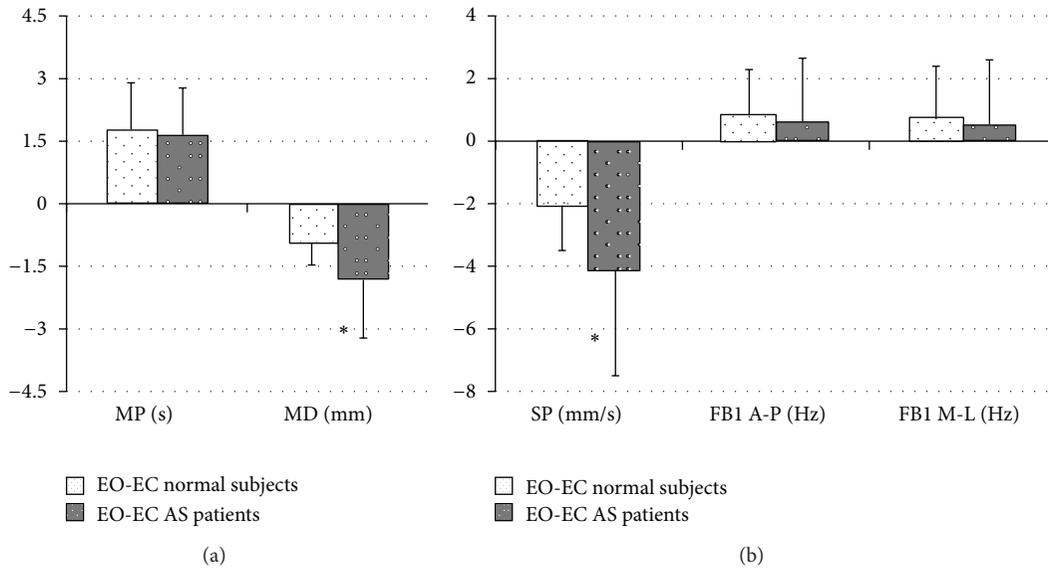


FIGURE 3: Mean  $\pm$  standard deviation of the differences between EO and EC values of the structural parameters (MP and MD, left side) and global parameters (SP, FBI along A-P and M-L directions, right side), for normal subjects (white columns with black dots) and AS patients (black columns with white dots). The asterisks represent significant difference ( $P < 0.05$ ).

differences in MD mean values, between the groups for both visual conditions (Figure 4), as well as for the  $\Delta$  mean MD (difference between EO and EC, Figure 3), could be explained by an alteration of the intrinsic feedback due to the mechanical properties of the lower limb muscles, modulated by the segmental reflexes. As shown by their larger support base (Figure 1) during EC, in AS patients the impaired ability of adopting anticipatory muscle activations in controlling the upright quiet stance posture could lead to an increased risk of falling. The impaired posture control has been already correlated with an increased risk of falling in elderly [39–41] and neuromuscular disease patients [42–44]. Falling and related problems, that is, fear of falling and daily life activity restriction, are known to contribute to consistently reducing quality of life [45]. Moreover, the correlation analysis suggests a strong link between the degree of posture stabilization, measured by MP parameter in EO and EC condition and spinal mobility assessment (BASMI). Accordingly, by BASMI evaluation, we could eventually predict the status of motor controller degradation. Our findings are in line with previous studies, which showed that AS exhibits negative effect on postural stability, [5, 6] even by different instrumental tools. These effects are more evident in the later stages of the disease [5].

In conclusion, our study, showing that the posture control in AS is not effectively improved by visual input, confirms the initial hypothesis about the existence of “forward” models, combining sensory inputs with motor commands. In particular, according to our finding, we could speculate that a chronic postural imbalance, linked to the inflammatory spine disease, may affect the central sensory feedback. As a consequence, these patients show a reduced the ability of visual input to improve body stability. Accordingly, these results may allow applying new rehabilitation methods based

on increasing proprioceptive inflow integration with rehabilitating exercises executed without visual information support. Finally, this study points out a further important issue on the correlation between severe AS and risk of falling. It highlights the importance of developing advanced rehabilitation programs aimed at reducing the CoP amplitude oscillations especially without visual input. These programs could increase the safety margin of the posture motor commands’ intervention. Indeed, monitoring patients’ balance may be of interest to develop new rehabilitation and protection methods aimed to increase postural stability and protect the elderly and severely affected patients from falling and its associated sequelae.

## List of Abbreviations

A-P:	Anteroposterior
AS:	Ankylosing spondylitis
BASDAI:	Bath ankylosing spondylitis disease activity index
BASFI:	Bath ankylosing spondylitis functional index
BASMI:	Bath ankylosing spondylitis metrology index
CoP:	Centre of foot pressure
EC:	Eyes closed
EO:	Eyes open
FBI:	Frequency bandwidth that includes the 80% of the area under the amplitude spectrum of the centre of foot pressure oscillations
MD:	Mean distance between one peak and another of the sway density curve of the centre of foot pressure

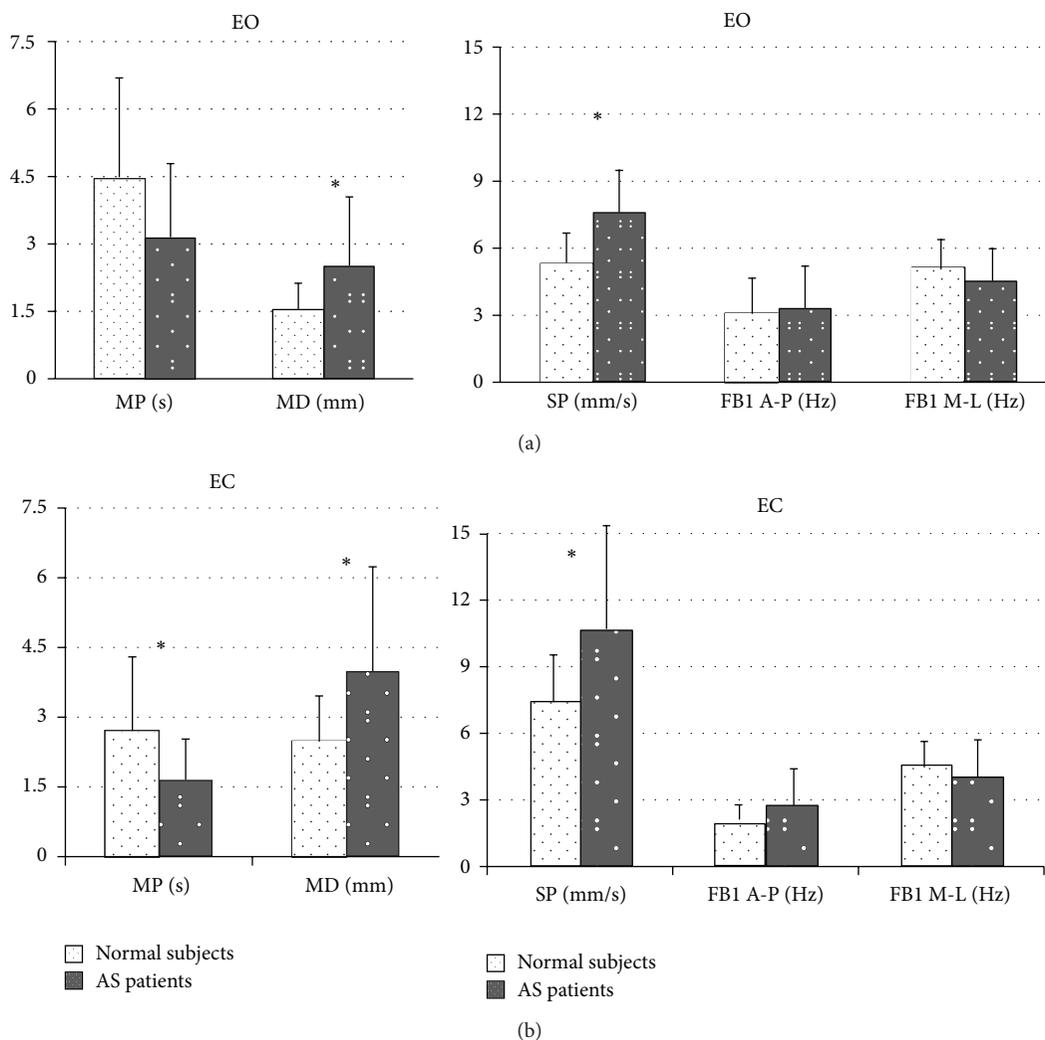


FIGURE 4: Structural parameters (MP and MD, left side) and global parameters (SP, FB1 along A-P and M-L directions, right side), for normal subjects (white columns with black dots) and AS patients (black columns with white dots). Data are expressed as mean ± standard deviation. The asterisks represent significant difference ( $P < 0.05$ ).

M-L: Mediolateral

MP: Mean amplitude of the peaks of the sway density curve of the centre of foot pressure

SDC: Sway density curve of the centre of foot pressure

SP: Sway path of the centre of foot pressure.

### 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

# Use of a Robotic Device for the Rehabilitation of Severe Upper Limb Paresis in Subacute Stroke: Exploration of Patient/Robot Interactions and the Motor Recovery Process

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This pioneering observational study explored the interaction between subacute stroke inpatients and a rehabilitation robot during upper limb training. 25 stroke survivors (age  $55 \pm 17$  years; time since stroke,  $52 \pm 21$  days) with severe upper limb paresis carried out 16 sessions of robot-assisted shoulder/elbow training (InMotion 2.0, IMT, Inc., MA, USA) combined with standard therapy. The values of 3 patient/robot interaction parameters (a guidance parameter: Stiffness, a velocity-related parameter: Slottime, and Robotic Power) were compared between sessions 1 (S1), 4 (S4), 8 (S8), 12 (S12), and 16 (S16). Pre/post Fugl-Meyer Assessment (FMA) scores were compared in 18 patients. Correlations between interaction parameters and clinical and kinematic outcome measures were evaluated. Slottime decreased at S8 ( $P = 0.003$ ), while Guidance decreased at S12 ( $P = 0.008$ ). Robotic Power tended to decrease until S16. FMA scores improved from S1 to S16 (+49%,  $P = 0.002$ ). Changes in FMA score were correlated with the Stiffness parameter ( $R = 0.4$ ,  $P = 0.003$ ). Slottime was correlated with movement velocity. This novel approach demonstrated that a robotic device is a useful and reliable tool for the quantification of interaction parameters. Moreover, changes in these parameters were correlated with clinical and kinematic changes. These results suggested that robot-based recordings can provide new insights into the motor recovery process.

## 1. Introduction

After stroke, most patients have residual upper limb (UL) motor impairments, leading to long-term limitations in function which impact quality of life [1, 2]. Motor recovery is often poor with only one-third of all stroke patients regaining some dexterity within six months [3].

Over the last 2 decades, many studies have investigated the neuroplastic changes which occur after the acute event as well as optimal strategies to restore lost motor function. This growing body of evidence has demonstrated that large numbers of movement repetitions [4–6], carried out within intense [7–9] and specific task-oriented [10, 11] training programs, are required to drive optimal neuroplastic changes and to improve function. This scientific knowledge

has stimulated the development and the use of technological devices, referred to as rehabilitation robots, to address the need for intensive training. This training is carried out under the supervision of therapists. Advanced robotic systems can provide repetitive, reproducible, and interactive forms of physical therapy which can be quantified. Since the first clinical studies of the MIT-MANUS robot [12] at the Massachusetts Institute of Technology (MIT), this innovative therapeutic tool has been clinically studied for the rehabilitation of the paretic upper limb, mainly after stroke. There are a multitude of studies of patients in the acute/subacute phase of stroke recovery [12–16] as well as in the chronic phase [17–20]. The results were very promising, showing that robotic therapy is safe and well tolerated [12, 15, 19, 20] and that it has a positive impact, improving motor impairments. These results

led to the endorsement of the use of upper extremity robotics in the 2010 guidelines of the American Heart Association for Stroke Care [21].

Robot-mediated training is highly repetitive in nature. Indeed, such systems allow stroke patients, including those with severe impairment, to repeat movements hundreds of times. This is physically impossible using usual treatment methods [22]. This feature is mostly due, in the most advanced systems, to the use of active robotic controllers [23]. The principle paradigm implemented to date consists of performance-based algorithms that enable the robot to adjust the mechanical assistance provided during the training session according to the patient's motor performance. Most robotic devices use assisted-as-needed programs, the aim of which is to provide only as much assistance as the patient requires to complete the task. However, a potential problem with most robotic controllers is the provision of excessive assistance. This can encourage patients to minimize their efforts, resulting in a reduction of experience-dependent plasticity [24, 25]. There is currently a lack of literature regarding how humans and particularly stroke survivors interact with robotic-therapy devices.

In addition to therapeutic effects, some robots can assess motor performance during robot-assisted tasks by recording biomechanical parameters (mainly position and speed of the hand). Some robots can provide new insights into the effectiveness of treatment through the capture of motion kinematics [26–28] and some can measure and record patient-robot interactions during training. Thus, some devices can continuously track motor performance through the measurement of specific indicators, including the patient's actual level of participation. Such data are difficult to obtain in usual care.

The aim of the present study was to investigate patient-robot interactions and to analyze changes and potential correlations with clinical and kinematic outcome measures during an upper limb robot-assisted training program in subacute stroke patients with severe motor impairments.

## 2. Materials and Methods

**2.1. Participants.** From October 2010 to March 2013, 48 inpatients involved in the upper limb robotic program were screened for inclusion in this observational study. These patients had undergone an upper limb robot-assisted rehabilitation program administered as part of usual care for moderately to severely motor impaired inpatients admitted to the Neurorehabilitation Unit at Les Trois Soleils Rehabilitation Center (Boissise-le-Roi, France). 25 stroke survivors (13 females, age  $55 \pm 17$  (19–88) years, 21 ischemic strokes, and 4 hemorrhagic strokes) were enrolled. The inclusion criteria were the following: being over 18 years old, with moderate to severe upper limb paresis defined by a low motor score ( $\leq 35$  on the Fugl-Meyer Assessment (FMA) scale [29, 30]), being in the subacute phase of stroke (time since stroke,  $52 \pm 21$  days), with a single lesion confirmed on CT scan or MRI, and with sufficient understanding to participate in rehabilitation exercises (see Table 1). Finally, patients had to have carried out the whole upper limb robot-mediated training using the assistive robotic mode (patients who had used the passive and

TABLE 1: Patient demographics.

Characteristics ( $n = 25$ )	
Gender (male/female)	12/13
Mean age $\pm$ SD (years)	$55.5 \pm 17$
Time since stroke, mean $\pm$ SD (days)	$52.2 \pm 21.6$
Type of stroke (H/I)	4/21
FMA score S1 ( $n = 18$ ) mean $\pm$ SD [range]	$19 \pm 8.5$ [7–35]
FMA score S16 ( $n = 18$ ) mean $\pm$ SD [range]	$28 \pm 15.3$ [9–57]

H, hemorrhagic; I, ischemic; FMA, Fugl-Meyer Assessment; SD = standard deviation.



FIGURE 1: Experimental setup (upper left, setup with an individual; upper right, “clock exercise”; lower, Arm robot).

active modes within some robotic sessions were not included). This observational study was approved by the “CPP Ile de France 1” Ethics Committee.

### 2.2. Interventions and Apparatus

**2.2.1. Apparatus.** The InMotion 2.0 Arm robot (Interactive Motion Technologies, Inc., Watertown, MA, Figure 1), the commercial version of the MIT Manus, was used for the study [31]. This device is a 2 translational degrees-of-freedom planar robot that emphasizes shoulder (flexion/extension) and elbow (flexion/extension) movements in the horizontal plane. It was designed to have low intrinsic endpoint impedance with a low inertia. This device has several treatment modes, including an adaptive (or assist-as-needed) program using a performance-based algorithm that adjusts forces to assist or challenge the patient's movement according to his/her motor performance. Particularly, if a task cannot be completed volitionally, the robot provides assistance to reach the target.

The point-to-point unconstrained reaching program is mainly used for the evaluation of motion kinematics.

**2.2.2. Interventions.** All the patients underwent 16 sessions of upper limb robot-assisted training in addition to their usual

TABLE 2: Changes in interaction parameters and number of movements from S1 to S16.

$n = 25$	S1	S4	S8	S12	S16
Stiffness (N/m) mean $\pm$ SD	247 $\pm$ 28	240 $\pm$ 34	231 $\pm$ 16	221 $\pm$ 44 <sup>a</sup>	218 $\pm$ 44 <sup>a</sup>
Slottime (s) mean $\pm$ SD	1.48 $\pm$ 0.31	1.33 $\pm$ 0.23	1.27 $\pm$ 0.28 <sup>a</sup>	1.24 $\pm$ 0.27 <sup>a</sup>	1.27 $\pm$ 0.29 <sup>a</sup>
Robot (active) Power (mwatt) mean $\pm$ SD	95.5 $\pm$ 26	83.6 $\pm$ 26	82.6 $\pm$ 34	80.2 $\pm$ 34	82 $\pm$ 34
Number of movements mean $\pm$ SD	614 $\pm$ 250		780 $\pm$ 271		857 $\pm$ 342

<sup>a</sup>Versus S1,  $P < 0.05$ .

stroke rehabilitation program. Each session lasted for 45 minutes, 4 days per week.

During the training, the patient was seated on an adjustable chair in front of a monitor which displayed goal-directed exercises. The trunk was restrained by a harness to decrease compensatory movements. The paretic limb was supported at the elbow by a splint. The shoulder was in 45° elevation and the elbow slightly flexed. The wrist was in a neutral position and the fingers were placed around the handle (Figure 1).

The patient held the robot handle to perform the exercises. The motor tasks involved point-to-point gravity-compensated reaching towards 8 visual targets displayed in the 8 compass directions on the monitor and presented in a clockwise order. Each target was 14 cm from the center of the monitor. The patient was instructed to perform as many accurate movements as possible in the allocated training time. The training consisted of series of 320 repetitions (4 blocks of 80 movements). Patients were allowed a 1-to-3-minute break after each block. A summary graph also displayed patient's performance and several interaction parameters after each block (Figure 2).

During the training, patients performed an average of 614  $\pm$  250 movements during the first session (S1), 780  $\pm$  271 movements at the midpoint of the training (session 8, S8), and 857  $\pm$  342 movements during the last session (S16) (see Table 2).

Standard care for the paretic upper limb consisted of one-hour occupational therapy sessions 5 days per week; this program involved passive stretching within submaximal ranges of motion with inhibition of spasticity if necessary [32], active assisted movements, reaching movements with or without elbow support, and grasping tasks that were tailored to the abilities of each patient [33].

This comprehensive program of care also included one-hour daily (5 days a week) sessions of physical therapy based on lower limb rehabilitation (without upper limb therapy) and, if necessary, one hour of speech therapy 3-4 times a week.

**2.3. Robot-Based Outcome Measures.** The robotic device measures several parameters related to patient-robot interactions, indicating the level of assistance and/or challenge provided by the robot while the patient performed the reaching task. The values recorded after the 80th movement (out of 320) were analyzed for the 25 patients. This was because, during the first 80 movements, these parameters are frequently adjusted by the robot (every 16 movements).

The following parameters were analyzed.



FIGURE 2: Performance feedback (displayed after each block of 80 movements).

*Stiffness* is a parameter of lateral guidance. The robot adapts the stiffness of the side walls, thus regulating the amount of guidance given to the patient to produce straight movements. As patients get better at aiming, the amount of side guidance is reduced to challenge the patient to make even straighter movements. Stiffness is defined as force/displacement and is measured in Newton/meters. The default stiffness of 200 N/m was used to begin with and the adaptive algorithm then adjusted the stiffness according to the movements performed by the patient.

*Slottime* is the time allotted to the patient to achieve the task. The initial time allowed is 2 seconds. As the patient moves faster, the time is gradually decreased to 1 second. This is a velocity-related parameter.

*Robot Power* is defined as force \* velocity, calculated from the force transducer measurements (force) and the position measurements in the direction of the target (velocity). If the patient performs the whole movement without assistance, the value will be close to zero, that is, minimum interaction force registered in the transducer.

The interaction parameters were analyzed at S1, S4, S8, S12, and S16.

In addition to the interaction parameters, 2 kinematic metrics calculated from trajectory recordings carried out during a robot-based evaluation (80 movements toward 8 targets) were analyzed at S1 and S16 in 19 patients (data missing for 6 patients); the mean velocity (m/s) and the movement accuracy were calculated as the mean deviation from the straight line (m).

**2.4. Clinical Outcomes.** Motor impairment was measured before the first session and after the last session using the upper extremity motor section of the Fugl-Meyer Assessment

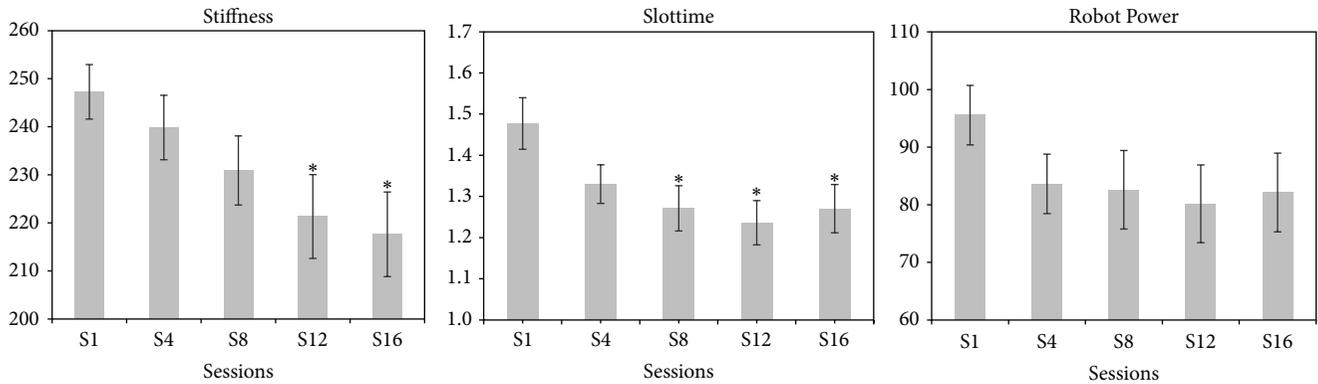


FIGURE 3: Changes in interaction parameters over the training period. \*: versus S1,  $P < 0.05$ .

(FMA) in 18 patients (7 patients with incomplete data). The FMA scale measures the ability to move the paretic arm, including items related to movements of the shoulder, elbow, wrist, and hand. Each item is rated on a 3-point scale (maximum score, 66 points).

**2.5. Statistical Analysis.** The values of the 3 interaction parameters were compared across sessions (S1, S4, S8, S12, and S16) in 25 patients (Table 2; ANOVA, post hoc Tukey test). A descriptive analysis was performed to compare FMA scores between S1 and S16 in 18 patients ( $t$ -test).

Pearson's coefficients were used to explore correlations between the interaction parameters, changes in FMA score, and selected kinematic parameters, as well as potential correlations between the 3 interaction parameters (S1 versus S16).

### 3. Results

Slottime decreased at S8 ( $P = 0.003$ ). Stiffness decreased at S12 ( $P = 0.008$ ). There was a trend towards a decrease in Robot Power (Figure 3).

The FMA score improved significantly from S1 to S16 (+49%,  $P = 0.002$ ).

The regression analysis showed that the change in FMA score was correlated with the change in Stiffness ( $r = 0.4$ ,  $P = 0.003$ ) (Figure 4) but not with the other interaction parameters. There was a good correlation ( $r = 0.35$ ) between Slottime and the mean change in velocity. There was no correlation between Stiffness and changes in movement accuracy. Finally, there was a negative correlation between the change in Stiffness and the change in Slottime ( $r = -0.6$ ,  $P = 0.001$ ) (Figure 5).

### 4. Discussion

The present study is, to our knowledge, the first to analyze how patients with moderate to severe motor impairment following stroke interact with a rehabilitation robot during upper limb training carried out as part of a stroke rehabilitation program in the subacute phase. Moreover, the nature

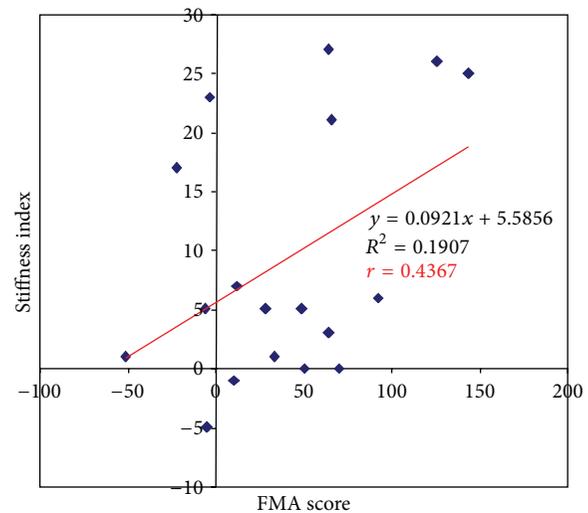


FIGURE 4: Correlation between FMA score and changes in Stiffness.

of the interactions and how they changed during the robot-mediated program were evaluated. Potential correlations with clinical scores and kinematic metrics were also analyzed.

The evaluation of interaction parameters, which is difficult or impossible in usual care, appears to be of critical interest because it affords an insight into the level of active participation of patients, as well as both quantitative and qualitative motor performance. In fact, some recent results from studies of robotic devices for gait training [34–36] demonstrated that the patient's level of engagement is a crucial determinant of robot-mediated rehabilitation. Such rehabilitation is more effective when the user actively participates in the movement, and the use of full and passive guidance could have a negative impact on recovery in stroke patients. A previous study showed that continuous passive motion using an upper limb robotic system did not provide any advantage over conventional therapy [37].

The present study confirmed that the robot used for the rehabilitation program was a truly interactive device, since it adjusted its action according to changes which occurred

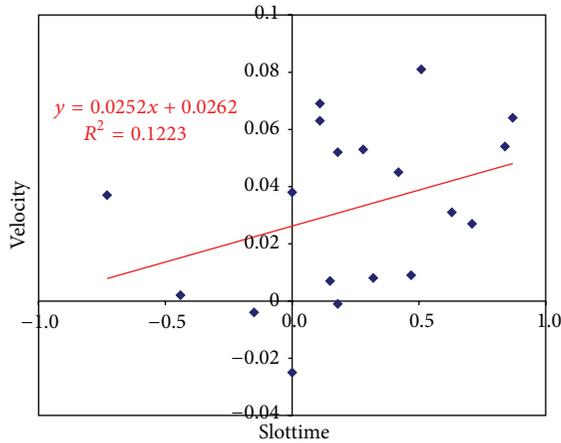


FIGURE 5: Correlation between Slottime and velocity.

in the patient's motor performance during the training. This approach was pioneering but it fully depended on the design of the robot used in the study which enabled the recording of and easy access to several interaction parameters; indeed, we were able to specify the type of assistance provided to the patient by the robot, showing that interactions were multimodal. The robot recorded kinematic data (mainly velocity and position) and, through the use of a performance-based algorithm, applied forces laterally, which guided the hand to assist aiming (Stiffness), as well as "longitudinally" to facilitate reaching to the target (Robotic Power). An additional and intricate parameter took into account the time taken to reach the target in order to challenge the patient to perform the task even faster (Slottime). These parameters were used to create both assistive and challenging effects. The results demonstrated that patient/robot interactions changed "positively" over the training period, following dynamic and differentiated processes. In fact, the velocity-related parameter (Slottime) decreased earlier than the lateral guidance parameter (Stiffness). The decrease in Robotic Power was smaller. This latter finding could be explained by the characteristics of the patients and the duration of the training. Indeed, the patients all had moderate to severe motor impairments and it was possible that the 5 weeks of training were insufficient for significant reductions in this type of robotic assistance to occur.

This study also showed that changes in the interaction parameters were well correlated with clinically evaluated motor performance and certain kinematic parameters (velocity). These results might indicate that these parameters could be reliable indicators of objective motor performance. This is novel because no other study has used such a correlation approach to evaluate interaction.

Another finding is that, above providing movement support and modeling the time course of changes in motor performance, the robot challenged the patient, thus promoting further motor improvement (by decreasing lateral guidance and the time allocated to perform the task). This result is promising as it relates to the need to design robotic algorithms which take into account the fact that the human motor

control system might reduce its participation when the controller is too compliant [25]. The present results suggest that robotic rehabilitation based on an adaptive program optimized the participation of patients, including those with severe motor impairment, thus resulting in a potential enhancement of experience-dependent plasticity.

As the patients included in this study were in the subacute phase of stroke, these results might provide an insight into the process of motor recovery in the early stages after stroke. In fact, the results suggest that movement velocity recovered before accuracy. This is concordant with previous work [38, 39]. The negative correlation between lateral guidance and the velocity-related parameter also suggested that there may be a speed/accuracy tradeoff in motor performance.

The study also highlighted that robotic devices could provide a compliment to clinical scales by quantifying motor performance, including longitudinal measurements of active participation. This latter parameter is useful to gain an understanding of the recovery process, besides motion kinematics which are typically evaluated in robotic studies [40, 41]. Even if the discriminant validity of kinematic variables as measures of UL impairment is still unclear, it appears that kinematic assessments extend clinical scales, assessing sensorimotor function in a more objective and reliable way and in repeatable conditions [42, 43].

The results reported should be interpreted with caution because the sample was small and the study was observational. Other limitations include the fact that data were incomplete for some patients.

## 5. Conclusions

This study used a novel approach to evaluate motor performance in the subacute phase of stroke. The results demonstrated that patient/robot interaction parameters are valid, relevant, and reliable variables for the quantification of active participation and motor performance in subacute stroke patients. Indeed, these parameters were correlated with clinical scores and with some kinematic parameters. These findings suggested that the analysis of patient/robot interactions might provide new insights into the motor recovery process.

## Conflict of Interests

The authors have no conflict of interests to declare or any financial or other interests in the manufacturer or distributor of the device used in the present study.

## Acknowledgments

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## Research Article

# A Game System for Cognitive Rehabilitation

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Brain injury such as traumatic brain injury (TBI) and stroke is the major cause of long-term disabilities in many countries. The increasing rate of brain damaged victims and the heterogeneity of impairments decrease rehabilitation effectiveness and competence resulting in higher cost of rehabilitation treatment. On the other hand, traditional rehabilitation exercises are boring, thus leading patients to neglect the prescribed exercises required for recovery. Therefore, we propose game-based approach to address these problems. This paper presents a rehabilitation gaming system (RGS) for cognitive rehabilitation. The RGS is developed based on a proposed conceptual framework which has also been presented in this paper.

## 1. Introduction

Brain injury such as traumatic brain injury (TBI) and stroke is the leading cause of long-term disabilities in many countries. Cognitive impairments occurring after sustaining a brain injury include attention, memory, and executive function deficiencies. These consequences dramatically affect patients' lives and limit the performance of their everyday activities [1].

There are many issues which are involved in decreasing rehabilitation effectiveness and competence. The increasing rate of brain damaged victims results in limited human resources and facilities, thus burdening the healthcare systems; for example, in the United States alone, 7% of the population or approximately 20 million people are suffering from cognitive disabilities [2]. In addition, the heterogeneity of impairments that patients suffer from is a relevant factor for planning, developing, and evaluating treatments. Hence, tailoring individualized rehabilitation tasks required for each patient involve a higher cost. The health care costs for patients after sustaining brain injuries are among the highest compared with other healthcare services in many countries [3].

On the other hand, studies revealed that majority of patients (75%) with traumatic brain injuries are less than 35 years in age [4], and this age group is more inclined to play games on computers and/or handheld games devices

compared to other old age groups. Current research findings revealed that the brain has the ability to cure itself following an injury through repetitive, intensive, and task oriented training [5, 6]. However, brain damaged patients commonly reported that traditional rehabilitation exercises can be boring due to their repetitive nature which further lead them to neglect the exercises required for recovery [7]. In addition, patient's motivation is found to be an important factor for rehabilitation success and is often utilized as a determining factor in the outcome of rehabilitation [8]. The therapists' main problem is to find a way to encourage patients to actively take part in a rehabilitation program [9, 10]. Therefore, with such substantial effects on the quality of life of millions of patients and on healthcare systems worldwide, a feasible game-based intervention that can increase rehabilitation adequacy and effectiveness is crucial. This paper presents a rehabilitation gaming system (RGS) that would address these issues. The RGS is developed based on a proposed conceptual framework described in the following sections.

## 2. Proposed Conceptual Framework

Research on serious games for people with cognitive disabilities is still in its infancy, compared to other types of disabilities [11]. The proposed conceptual framework for designing a brain injury cognitive rehabilitation gaming

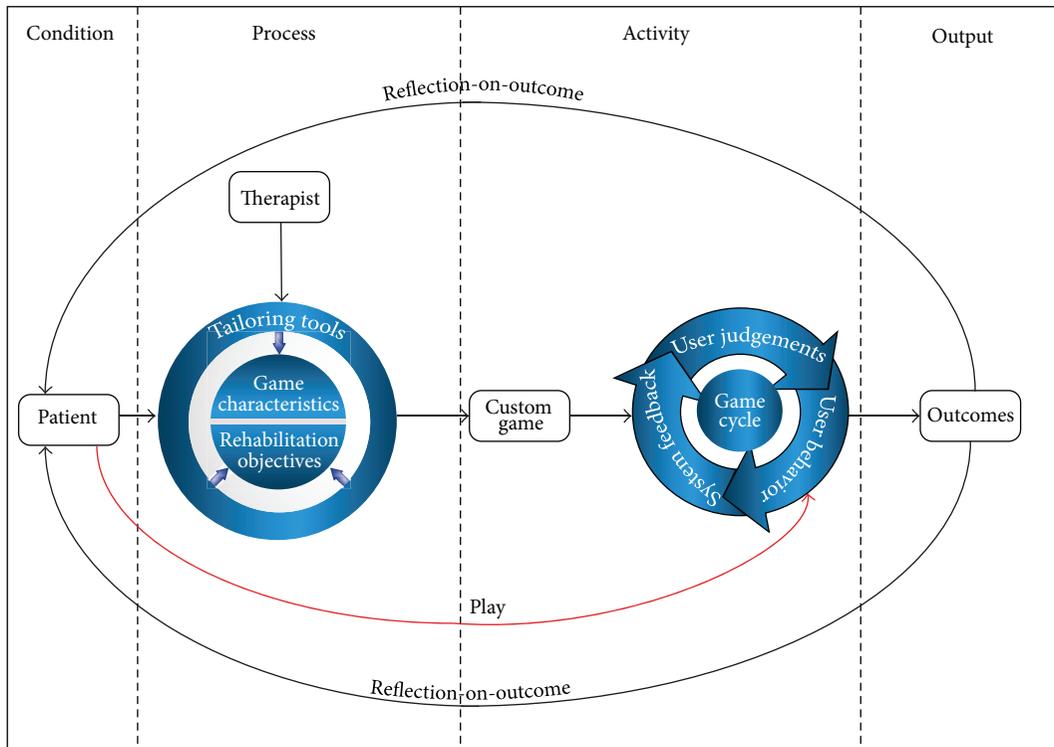


FIGURE 1: A conceptual framework for designing brain injury cognitive rehabilitation game.

system is illustrated in Figure 1. The aim is to establish a framework that can be used by the game developers and practitioners while designing game systems for cognitive rehabilitation. The framework was constructed from the results of our investigation and other related literatures [12–14]. Basically, the proposed framework consists of four components, namely, condition, process, activity, and outcome. Each one of these parts plays a crucial role in designing feasible game-based cognitive rehabilitation intervention that can increase rehabilitation effectiveness and competence. The following sections describe these components in detail.

**2.1. Condition.** The main factor that causes mismatch between a patient and potential technology is inadequate evaluation and assessment of the patient’s needs and preferences [1, 15]. The difficulty of matching a patient and technology emerges not only from his unique combination of cognitive, sensory, and physical abilities but also from his expectations and reactions to the new technology interventions. These reactions arise from personal preferences, needs, abilities, and prior technology experiences [15].

Cognitive assessment practice typically begins with simple tests such as the Mini Mental State Exam (MMSE). In case of MMSE poor performance, it is fundamental to conduct a more in-depth evaluation using the neuropsychological assessment (NA) [16]. In addition, investigation of personal and health related factors as well as environment may help in the identification of factors that influence patient’s functional ability and recovery [17]. The International Classification of Functioning, Disability and Health (ICF) model of human

functioning and disability focuses on the “components of health” instead of the “consequences of disease” [18]. The ICF can be used as a valuable framework to manage and classify patient’s information and it helps rehabilitation professionals in identifying aspects of a patient’s condition that affect his/her recovery [19]. Therefore, game-based rehabilitation interventions are most effective when they are shaped to meet a patient’s particular needs and preferences, not when they are prescribed as an isolated intervention of addressing particular cognitive and/or physical deficiencies. This process needs to begin with a comprehensive evaluation that gives the rehabilitation professionals an opportunity to assess the patient’s ability, needs, preferences, and expectations. The evaluation generates information that can assist in tailoring game interventions which further enhance the quality of training and positively affect the self-esteem and motivation of patients.

**2.2. Process.** Garris et al. [20] described game characteristics in terms of six categories: fantasy, rules/goals, sensory stimuli, challenge, mystery, and control. They further argued that success in pairing game’s characteristics with appropriate instructional practices could trigger individuals’ motivational forces towards achieving the intended outcome. In our proposed framework as shown in Figure 1, the “tailoring tools” are responsible for mapping a game’s characteristics with the intended rehabilitation objectives. For example, a “challenge” is a crucial game characteristic. Providing optimal challenge for a specific patient means matching the game difficulty to the patient’s ability; being that it is neither too

easy nor too difficult. To enable the optimal challenge, it is necessary to continuously adapt the game or create new game levels using “tailoring tools” to match the patient’s existing skills.

Although a fully automated rehabilitation intervention which assesses the patient’s deficiencies and then uses this assessment to create an individualized rehabilitation plan is virtually possible, such intervention has low chances to be medically accepted [21]. Rehabilitation professionals’ experience in formulating and determining rehabilitation objectives and selecting exercises and facilities to attain those objectives has to be recognized. However, compromising between a fully automated rehabilitation intervention and a fully therapist-dependent rehabilitation intervention is probably the best alternative [21].

In a physical rehabilitation context, although some researchers have built games to address particular deficits, these customized games take time to create, are expensive, and cater only a small number of brain damaged patients. Environments with authoring tools that decrease time and expense enable therapists to quickly create games tailored to individuals with brain injury [22]. However, in cognitive rehabilitation, where there is diversity and heterogeneity of cognitive impairments, environments with “authoring tools” to create customizable games can be more cost-effective and can provide feasible games that can meet the specific needs of brain injured individuals.

The complexity of a game environment depends on its “tailoring tools” and ability of these tools to map the intended rehabilitation objectives with various game characteristics based on the needs and preferences of the patient. However, rehabilitation professionals often do not possess advanced knowledge and skills to understand the underlying design and development. Therefore, the game environment and its tailoring tools should be intuitive enough without the need of much technical knowledge.

**2.3. Activity.** A custom game is the output of the process part as shown in Figure 1, whereby the game is ready to be played by the patient. Retention of patient attention and his/her deep involvement depends on the effectiveness of the tailoring of these game activities by therapists. If the therapist succeeds in mapping the game’s characteristics with the intended rehabilitation goal in the game, this will produce a repeating game cycle [20, 23]. The game cycle may help in sustaining patient’s engagement in the rehabilitation intervention, which in turn leads to specific cognitive and affective outcomes.

**2.4. Output.** The game play activities generate specific outcomes which tell the level of patient achievement in playing the game. This achievement can be as simple as describing the game scores such as the total amount of assets collected and the time taken to achieve the goal within the game; or it can be extended to describe “changes of patient’s outcomes,” which involves measuring improvements in a given cognitive function over time. This achievement can serve the purpose of patient assessment. Hence, new game playing activities should be modified and adapted to suit

the patient’s level. Therefore, outcomes play a crucial role and can be used as monitoring and tracking mechanisms by therapists. This promotes the possibility of unsupervised rehabilitation that can be continued after the patient is discharged from inpatient rehabilitation services. Thus, patients do not need long instructions and supervision by therapists.

Reflections on outcomes: motivation and engagement in game-based training will be achieved if the patient believes in potential success during game play. This perception strengthens the patient’s confidence and can be an incentive for him/her to exert more effort to attain the intended game goal. This can be reflected through outcomes that reflect his/her performance in game experience. Moreover, outcomes enable therapists to capture changes in patients’ skills, what they are able to do, their level of task performance, and affective reactions. Therefore, reflection on outcome guides therapists to continuously adjust and modify the game according to the patient’s existing skills and expectations through therapist-oriented tailoring tools offered by the game environment. To demonstrate the implementation of the proposed framework, a rehabilitation gaming system (RGS) prototype has been developed and described in the next sections.

### 3. Prototype Development

**3.1. Design Considerations.** There are pressing needs for new strategies to increase capacity while optimizing the quality of rehabilitation care. Therefore, the proposed framework for designing game-based intervention for brain injuries cognitive rehabilitation was implemented according to the following design strategy. The first consideration for the prototype development process was to use the web. Development of a web-based rehabilitation platform is a very promising foreseeable solution. Web-based platform can be used in internal networks as well as long-distance public networks for remote patient access from home at no cost. Web is a cross-platform environment that can be accessed from both desktop and mobile platforms regardless of the operating system used. Such systems could be utilized in the clinical setting for inpatient and outpatient rehabilitation. The second consideration for the development process was to use Adobe Flash. This software is a multimedia platform that is well known for creating simulation and games that can be easily viewed on the web, and this is considered as one of the positive factors for using this software in the development process. A negative aspect of using Adobe Flash is the complexity of using the Action Script programming language to create the game system; this was one of the major challenges in this study, which resulted in the time-consuming development process.

**3.2. Prototype Overview.** The main interface of the RGS is shown in Figure 2. The user (i.e., therapist and/or patient) can get into the next level only after filling in the user’s name and password fields. In the next sections, therapist modules are described followed by patient’s game interfaces.

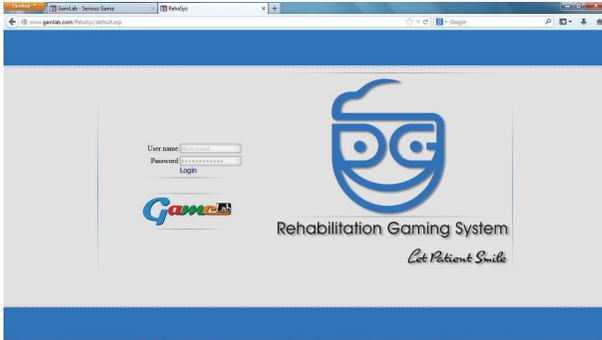


FIGURE 2: Rehabilitation gaming system main interface.

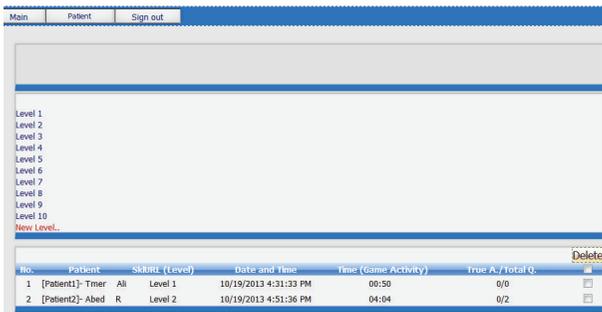


FIGURE 3: Screenshot of therapist's main interface.

**3.2.1. Therapist.** Once the therapist logs into the RGS system, the therapist's main interface will appear as shown in Figure 3. This interface can be described as follows: at the top of the screen there are three buttons ("main," "patient," and "sign out"). The "main button" allows therapist to go back to the main interface. The "patient button" allows therapist to add new patient and access and modify patients' information (details in Section (a)). Clicking on the button marked "sign out" exits the user from RGS.

The middle of the screen shows the list of game's levels that were previously created by the therapist. Therapists can edit and modify them by clicking on the game's level name or by clicking on "new level button" to create a new game level (details in Section (b)). The bottom of the screen displays a patient's results. Once the patient finishes the game exercises, therapists can log into the system and see the details such as patient's name, the levels that were played by the patient, the time and date when the patient accessed the game system, and the total amount of time taken by the patient to finish the game level, and the total correct answers are also presented. Therefore, therapists can easily track patient's performance online, which enables him/her to adjust and modify the game's tasks.

*(a) RGS Patient Editor.* As shown in Figure 4, the RGS patient editor allows therapists to add new patient's information such as user name, password, first name, last name, email, phone number, gender, and case description.

On top of that, RGS patient editor allows therapist to assign game levels to the patients. Also, it allows therapist to search for certain patients through selecting one

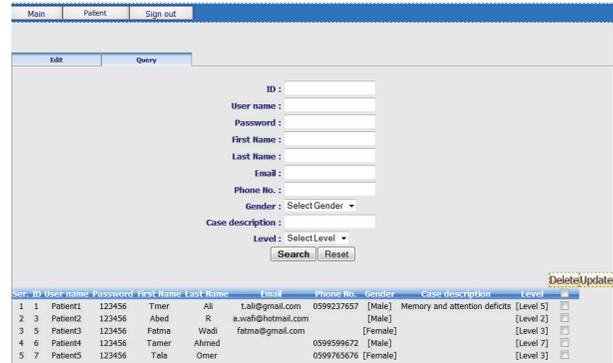


FIGURE 4: Screenshot of RGS patient editor.

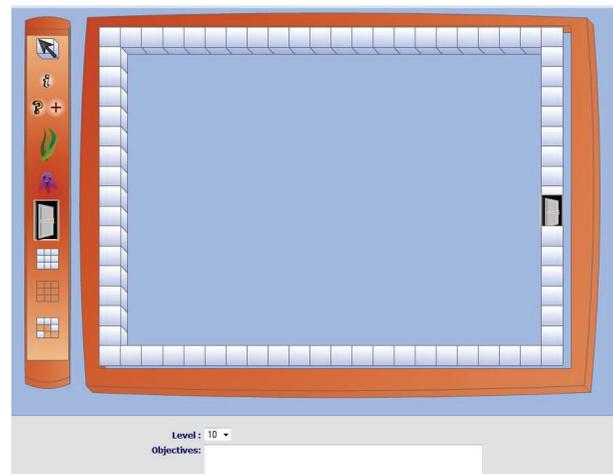


FIGURE 5: Screenshot of game's design environment.

option/criteria and clicking on the button marked "search." The system will then generate a query, which would consult the RGS database for the desired information. A list of patients' details will be presented, so therapists can just click on the "check box" that appears beside each patient's record, allowing the therapist to "delete" or "update" patient's information. In the context of this research, at the end, the RGS will be integrated with the management information system of the rehabilitation hospital. Therefore, patient's information will be captured from hospital information system. However, this module (i.e., create and modify patient profile) was purposely developed so that RGS could be used as a standalone system. Moreover, the generated game's interfaces for the patient are textless; also therapists have the ability to use any language according to their patients; hence the RGS can be used worldwide. Furthermore, it offers a new feasible and cost-effective alternative for rehabilitation.

*(b) Game Design Editor.* There is no limit to the number of mazes that therapists can create. As shown in Figure 5, the game design environment lets therapists create the game and save it. To the right is an empty field surrounded by a border. On the left, there are tools that can be used by therapists to build and tailor the game within the field on the right.

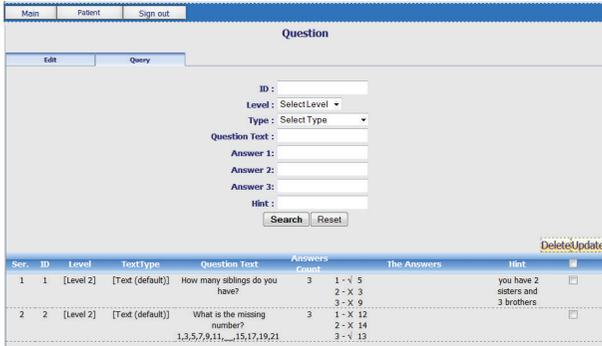


FIGURE 6: Screenshot of RGS questions editor.

Therapists can simply click on a particular tool to activate it and then start the game design process on the empty field. In case of error, like putting the wrong game object in the wrong place, the therapist can simply select the correct one from the “tools” panel and drop it over the wrong one. The game object will instantly turn into the correct choice. At the bottom part of the game design environment, there is a text box and a save button. Therapists can enter texts in the text box describing the objectives of the game and the instructions on how to play the game. When patients log into RGS to play the assigned game, these texts will be launched as a game introductory. In the end, after the therapist completes the design, he/she can simply click on the button marked “save” and the final result will be saved.

(c) *RGS Questions Editor.* As shown in Figure 6, the RGS questions editor enables therapists to create customized questions with different difficulty levels. Patients will have to answer a number of questions. The questions can be about patient’s background such as “How many siblings do you have?” and/or general knowledge questions such as “How many days are there in a week?” and/or mathematical equations such as “ $5 - 2 = ?$ ” Each question has three possible answers. By clicking on one choice of answers, the game will provide feedback to the patient about his answer. The complexity and the number of questions for patients depend upon the game level designed by the therapist. Patients’ answers will allow therapists to analyze the memory and/or cognitive progressions of the patient.

(d) *Creating a New Game Level.* With patient assessment, it is possible to “prescribe” training that targets specific cognitive functions. Therapists, based on their patients’ abilities, limitations, and preferences, create this game training. As shown in Figure 7, therapists can access the RGS design environment and start the design process by using the tailoring tools, drawing the maze’s pathway, adding the game’s objects, identifying the behavior of the opponents, such as how they move or react during game play, and editing and adding questions to the playing field.

3.2.2. *Patient Game Interfaces.* Once a patient’s profile is created and the game training is tailored and assigned by the therapist, the patient will then have access to a personalized

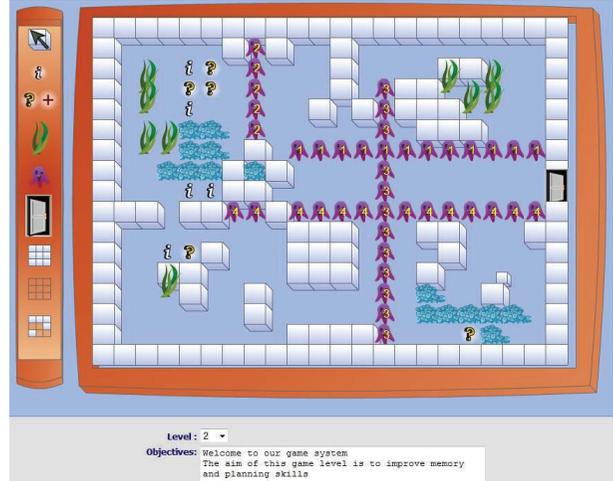


FIGURE 7: Screenshot of game level design by therapist.



FIGURE 8: Screenshot of introductory screen.

set of game tasks. Patients will have to log into the system. This can be made by either patients or caregivers for those who have interaction difficulties. A short introductory screen first will appear to explain the game’s purpose and/or how it is played as shown in Figure 8. Patients can then start the game, as shown in Figure 9, by clicking on the “green arrow button” at the bottom of the screen.

Patients control the yellow smiley face character using the keyboard’s arrow keys and steer it from the upper left corner of the playing field to the destination of a maze (i.e., the opened door). Unfortunately, there are opponents blocking the way which are continuously moving. Players need to avoid collision with them to reach the destination. These restrictions force patients to watch out while walking and they must plan their pathway before moving to the maze’s destination. Whenever the smiley face hits one opponent while moving, it will fly and then automatically move back to the playing field. The length of time taken to reach the destination is recorded. The quicker the player can reach the destination, the better the results he/she can achieve.

Moreover, while patients go through the maze toward the destination, he/she will be asked to answer some customized questions. There are a number of question marks on the playing field. Once the smiley face hits a question mark, it



FIGURE 9: Screenshot of patient's gameplay interface.

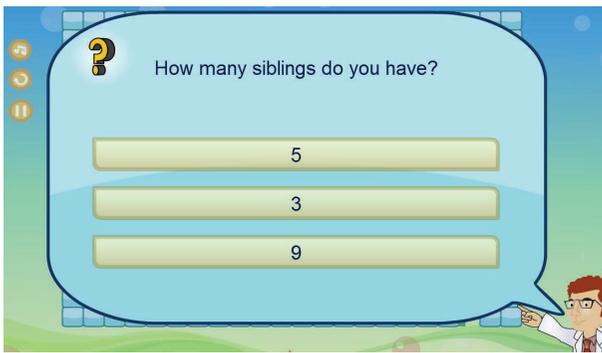


FIGURE 10: Screenshot of game's question interface.

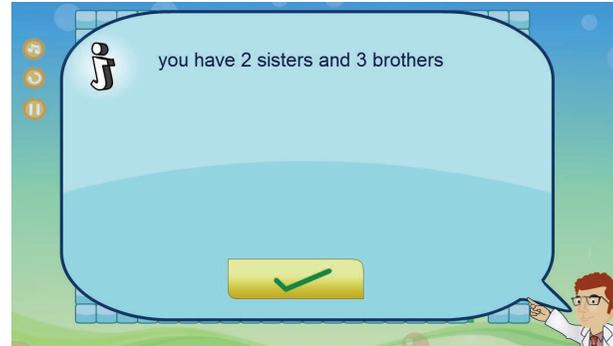


FIGURE 11: Screenshot of helping aids interface.



FIGURE 12: Screenshot of patient's score interface.

will disappear and the question will be presented via a pop-up screen as shown in Figure 10. The patient will have three choices of answers: when he/she thinks that he/she has found the answer, he/she can click on it and he/she will be given a feedback telling him/her if he/she has chosen correctly. Then, the patient can press the space bar or simply click the "arrow button" to continue playing. In addition, help icons labeled "i" are available on the playing field; patients can use them to answer the questions as shown in Figure 11.

Outside the maze, in the upper left corner, three buttons are provided for the patient. The first one allows patients to replay the game level. The second button can pause the game and the last button can stop the game's music. Patients continue until they reach the maze destination (i.e., when the smiley face touches the opened door). The score will be calculated based upon the total time taken to finish the game level and the number of questions the patient has answered correctly.

The result will be presented as shown in Figure 12 and automatically saved. The patient then will be given the opportunity to go to the next game level by clicking on the "arrow button."

#### 4. Conclusion

In this paper, a conceptual framework for designing game-based cognitive rehabilitation system is presented. This framework can be a useful guide for game designers, developers, and practitioners in designing a rehabilitation gaming

system that can significantly affect patients, therapists, and health care systems. Every component of this framework plays a crucial role to provide game-based intervention that can increase rehabilitation effectiveness and competence. To demonstrate the implementation of the framework, a rehabilitation gaming system (RGS) is developed. RGS benefits rehabilitation process and fulfills its real needs such as increasing patient's motivation by providing individualized rehabilitation game experience while simultaneously reducing the development costs associated with it and allowing therapists to track patient's activities and to assess their progress. Furthermore, it is likely to open new opportunities for home-based and unsupervised rehabilitation. The question arises on whether such interventions will be accepted by the target group (i.e., patient and therapist). In the near future, evaluation of the developed rehabilitation gaming system targeting therapists will be conducted to determine their satisfaction.

#### 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

# Grasps Recognition and Evaluation of Stroke Patients for Supporting Rehabilitation Therapy

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Stroke survivors often suffer impairments on their wrist and hand. Robot-mediated rehabilitation techniques have been proposed as a way to enhance conventional therapy, based on intensive repeated movements. Amongst the set of activities of daily living, grasping is one of the most recurrent. Our aim is to incorporate the detection of grasps in the machine-mediated rehabilitation framework so that they can be incorporated into interactive therapeutic games. In this study, we developed and tested a method based on support vector machines for recognizing various grasp postures wearing a passive exoskeleton for hand and wrist rehabilitation after stroke. The experiment was conducted with ten healthy subjects and eight stroke patients performing the grasping gestures. The method was tested in terms of accuracy and robustness with respect to intersubjects' variability and differences between different grasps. Our results show reliable recognition while also indicating that the recognition accuracy can be used to assess the patients' ability to consistently repeat the gestures. Additionally, a grasp quality measure was proposed to measure the capabilities of the stroke patients to perform grasp postures in a similar way than healthy people. These two measures can be potentially used as complementary measures to other upper limb motion tests.

## 1. Introduction

Stroke survivors are often unable to perform the fine motor control activities which are required during activities of daily living, amongst which grasping is one of the most recurrent. Robots represent an appealing tool for exercise-based approach in neurorehabilitation due to their ability to deliver repetitive training.

Previous studies indicate that the inclusion of robot rehabilitation training improves short- and long-term motor control of the impaired upper limb of patients after a stroke [1, 2]. However, evidence of the transfer of robotic training effects to activities in daily life is limited [3]. Therefore, the inclusion of functional tasks, such as grasping objects, is vital to increase the practice and improvement in such activities for stroke rehabilitation [4, 5]. A next step in this direction

is to incorporate the detection of various grasps in the robot rehabilitation frameworks and to evaluate their quality in order to detect any improvement in the rehabilitation process.

The SCRIPT (Supervised Care & Rehabilitation Involving Personal Telerobotics) project aims at delivering an affordable system for home-based rehabilitation of the hand and wrist for stroke survivors [6]. A passive exoskeleton (Figure 1) has been developed within the project in order to facilitate the patients' fingers and wrist extension. Self-administered training at home is performed through repetitive interaction with games based on functional exercises (rehabilitation games) to enhance engagement. The games are controlled by wearing the orthosis and performing several arm and hand movements. The challenge is to use the limited set of sensors provided by the device to recognize the various grasp postures and incorporate them into the games.

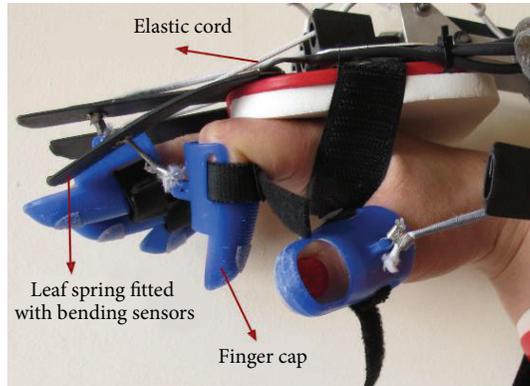


FIGURE 1: SCRIPT passive orthosis showing the bending sensors and leaf springs.

In previous work [7], we implemented and tested different methods for recognizing four different grasp postures performed while wearing the SCRIPT orthosis. We found that with the support vector machines (SVM) method, we could achieve an overall accuracy of more than 90% with small computational time. SVM has been successfully applied to the classification of a variety of biomedical conditions [8–10] and more specifically to study different aspects of stroke patients, including the classification of carotid artery plaques [11], the study of dietary patterns [12], or using readings of a shoe-based sensor to identify sitting, standing, and walking postures [13]. However, few studies have been reported on the classification of grasp postures. Tavakolan et al. [14] used SVM for pattern recognition of surface electromyography signals of four forearm muscles in order to classify eight hand gestures. On the other hand, Puthenveetil et al. [15] used linear discriminant analysis to classify hand preshapes in poststroke patients using data from the CyberGlove. In this study, the SVM approach was selected to perform the gesture recognition.

We conducted an investigation in a group of eight poststroke patients to determine the accuracy of the gesture recognition wearing the SCRIPT device using the SVM approach and compared it with the samples obtained for a group of ten healthy subjects, and their relationship to the level of hand impairment. The goal was to assess the grasp recognition with this particular user group, in order to confirm its suitability prior to deployment in a larger scale clinical evaluation of the prototype orthotic device and the SCRIPT system. Successful grasp recognition will provide a more versatile set of gestures to the therapeutic human-machine interaction system that acts as a medium to support home-based rehabilitation. After a given grasp posture is recognized, the aim is to evaluate the quality of the grasp in order to measure the patient progress throughout the rehabilitation process. In this study, we propose a grasp quality measure that can be calculated with the available sensor readings of the orthosis. These measures are compared with the results of other commonly used upper limb motion tests such as the action research arm test (ARAT) [16].

TABLE 1: Details of healthy participants.

Id	Country	Age	Gender	Dominant
1	UK	27	Male	Right
2	UK	32	Male	Right
3	UK	33	Male	Right
4	UK	30	Female	Right
5	UK	32	Female	Right
6	Italy	47	Female	Left
7	Italy	50	Male	Right
8	Italy	52	Female	Right
9	Italy	50	Male	Left
10	Italy	51	Female	Left

## 2. Materials and Methods

**2.1. Device.** The SCRIPT passive exoskeleton [17] is a passive device which applies external extension torques on the fingers through five leaf springs which are connected to the finger caps through an elastic cord (Figure 1). The elastic cord enables the fingers freedom of movement relative to the leaf spring and also allows adjusting the level of extension support provided by the device. The leaf springs are fitted with commercial resistive flex sensors [18] which measure their flexion with a resolution of 1 degree. However, the deflection of the leaf spring is not the actual flexion angle of the finger but the two quantities are related by a monotonically increasing function [17]. Subjects are free to laterally abduct/adduct the fingers and oppose the thumb, but these movements are not supported nor sensed by the orthosis. The orthosis measures only overall finger flexion in a range from 0 to 90 degrees and wrist flexion and extension in a range from 90 to  $-45$  degrees.

It is important to note that this passive orthosis, as many of those in the field of rehabilitation robotics, may not conform to the conventional definition of a robot, although consisting of sensors, passive actuators, and a decision making component. However, our research is applicable in the field of robot-assisted training and rehabilitation robotics, where sensors are utilized towards benchmarking motor performance aiding a meaningful two-way interaction between the human and the machine.

**2.2. Study Participants.** Two groups of participants were recruited for this study: a group of healthy subjects and a group of stroke patients. In the first group, ten healthy subjects with no previous injuries of fingers, hand, or wrist volunteered to participate in this study (Table 1). This study was carried out at the University of Hertfordshire, UK, approved by the university ethics committee (Ethics protocol number COM/ST/UH/00008) and at the IRCCS San Raffaele Pisana (Rome, Italy). Participants were recruited among faculty staff by advertising on an internal mailing list.

The group of stroke patients were selected with the criteria used in the SCRIPT project in order to enable them the use of the orthosis [19]. They were patients with a unilateral ischemic or hemorrhagic stroke between 6 months and 5 years ago. They had limitations in arm and hand function,

while being able to actively flex the elbow by at least  $15^\circ$  and to actively flex their fingers by a quarter of their passive range. They also had the ability to understand and follow instructions. For stroke patients, individually fitted orthoses were used. With these criteria, a total of eight patients were selected (Table 2). One patient was recruited at the Rehabilitation Centre Het Roessingh (Enschede, the Netherlands) and seven at the IRCCS San Raffaele Pisana (Rome, Italy). In both cases, the experiment was part of an on-going clinical study, for which ethical approval had been obtained at the respective local ethics committees.

The action research arm test (ARAT) [16] and Fugl-Meyer assessment (FM) [20, 21] have been used as quantitative measures to evaluate the arm motor recovery after stroke. The ARAT consists of four sections: (A) grasp, (B) grip, (C) pinch, and (D) gross arm movement. The FM assessment for the upper extremity consists on a scale of 66 points divided in three subsections: proximal (shoulder and elbow), distal (wrist and hand), and coordination (test of tremor, dysmetria, and time). Details of the results of these tests for the stroke patients can also be seen in Table 2.

**2.3. Grasp Gestures.** In recent literature, several activities of daily living have been identified as the most important to train after stroke [4, 22, 23]. They include eating with cutlery, drinking, holding objects while walking, taking money from purse, open/close clothing, combing hair, and knob manipulation. We have selected, in agreement with healthcare professionals, three grasp gestures needed in order to perform these tasks (shown in Figure 2). Two are classified as precision grips: the tripod (the thumb opposes the index and middle finger) and the lateral grasp (the thumb holds an object against the side of the index finger). The third one, the cylindrical grasp, is classified as a power grasp (all fingers make contact with the object). Keller et al. [24] identified the tripod and the lateral grasps as the most frequently used prehensile patterns for static and dynamic grasping, respectively. The relaxed posture of the hand was used as the forth gesture in order to enable the recognition when the subjects were not performing any grasp.

Two of the selected gestures are evaluated in the ARAT: the tripod grasp is tested in section A (grasp subtest) and the cylindrical grasp in section B (grip subtest) of ARAT. It is expected that the recognition of the gestures is related to the ability to move the hand measured by ARAT.

**2.4. Grasp Recognition Method.** The problem of hand posture detection has been previously approached using vision-based [25, 26] or glove-based [25, 27–29] methods depending on the constraints of the specific applications. In our case, given the bulk of the device, vision-based approaches for the recognition of the hand postures are not suitable as the hand is practically occluded and therefore the recognition should be based on the sensory readings for each finger provided by the exoskeleton sensors. In a previous work, we compared several glove-based approaches to recognize grasp postures performing experiments on five healthy participants wearing the SCRIPT passive exoskeleton [7]. We compared

three methods: one based on the statistics of the flexion data, another based on neural networks, and finally one based on support vector machines (SVM). We found that with the last method, we could achieve an overall accuracy of more than 90% with small computational time (<60 ms). Therefore, in this study, we used the SVM approach to determine the accuracy of recognition for healthy participants and stroke patients wearing the SCRIPT device.

SVM is a popular machine learning technique for classification. A support vector machine is a supervised learning classifier that constructs a set of hyperplanes in a high-dimensional space (support vectors) that are used to classify the data. A good separation is achieved by the hyperplane that has the largest distance to the nearest training data point of any class. The hyperplanes are found solving the following optimization problem [30]:

$$\min_{\omega, b, \varepsilon} \quad \frac{1}{2} \omega^T \omega + C \sum_{i=1}^l \varepsilon_i \quad (1)$$

subject to  $y_i (\omega^T \varphi(x_i) + b) \geq 1 - \varepsilon_i, \quad \varepsilon_i \geq 0,$

where  $\{(x_i, y_i) \mid x_i \in R^p, y_i \in \{-1, 1\}\}$  are the training set of  $l$  instance-label pairs,  $x_i$  is  $p$ -dimensional real vector,  $\omega$  the normal vector of the hyperplane, and  $C > 0$  the penalty parameter of the error term. The training vectors  $x_i$  are mapped into a  $p$ -dimensional spaces by the function  $\varphi$  and in order to create nonlinear classifiers a kernel function is used. In our work, we used a radial basis function (RBF) as the kernel function, given that it can handle the case when the relation between class labels and attributes is nonlinear [31]. It is defined as

$$K(x_i, x_j) = \exp(-\gamma \|x_i - x_j\|^2), \quad \gamma > 0, \quad (2)$$

where  $\gamma$  is the kernel parameter. Therefore, two parameters are needed:  $C$  and  $\gamma$ . In order to find the best values for these parameters, we used a  $v$ -fold cross-validation technique, dividing the training set for one subject into  $v$  subsets of equal size and testing the classifier on the remaining  $v - 1$  subsets. In this work,  $v$  was taken as 5, the value of the cost parameter  $C$  was varied as  $C = 2^x, x = \{-5, \dots, 5\}$  and the value of the kernel parameter  $\gamma$  was varied as  $\gamma = 2^y, y = \{-4, \dots, 0\}$ . The values that gave the highest validation accuracy were:  $C = 4$  and  $\gamma = 1$ .

The method was implemented in Python using the *LIB-SVM* package (<http://www.csie.ntu.edu.tw/~cjlin/libsvm>). The flexion angles were normalized in the range from 0 to 1 (corresponding to 0 to 90 degrees) and the selected error to stop the training phase was set to 0.001.

**2.5. Grasp Evaluation Method.** In the field of robotics, many grasp quality measures have been developed that allow the comparison of different aspects of the robotic grasp [32]. In [33], the most common robot grasp quality measures have been adapted to the evaluation of the grasp of the human hand. From this set of quality measures, the one proposed by [34], which measures how close a given grasp is to a reference posture, is the only one that can be used with the sensor

TABLE 2: Details of the stroke patients.

Id	Country	Age (yrs.)	Time since acute event	Gender	Affected	Dominant	Stroke type	ARAT (max 57 points)				Total	FM (max 66 points)			Total
								A	B	C	D		Proximal	Distal	Coordination	
1	Holland	69	13 months	Male	Left	Left	Ischemia	18	10	17	8	53	16	4	49	
2	Italy	73	7 months	Female	Left	Right	Ischemia	12	8	12	9	41	12	3	43	
3	Italy	88	6 months	Male	Left	Right	Ischemia	12	8	6	6	32	12	4	43	
4	Italy	83	7 months	Male	Left	Right	Ischemia	6	4	0	3	13	11	3	42	
5	Italy	85	6 months	Male	Right	Right	Ischemia	12	8	12	9	41	12	3	43	
6	Italy	72	6 months	Male	Left	Right	Ischemia	18	12	12	9	51	28	3	54	
7	Italy	81	7 months	Female	Left	Right	Ischemia	18	12	12	9	51	35	3	62	
8	Italy	69	7 months	Male	Left	Right	Ischemia	12	8	6	6	32	26	3	41	

ARAT = arm scores of the action research arm test; FM = arm scores of the Fugl-Meyer motor assessment.



FIGURE 2: Selected gestures to be recognized performed wearing the SCRIPT device.

information provided by the SCRIPT orthosis. This index has been adapted for this study as follows:

$$GQ = 1 - \sum_{i=1}^n \omega_i \left( \frac{y_i - a_i}{R_i} \right)^2, \quad (3)$$

where  $n$  is the number of hand joints,  $\omega_i$  is a weight factor,  $y_i$  is the current finger flexion angle, and  $R_i$  is the joint angle range used to normalize the index calculated as the maximum between the reference posture  $a_i$  and either the upper or lower angle limit. The index has to be maximized, so that the grasp is optimal when all joints are at the reference posture, having a quality measure of one, and it goes to zero when all its joints are at their maximum angle limits.

The reference posture  $a_i$  is taken in this case as the one performed by the healthy subjects. This measure then enables the evaluation of how far is a poststroke patient from performing a grasp in a way similar to a healthy subject. However, the postures obtained from healthy subjects performing each gesture are likely to have variations between subjects, especially when some of the fingers were not playing an active role in the grasp (e.g., the ring and little fingers in the tripod grasp). In order to consider this variance, we have included a weight factor  $\omega_i$  that will give high scores to fingers whose postures have small standard deviation and vice versa. The

corresponding weights for each finger and a given gesture can be calculated as

$$w_{ig} = \frac{\alpha_g}{\sigma_{ig}}, \quad (4)$$

where  $i$  and  $g$  are the given finger and gesture;  $\sigma_{ig}$  is the standard deviation of the finger flexion over all subjects for the given finger and grasp; and  $\alpha_g$  is calculated as

$$\begin{aligned} \alpha_g = & 5 \prod_{i=1}^5 \sigma_{ig} \\ & \times \left( \left( \prod \sigma_{ig} \right)_{i=1,2,3,4} + \left( \prod \sigma_{ig} \right)_{i=1,3,4,5} \right. \\ & + \left( \prod \sigma_{ig} \right)_{i=1,2,4,5} + \left( \prod \sigma_{ig} \right)_{i=1,2,3,5} \\ & \left. + \left( \prod \sigma_{ig} \right)_{i=2,3,4,5} \right)^{-1}. \end{aligned} \quad (5)$$

**2.6. Experimental Protocol.** The participants took part in one session lasting half an hour conducted by a researcher or a therapist. They were asked to wear a SCRIPT passive orthosis on the impaired hand or one of their hands (in the case of healthy participants) while sitting in front of a PC. Subsequently, they were instructed to mimic the picture of

a gesture shown on the screen (Figure 2). The participant then confirmed that he/she achieved the desired gesture and then, pressing a button on the screen, the flexion angles of the gesture were saved. After confirmation, they were asked to relax the hand and press a button. At that moment, the angles of the relaxed posture were also saved.

Each subject performed 6 repetitions of each gesture in a pseudorandom sequence, resulting in the capture of 24 gestures. Data were then postprocessed by Python ad hoc applications. The results of the classification system were calculated using the following values for each gesture  $i$ :

- (i) True positives (TP): gestures correctly classified as gesture  $i$
- (ii) False negatives (FN): gestures  $i$  incorrectly classified as other gestures
- (iii) False positives (FP): other gestures incorrectly classified as gesture  $i$
- (iv) True negatives (TN): other gestures correctly classified as nongesture  $i$ .

These are commonly used to evaluate the sensitivity and specificity of a clinical test in its ability to confirm or refute the presence of a disease. The sensitivity (true positive rate) refers to the ability of the test to correctly identify those patients with the disease and the specificity (true negative rate) refers to the ability to correctly identify those patients without the disease [35]. In this study, the outcome of the test is not binary therefore we calculated the performance measures focusing on each gesture. We used the accuracy (ability to correctly identify positive and negatives), true positive rate (ability to correctly identify the positives), and false positive rate (lack of ability to correctly classify the negatives). The last one is measuring the opposite of the specificity as it enables us to focus on evaluating the recognition performance for a given gesture instead of looking at the classification of other gestures. They are calculated as follows:

$$\begin{aligned} \text{Accuracy} &= \frac{\text{TP} + \text{TN}}{\text{Total testing data}}, \\ \text{True positive rate (TPR)} &= \frac{\text{TP}}{\text{TP} + \text{FN}}, \\ \text{False positive rate (FPR)} &= \frac{\text{FP}}{\text{TN} + \text{FP}}. \end{aligned} \quad (6)$$

**2.7. Data Analysis.** The data acquired for each subject was divided into two sets for training and testing purposes. As the patients will need to perform the training procedure each time before starting the games (or the games will not be able to reliably recognize their gestures), the less number of samples required to train the model the better. In [7] it has been shown that a high accuracy can be achieved with 4 training samples, thus allowing very short calibration time and making this approach suitable for home-based rehabilitation. Then a training set using 4 samples per gesture was used to train the model. Results were considered taking into account all the possible permutations of 4 training samples.

The overall results of the recognition performance are summarized as median and interquartile range (using box plots) differentiated between the participant's type (healthy or stroke patient). The information provided by the different performance measures is compared using a Pearson correlation in order to determine if we can rely on one measurement for the recognition assessment. We also compared the variability of the flexion angles over all subjects performing the different grasp gestures and the results of the accuracy of gesture recognition per participant.

Additionally, the results of the proposed grasp quality measure are also presented summarized using box plots showing the variation between the participant's type, the different gestures and the intervariability between the participants. In order to assess how these three parameters influenced the results of the grasp quality measure, a multiple regression analysis was performed. This type of analysis can be used to consider multiple independent variables calculating the least square estimates for a data set [36]. Using this analysis, we can devise our model using the following equation:

$$\begin{aligned} \text{GQ}_i &= b_0 + b_1 \text{Participant type}_i \\ &+ b_2 \text{Grasp type}_i + b_3 \text{Subject}_i + \varepsilon_i, \end{aligned} \quad (7)$$

where  $b_0$  represents the constant and  $\varepsilon$  is the modelling error. In this case, predictors are classification variables with more than two categories, therefore dummy coding is required to include them in the regression equation. The technique to do this coding is to create an independent variable for each independent category except one as a dichotomy. The omitted variable provides a baseline for comparison while avoiding multicollinearity [36]. In this analysis, we used "healthy participants," "relaxed posture," and "subject 1 (healthy)" as our base line for each one of our predictors. The "Enter" method was used in order to force the model to consider all variables as significant variables in the model.

It is intuitive to assume that the level of impairment of the patients should be correlated with their ability to consistently perform the gestures in a similar way (recognition accuracy) and similar to the ones performed by healthy subjects (GQ measure). As the ARAT and FM tests are common ways to evaluate the level of capabilities of the upper limb, we correlated the accuracies of recognition and the grasp quality with the results of these test using the Pearson coefficient. The IBM SPSS statistical package for Windows version 21.0. was used to perform the analysis of the data.

### 3. Results

This section presents the results of the experiments in two parts: the results of the recognition of the different gestures performed by healthy subjects and stroke patients and the results of the evaluation of those grasps.

**3.1. Grasp Recognition Results.** The results of the true positive rate of the training and testing phases of the experiments are presented in Table 3. The number of gestures correctly

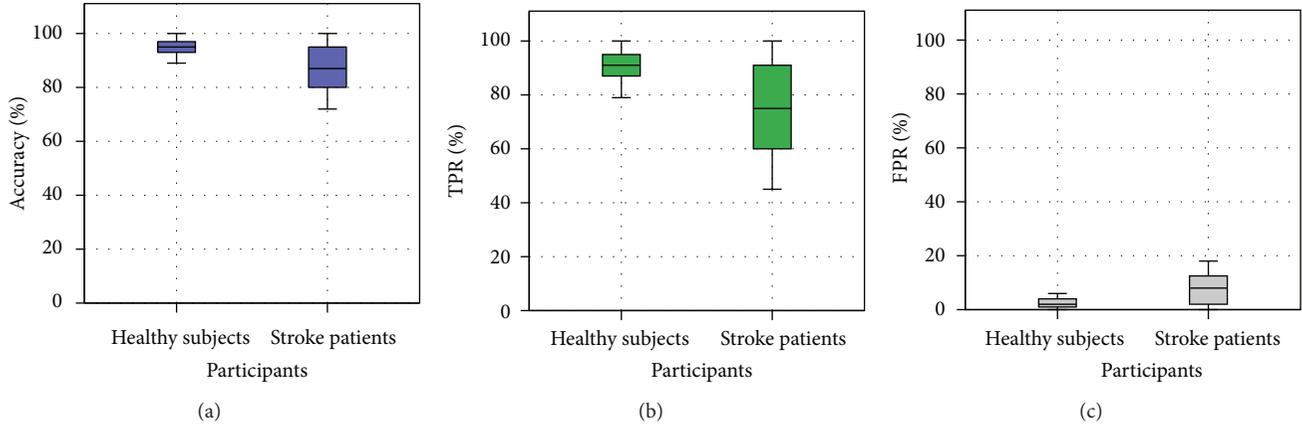


FIGURE 3: Performance measures evaluating the recognition of grasp postures: (a) accuracy, (b) true positive rate and, (c) false positive rate. Results are presented discriminated between healthy and stroke participants.

TABLE 3: Training time and true positive rate (%) between the training and testing phases.

		Mean training time (ms)	Mean training TPR (%)	Mean testing TPR (%)
Healthy subjects	1	1.800 ± 0.6	58 ± 14	87 ± 2
	2	1.733 ± 0.6	100 ± 0	100 ± 0
	3	2.200 ± 0.7	84 ± 14	91 ± 0
	4	1.933 ± 0.5	79 ± 25	98 ± 3
	5	1.733 ± 0.6	84 ± 14	91 ± 0
	6	1.867 ± 0.5	58 ± 10	83 ± 4
	7	2.200 ± 1.4	63 ± 9	85 ± 4
	8	2.267 ± 1.6	98 ± 8	97 ± 4
	9	2.200 ± 0.6	31 ± 9	91 ± 1
	10	1.867 ± 0.5	84 ± 16	90 ± 2
		1.980 ± 0.5	74 ± 24	91 ± 6
Stroke patients	1	2.600 ± 0.7	100 ± 0	99 ± 2
	2	2.200 ± 0.4	61 ± 15	91 ± 4
	3	2.200 ± 0.4	38 ± 6	59 ± 4
	4	1.933 ± 0.7	38 ± 11	56 ± 4
	5	2.467 ± 0.6	38 ± 6	58 ± 4
	6	2.067 ± 0.6	33 ± 6	68 ± 5
	7	2.067 ± 0.3	47 ± 19	93 ± 2
	8	2.133 ± 0.4	43 ± 26	78 ± 3
		2.208 ± 0.6	50 ± 25	75 ± 17

recognized greatly increased from the training to the testing phase. It can clearly be seen that the testing overall recognition performance of gestures performed by the stroke patients is lower than the ones performed by the healthy subjects, as it was expected, but still they produced a high percentage of true positive recognized gestures (TPR mean of 75%). The computational time taken for training the model was on average 1.98 ms for healthy subjects and 2.21 ms for stroke patients.

In order to evaluate in more detail the recognition results, the selected performance measures are shown in Figure 3. Ideally, the accuracy and true positive rate should be close to 100% and the false positive rate close to zero. In this case, the median accuracy of recognition of gestures performed by healthy subjects was over 95% and 87% for stroke patients. The true positive rate showed lower values with respect to the accuracy: median values for healthy subjects of over 91% and for stroke patients of over 75%. The healthy subjects showed median values of false positive rates of 2% and 8% for stroke patients.

The different performance measures can also provide information about the specific recognition of each gesture. Specifically, the true positive rate refers to the ability of the method to correctly identify a specific gesture, whilst the false positive rate refers to the percentage of gestures wrongly classified. Figure 4 shows the accuracy, true positive rate, and false positive rate for each gesture. The difficulty of recognition of the different grasp gestures is different for the healthy subjects or stroke patients. The relax posture is a clearly distinctive gesture, therefore showing the best accuracies, best TPR and FPR values close to zero. For healthy subjects, the tripod and cylindrical gestures were the most difficult to be recognized and the most misrecognized. For stroke patients, the three gestures presented similar difficulties to be recognized and the tripod and lateral grasp were the most misrecognized.

Table 4 presents the Pearson correlation values of the different performance measures for the given conditions. The accuracy is inversely correlated with the false negative rate over all conditions (correlation coefficient  $C < -0.97$  with statistical significance) and also highly correlated with the TPR ( $C > 0.99$ ). The different measures of performance provide specific information on the performance of recognition, but the accuracy could be selected as the overall measure of recognition performance. Therefore, for the following analysis, the accuracy will be used.

It is expected that the grasping capabilities of the participants affect the recognition of hand postures, especially in the case of stroke patients. Therefore, we also studied

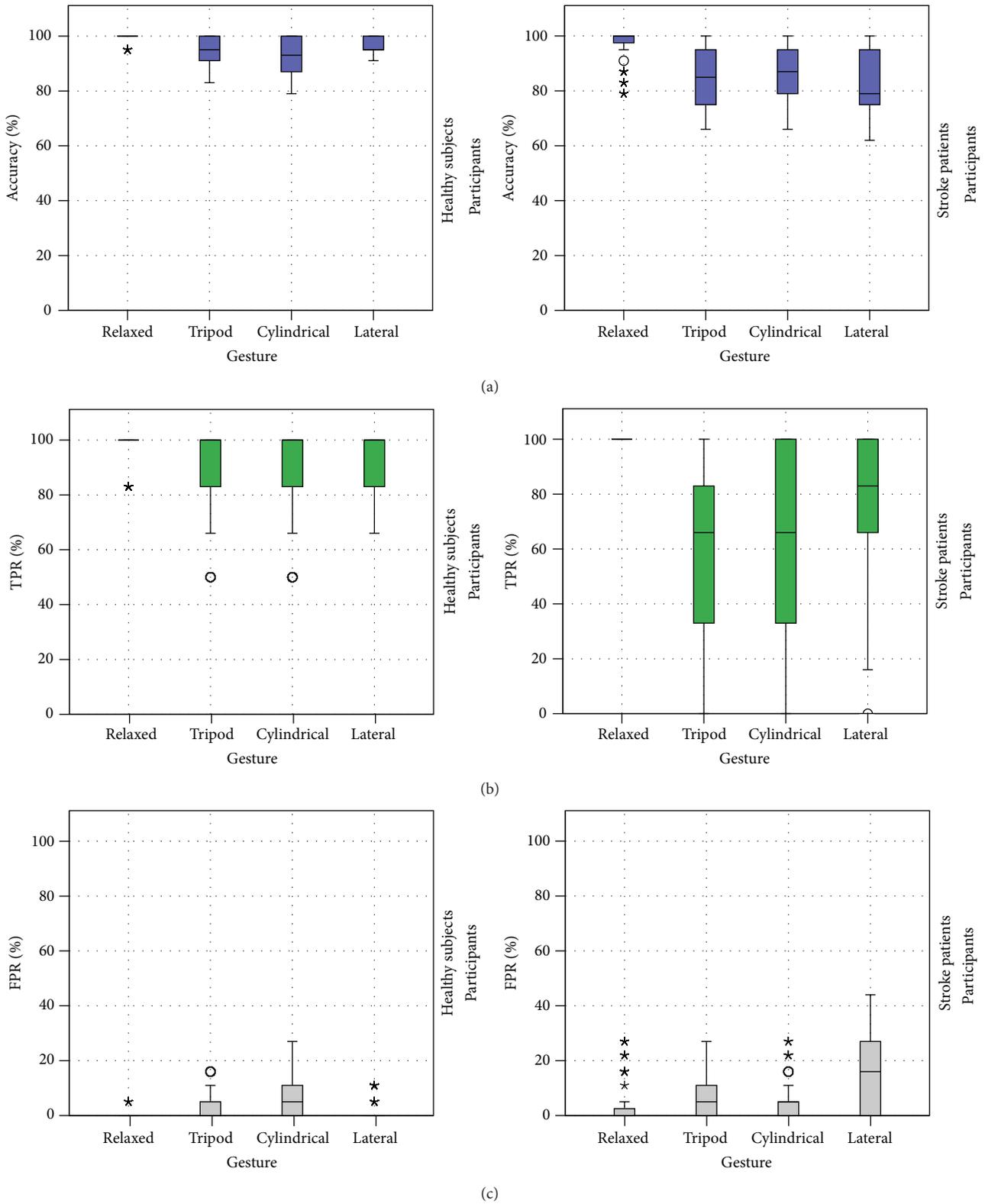


FIGURE 4: Performance measures for each gesture evaluating the recognition of hand postures: (a) accuracy, (b) true positive rate, and (c) false positive rate.

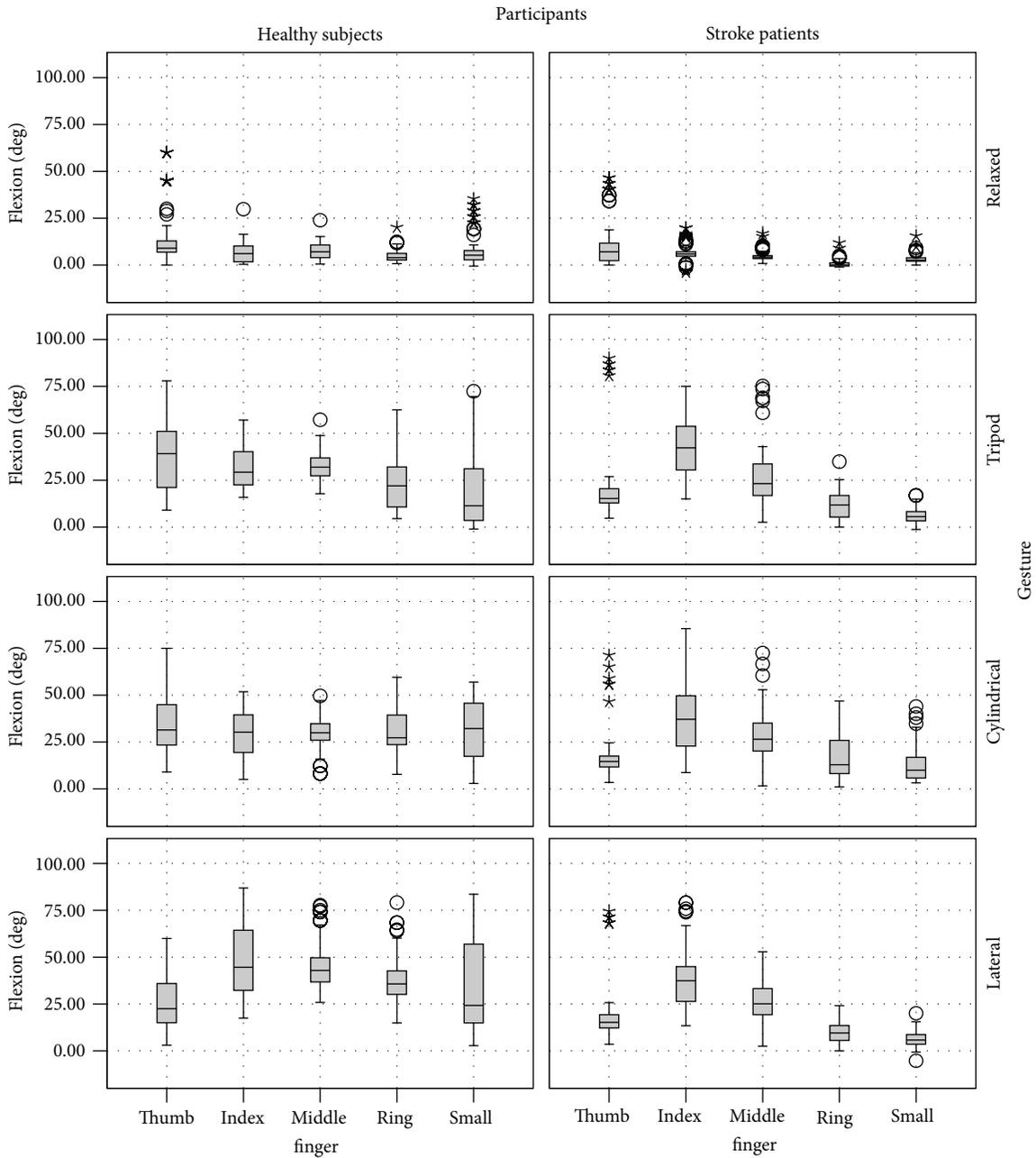


FIGURE 5: Variability of the different finger flexion values produced by all subjects while performing the various gestures.

TABLE 4: Correlation between performance measures.

Performance measures	Participant	Pearson correlation
Accuracy versus TPR	Healthy subject	0.999*
	Stroke patient	1.000*
Accuracy versus FPR	Healthy subject	-0.973*
	Stroke patient	-0.997*

\*Correlation is significant at the 0.01 level (2-tailed).

the variability of the finger flexion to produce the different gestures and their impact on the recognition.

Figure 5 shows a summary of the variability of the flexion angles over all subjects performing the different grasp gestures. As expected, the relaxed posture is the most consistent over all gestures. For healthy subjects, the ring finger and little finger are the ones with greater variation as they can freely move and are not actively participating in the tripod grasp. The ranges of variation for stroke patients are not similar to the healthy subjects. As the patients have different levels of impairment, the flexion of each finger has higher ranges across stroke patients with several outliers, except in the case of the ring finger and little finger where the variation was quite small (max 45 degrees). This might be

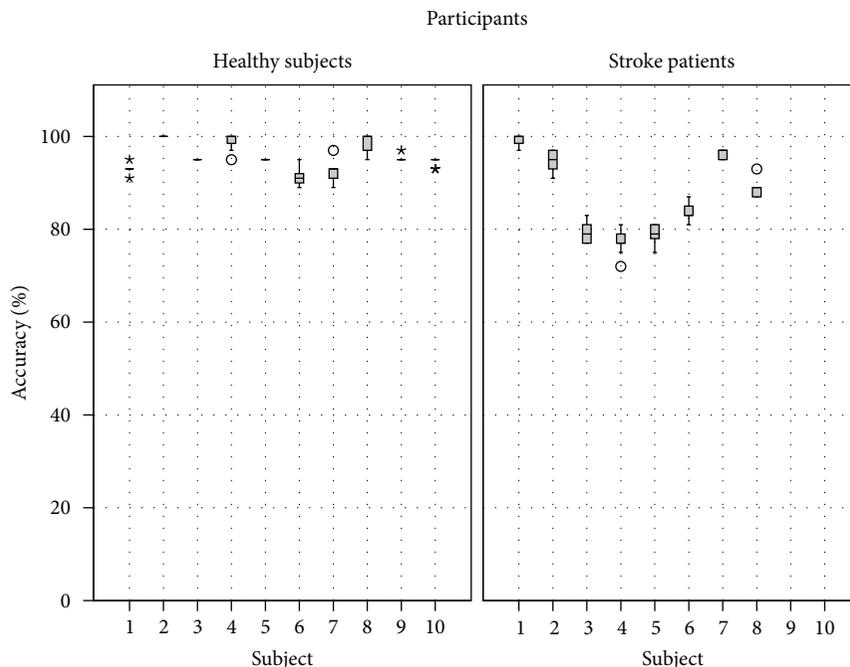


FIGURE 6: Results of the accuracy of gesture recognition per participant. The left graph presents the results for healthy participants and the right one for stroke patients.

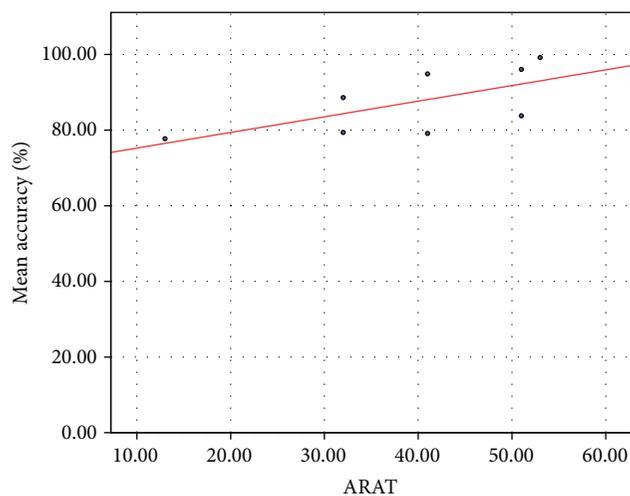


FIGURE 7: Association between accuracy of recognition and the results of the ARAT.

due to the limited mobility of these fingers and therefore the patients had to rely on the thumb, index finger, and middle finger to grasp the objects.

The results of the accuracies of recognition of the different gestures per participant are presented in Figure 6. The variation of accuracies of healthy subjects (between 89 and 100%) was smaller than the ones of stroke patients (between 72 and 100%).

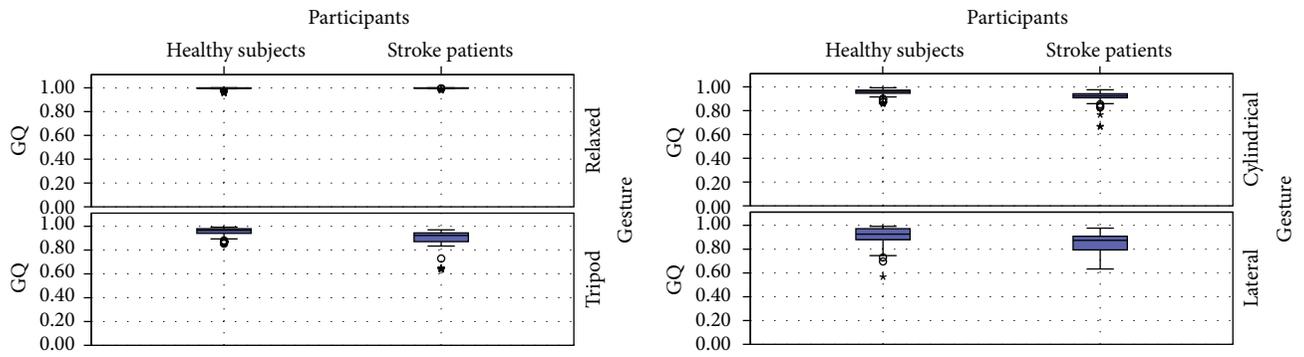
The correlation results of the recognition accuracies with the results of the arm motor recovery tests are presented in Table 5. The higher positive correlations are between the

TABLE 5: Correlation between accuracy and common arm motor recovery tests.

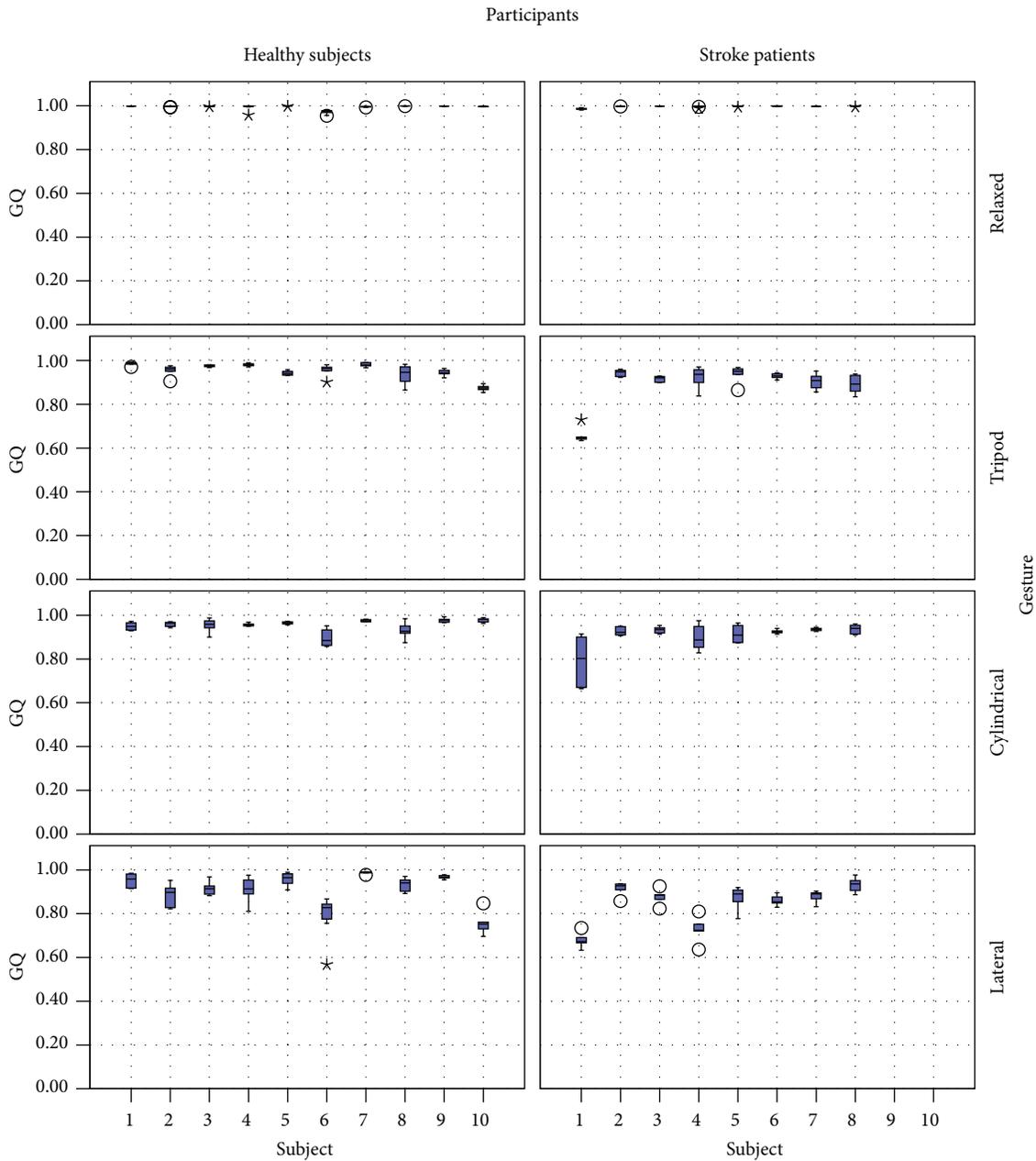
Test compared with mean accuracy	Pearson correlation	Sig.
ARAT_total	0.651	0.040
ARAT_A	0.630	0.047
ARAT_B	0.532	0.087
ARAT_C	0.680	0.032
ARAT_D	0.502	0.103
FM_total	0.461	0.125
FM_proximal	0.481	0.114
FM_distal	0.386	0.172
FM_coordination	0.141	0.370

accuracies and the ARAT test, especially to sections A (grasp subtest) and C (pinch subtest). This is not surprising given that the ARAT specifically assesses dexterity; while FM is a much broader measure of motor impairment. Figure 7 presents accuracy and ARAT score values for each subject to visually show the high association.

**3.2. Grasp Evaluation Results.** The results of the evaluation of each of the grasps using the proposed quality measure GQ are presented in Figure 8. The difference between the participant's type and the grasp performed are shown in Figure 8(a). The healthy subjects presented smaller variations than the stroke patients and their median grasp quality was higher. The higher variation was presented while the participants



(a)



(b)

FIGURE 8: Results of the grasp quality (GQ) for healthy subjects and stroke patients per gesture: (a) summary of the results and (b) results showing variability per participant.

TABLE 6: Correlation between the grasp quality measure (GQ) and common arm motor recovery tests.

Test compared with GQ	Relax posture		Tripod grasp		Cylindrical grasp		Lateral grasp	
	Pearson correlation	Sig.						
ARAT_total	-0.365	0.187	-0.387	0.172	-0.247	0.278	0.078	0.427
ARAT_A	-0.375	0.180	-0.443	0.136	-0.247	0.278	0.008	0.492
ARAT_B	-0.139	0.371	-0.214	0.305	-0.007	0.494	0.209	0.310
ARAT_C	-0.527	0.090	-0.502	0.102	-0.434	0.141	-0.055	0.448
ARAT_D	-0.077	0.429	-0.051	0.452	0.020	0.482	0.347	0.200
FM_total	-0.117	0.391	-0.133	0.376	-0.005	0.496	-0.037	0.465
FM_proximal	0.130	0.380	-0.083	0.422	-0.002	0.498	-0.071	0.434
FM_distal	-0.042	0.461	-0.086	0.419	0.044	0.459	0.023	0.479
FM_coordination	-0.638*	0.044	-0.660*	0.038	-0.574	0.069	-0.441	0.137

\*Correlation is significant at the 0.05 level (1-tailed).

ARAT = arm scores of the action research arm test; FM = arm scores of the Fugl-Meyer motor assessment.

TABLE 7: Multiple regression results.

Model	R	R square	Adjusted R square	Std. Error of the Estimate	Change statistics				
					R Square change	F change	df1	df2	Sig. F change
1	0.780	0.609	0.597	0.04365	0.609	53.469	20	687	0.000

performed the lateral grasp: the grasp quality varied 24% (0.75–0.99) for healthy patients and 28% (0.67–0.95) for stroke patients.

The variability between different participants is shown in Figure 8(b). The grasp performed by healthy participants got a measure of quality above 0.85 for the relaxed, tripod, and cylindrical grasps. However, performing the lateral grasp presented a higher variability, especially for subjects 6 and 10 who obtained grasp qualities as low as 0.57 and 0.73, respectively. Stroke patients presented a higher variability between subjects, but in general there was a consistent grasp quality per subject and gesture (maximum 17%)—except patient 1 performing the cylindrical grasp who showed a variation of 25% (0.66–0.91).

The correlation results of the grasp quality with the results of the arm motor recovery tests are presented in Table 6. In general, there are no significant correlations between the tests and the results obtained from the quality measure, except for a negative correlation with the Fugl-Meyer coordination test performing the relax posture and the tripod grasp.

In order to assess how the quality measure GQ was influenced by the type of participant (healthy subject versus stroke patient), the gesture and the specific subject performing the grasps, a multiple regression analysis was performed. The results of the analysis are presented in Table 7. The  $R$  value of 0.78 shows that the predictors are a good estimation of the quality measure. The  $R^2$  indicates that the predictors account for 60.9% of the variation of the grasp quality measure and as the adjusted  $R^2$  is similar to it, it means that the model is able to generalize. Also, the standard error 0.04365 indicates that most sample means from the estimated values are similar to the quality measure means. The  $F$  change is an important value, as it indicates that this model causes  $R^2$  to change from zero to 0.609, which is significant with a probability less than 0.001.

#### 4. Discussion

In this study, we examined the performance of a technique based on support vector machines for the recognition of hand gestures using the finger flexion/extension angles, comparing a group of healthy subjects with a group of poststroke patients. The results for stroke patients in general show lower accuracies, lower true positive rates, and greater false positive rates than the ones performed by the healthy subjects. This is due to the fact that patient's impairment after stroke affects their ability to reproduce gestures with small variations and in a repeatable way. However, we hypothesized that the accuracy for gesture recognition for stroke patients can provide an insight into patients' ability to consistently reproduce gestures. This was corroborated with the high correlation between the recognition accuracies and the scores of the ARAT ( $C = 0.651$  for ARAT total) showing that the accuracy of recognition can be used as a measure of the grasping capabilities for the impaired hand. Therefore, this method shows potency to be used to detect changes related to lack of ability to reproduce gestures in a consistent way.

The results of the grasp quality measure showed that the healthy subjects presented smaller variations than the stroke patients and their grasp quality is higher showing that the measure can be used to assess the capabilities of stroke patients to perform grasp postures in a similar way than healthy people. However, the variation between healthy subjects is higher than we expected. This can be due to the mechanical design of the orthosis, as the readings of the sensors could vary from orthosis to orthosis depending on the level of tension of the elastic cords. With a more accurate device, as the currently developed next version of the SCRIPT orthosis, it is expected that the variability of the healthy postures will be reduced which will give more

consistent high quality values for the healthy participants and therefore would be more reliable for the evaluation of stroke patient grasps. Also, a broader study with a larger number of healthy subjects and stroke patients could give a more accurate measure of the quality of the grasp.

Also, the lack of correlation between the grasp quality GQ and the results of the arm motor recovery tests could indicate that this measure is perhaps providing different information about the patients, indicating specifically their level of ability to perform each of the different gestures which is not specifically measured by the common tests. The results of the multiple regression analysis showed that the type of participant (healthy subject or stroke patient), the grasp and the specific subject can account for 61% of the grasp quality variability, which indicate presence of other indicators that have not been considered in our model. We hypothesize that these variations could be due to differences influenced by gender, the hand performing the gestures (dominant or nondominant), or the level of disability of the stroke patients. Future work is needed to study the influences of these factors when more participants and more accurate readings are available.

The technique presented in this paper will be used for the recognition of hand postures needed for the activities of daily life while patients playing rehabilitation games with the SCRIPT orthosis over a period of six weeks. With these results, more exhaustive analysis can be performed which could provide insights into improvements on the ability to perform the grasp postures over time influencing the overall motor performance.

## 5. Conclusion

The results obtained with this study show that a technique based on support vector machines can be used to recognize different grasp gestures for stroke patients with a valid accuracy while playing rehabilitation games wearing a specially designed exoskeleton for their rehabilitation. This will allow the training of various grasp postures to improve the performance of these postures needed for several activities of daily living. Moreover, we showed that the accuracy of recognition can be used to assess the ability of the stroke patients to consistently repeat the gestures and the proposed grasp quality GQ can be used to measure the capabilities of the stroke patients to perform grasp postures in a similar way than healthy people. These two measures could be used as complementary measures to the other upper limb motion tests such as ARAT and they can be potentially applied to evaluate the grasp performance of patients using other orthosis able to measure finger flexion.

## 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

# Assessment of Waveform Similarity in Clinical Gait Data: The Linear Fit Method

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The assessment of waveform similarity is a crucial issue in gait analysis for the comparison of kinematic or kinetic patterns with reference data. A typical scenario is in fact the comparison of a patient's gait pattern with a relevant physiological pattern. This study aims to propose and validate a simple method for the assessment of waveform similarity in terms of shape, amplitude, and offset. The method relies on the interpretation of these three parameters, obtained through a linear fit applied to the two data sets under comparison plotted one against the other after time normalization. The validity of this linear fit method was tested in terms of appropriateness (comparing real gait data of 34 patients with cerebrovascular accident with those of 15 healthy subjects), reliability, sensitivity, and specificity (applying a cluster analysis on the real data). Results showed for this method good appropriateness, 94.1% of sensitivity, 93.3% of specificity, and good reliability. The LFM resulted in a simple method suitable for analysing the waveform similarity in clinical gait analysis.

## 1. Introduction

Instrumented gait analysis (GA) allows for gathering quantitative information about joint kinematics and kinetics of the musculoskeletal system during gait. A critical issue in the field of human movement analysis refers to the data comparison and the assessment of the deviation of the gait pattern under analysis from reference data through a few meaningful indices [1]. The need to assess the similarity between two curves of gait data is encountered in a large variety of different scenarios, for example when two datasets are obtained using different instrumentations [2–4], using different protocols [5–7], before and after treatment [8, 9], or especially for comparing pathological gait data related to a patient (or a group of patients) to physiological gait data related to healthy subjects [10–12].

In general, the comparison between pairs of kinematic or kinetic curves is performed by computing the Pearson

correlation coefficient ( $R$ ), which allows for quantifying the strength of a linear relationship between the two curves (i.e., their shape similarity, independently from their amplitude or their mean difference) [2, 13], the root mean square error (RMSE) which provides a positive global index [2], or the coefficient of multiple correlation (CMC) which is helpful when the reliability of a group of curves is under analysis [14, 15].

Another simple approach consists in computing the difference between parameters assumed to be representative of the entire curve, such as the range of motion (ROM), or computing the mean difference between the mean values of the curve under analysis ( $P_a$ ) in respect of the mean value of the reference curve ( $P_{ref}$ ). Some indices were proposed to compare the ROM of the two curves in order to quantify the differences in terms of pattern amplitude, such as the ratio index ( $RI_{ROM} = ROM_{Pa}/ROM_{P_{ref}}$ ) [16] or the symmetry angle [17]. Conversely, the mean difference ( $MD = \overline{P_a} - \overline{P_{ref}}$ )

is used to assess a vertical shift (offset) between the curve under analysis and the curve used as reference [2].

More complex methods have also been suggested, including extended indices [18], Fourier analysis [19], principal component analysis [20], eigenvectors [21], fractal methods [22], neural networks [23], pattern recognition techniques [24], or ninth order polynomial fitting [25].

Another approach commonly adopted in gait analysis consists in calculating the differences between the curves in correspondence of specific gait events (e.g., at foot contact or at foot off). However, for each curve several points can be extracted and their selection could be arbitrary, and it implies a loss of information about the entire pattern and hence it can be critical.

Finally, it is possible to combine the above described approaches, such as in the study of Crenshaw and Richards [21], who suggested to compute five parameters three of which based on eigenvectors computation, and the other two being MD and  $RI_{ROM}$ .

There is hence a large amount of research in the field, but the multiplicity of proposed methods suggests that a simple approach to perform a global comparison between gait curves through a few parameters with specific physiological meaning is still lacking. The aim of this study is to present and validate a simple method based on linear fitting applied to two datasets plotted one versus the other (linear fit method, LFM). This method was preliminary suggested by our group [26] and then, despite not being validated yet, applied for comparing gait data obtained in different laboratories [27]. Presented here is the validation of LFM, carried out in terms of appropriateness, sensitivity, specificity, and reliability.

## 2. Material and Methods

**2.1. Analytical Description of the Linear Fit Method.** Let  $P_a$  be the kinematic (or kinetic) dataset under investigation that should be compared with  $P_{ref}$ , that is, the reference dataset (as shown in the left plot of Figure 1). As usual in GA, the datasets are time-normalized between two selected events, such as two consecutive foot-strikes of the same limb which define the gait cycle [28]. Since  $P_a$  and  $P_{ref}$  are normalized in respect of gait cycle, they result in two arrays of real numbers of the same length. It is hence possible to plot  $P_a$  against  $P_{ref}$  to define a set of points in a Cartesian coordinate system  $((X_i; Y_i)$  with  $i = 1 \dots N$ ) where  $P_{ref}$  values correspond to  $x$ -values and  $P_a$  to  $y$ -values (Figure 1). The present method is based on applying a linear fit to this set of points. This fitting minimizes, in a least squares sense, the sum of the square vertical distances between the points and the fitting line (regression line, right plot of Figure 1):

$$Y_a = a_1 \cdot P_{ref} + a_0, \quad (1)$$

where  $Y_a$  represents the linear function which approximates  $P_a$  values by means of a linear transformation of values of  $P_{ref}$ ;  $a_1$  is the angular coefficient; and  $a_0$  is the intercept of the fitting line. The goodness of the fit can be easily assessed by the coefficient of determination  $R^2$  which coincides with, for

the properties of linear fit, the square of the Pearson's correlation coefficient  $R$ .

The LFM relies on the interpretation of the values of  $R^2$ ,  $a_0$ , and  $a_1$  for assessing curve similarity between  $P_a$  and  $P_{ref}$ .

The formulas for computing the above three parameters are

$$\begin{aligned} a_1 &= \frac{\sum_{i=1}^N (P_{ref}(i) - \overline{P_{ref}}) \cdot (P_a(i) - \overline{P_a})}{\sum_{i=1}^N (P_{ref}(i) - \overline{P_{ref}})^2}, \\ a_0 &= \overline{P_a} - a_1 \cdot \overline{P_{ref}}, \\ R^2 &= \frac{\sum_{i=1}^N (a_0 + a_1 \cdot P_{ref}(i) - \overline{P_a})^2}{\sum_{i=1}^N (P_a(i) - \overline{P_a})^2}, \end{aligned} \quad (2)$$

with  $N$  the length of datasets (corresponding to the 100% of gait cycle) and overline used for indicating the mean value of a dataset.

The meaning of LFM parameters is described below.

- (i)  $a_1$  measures the mean variation of  $P_a$  for every one-unit change in  $P_{ref}$ . It hence represents the amplitude scaling factor, that is, the factor for which  $P_{ref}$  should be multiplied to match  $Y_a$  except for a scalar addition.
- (ii)  $a_0$  predicts this scalar addition (shift), that is, the value of  $P_a$  when  $P_{ref}$  is equal to 0.
- (iii)  $R^2$  measures the strength of the linear relationship between  $P_a$  and  $P_{ref}$ , that is, the percentage of variance in  $P_a$  that can be matched by the variance in  $P_{ref}$ .

It should be noted that if  $P_a = P_{ref}$  then the values of LFM parameters are  $a_1 = 1$ ,  $a_0 = 0$ ,  $R^2 = 1$ . Further, the LFM has the advantage that, for its intrinsic linearity, the mean  $a_1$ - and  $a_0$ -values obtained from  $n$  comparisons of  $n$  different  $P_a$ -data sets with their mean pattern are equal to the ideal values:  $a_1 = 1$  and  $a_0 = 0$ . This is the case when  $n$  curves of healthy subjects are compared with a reference pattern obtained as their mean.

**2.2. LFM Validation.** To describe the application and the advantages of LFM in comparison with other parameters commonly used in GA, we first analyzed synthetic datasets generated from a real reference pattern (a physiological knee sagittal kinematics) in which mathematical transformations were applied in order to simulate specific gait pattern alterations. Synthetic data were used to allow for perfectly knowing the mathematical difference between  $P_a$  and  $P_{ref}$ .

LFM was then validated using real data in terms of (a) appropriateness (does it provide different results for patients when compared to healthy subjects?), (b) sensitivity (does it detect a specific difference in the curves?), (c) specificity (is it able to detect as pathological only actually pathological patterns?), and (d) reliability (can the measures be repeated accurately?).

In particular, to test the capacity of LFM to detect a difference when it is present, we have applied it on five different synthetic arrays of data ( $Y_i$ ,  $i$  from 1 to 4) obtained by altering the mean knee sagittal kinematics ( $P_a$ ) obtained

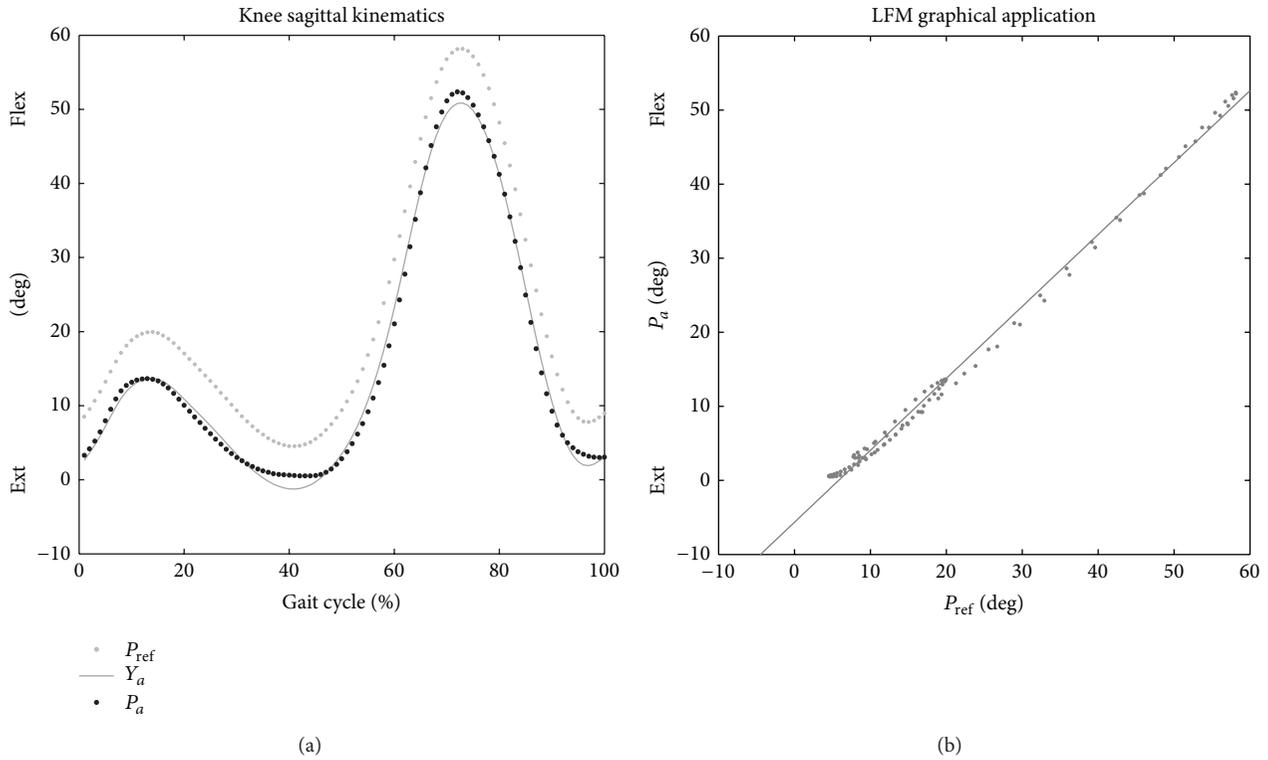


FIGURE 1: Two exemplificative knee sagittal kinematic datasets were compared in order to graphically illustrate the LFM. The circles represent the 100 values obtained for the two knee kinematics when time-normalized and reported in terms of gait cycle. On the left are the points for the investigated dataset  $P_a$  (black dots) and for the reference dataset  $P_{ref}$  (grey dots). The grey line represents the reconstructed curve  $Y_a$  obtained by the parameters of the linear fit applied to the values of  $P_a$  when plotted versus  $P_{ref}$  (right plot).

from 15 healthy subjects acquired by means of a stereophotogrammetric system during level walking. The mathematical reshape of these reference data allowed for examining the variation of LFM parameters when one (or more) specific feature of the curve was selectively altered in order to simulate a specific knee impairment ( $Y_1$ : hyperextended knee,  $Y_2$ : knee with reduced mobility,  $Y_3$ : stiff knee,  $Y_4$ : hyperflexed knee). For these curves, values of the three parameters obtained using LFM were compared with the values of three parameters commonly used in literature:  $RI_{ROM}$ , MD, and RMSE.

To assess the LFM appropriateness, the data relative to the sagittal kinematics of hip, knee and ankle of 15 healthy subjects were compared with those of 34 patients affected by cerebrovascular accident (CVA). The values of LFM parameters obtained for the two groups were hence compared by means of unpaired 2-tailed  $t$ -tests. For this and all the other statistical tests applied in the present study the threshold for statistical significance was set at 0.05.

Relevant mean, standard deviation, and 95% confidence interval ( $IC_{95\%}$ ) for each set of data were also computed.

LFM sensitivity and specificity were assessed performing a Wilks' lambda discriminant analysis computed on the above described real data. This analysis was performed to assess the capacity of LFM to cluster the subjects into two groups: healthy group and patient group.

The reliability of LFM was evaluated by computing the intraclass correlation coefficient (ICC (2, 1)) for hip, knee, and ankle sagittal kinematics.

Gait datasets were acquired using a 9-camera motion capture system (Smart-D system, BTS Bioengineering, Milan, Italy) to reconstruct the 3D position of 21 retroreflective spherical markers located on the subjects skin according to the conventional method [29], during level walking in barefoot conditions at self-selected speed. Datasets were related to three trials per side for each one of fifteen healthy subjects and six trials of affected side for each one of the 34 patients with CVA.

### 3. Results

**3.1. Comparison of LFM Parameters to Other Parameters on Synthetic Data.** Figure 2 shows four synthetic reproductions of knee impairment, and Table 2 reported the relevant mathematical equations applied to obtain these data, together with the values of computed parameters. The values obtained for  $RI_{ROM}$ , MD, and RMSE were misleading and less meaningful in respect of the values of LFM parameters. In detail, the analysis performed on the synthetic datasets  $Y_2$  and  $Y_3$  highlighted that when the offset between curves is evaluated using MD, its value is influenced by the amplitude differences (Table 1), whereas the LFM evaluated the shift independent of

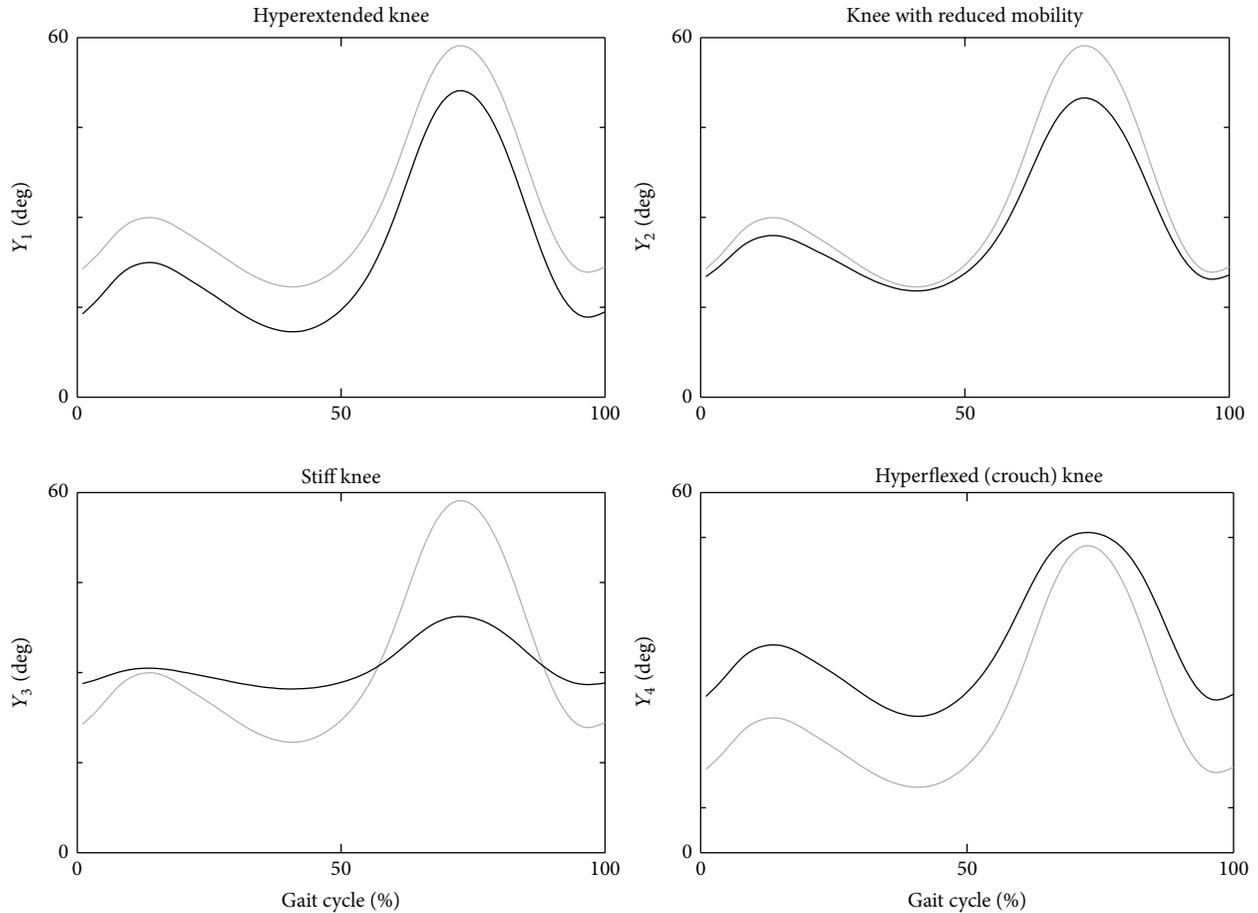


FIGURE 2: Four synthetic reshapes (black lines) mathematically obtained by a knee reference pattern (grey line), simulating four different impairments.

TABLE 1: Parameter values for the curves shown in Figure 2.

Pattern	$RI_{ROM}$	MD	RMSE	Linear fit method		
				$a_1$	$a_0$	$R^2$
$Y_1 = P_a - 10^\circ$	1	$-10^\circ$	$10^\circ$	1	$-10^\circ$	1
$Y_2 = 0.8 * P_a$	0.8	$-4^\circ$	$6^\circ$	0.8	0	1
$Y_3 = 0.3 * P_a + 15^\circ$	0.3	$0^\circ$	$12^\circ$	0.3	$15^\circ$	1
$Y_4 = -0.007 * P_a^2 + 1.2 * P_a + 15^\circ$	0.76	$14^\circ$	$15^\circ$	0.77	$19^\circ$	0.98

amplitude differences, by  $a_0$  and  $a_1$ , respectively. The  $RI_{ROM}$  and  $a_1$  values were very similar for the analysed synthetic data, despite  $a_1$  values were not dependent by the artefacts affecting the values of  $RI_{ROM}$ . The RMSE values ranged between  $10^\circ$  and  $15^\circ$  for three out of four investigated datasets ( $Y_1, Y_3, Y_4$ ), despite that the differences were due to an offset in  $Y_1$ , to a combination of offset and amplitude differences in  $Y_3$  or to shape dissimilarity in  $Y_4$ . In this last case, the mathematical transformation was not linear, but the value of  $R^2$  remained high, and the waveform differences were better quantified by parameters of LFM than by  $RI_{ROM}$ , MD, and RMSE.

**3.2. Appropriateness.** LFM has been applied to analyze the sagittal kinematics of healthy subjects (walking at  $68 \pm 10\%$  of their stature/s) and of patients with CVA (walking at  $42 \pm 15\%$ ,  $P < 0.001$ ). Mean, standard deviation, and confidence interval of the sagittal hip, knee, and ankle kinematics of the healthy subjects group were reported in Table 2. For healthy subjects, the mean  $a_1$  and  $a_0$  resulted to be equal to ideal values 1 and 0, respectively, whereas the mean  $R^2$  was just close to its ideal value for hip (0.99), knee (0.97), and ankle (0.89) joint (in mean 0.95). The statistical analysis reported in Table 2 showed that the values of the LFM parameters resulted to be statistically different between patients and

TABLE 2: Mean  $\pm$  standard deviation (SD) of the values of LFM parameters for normative data obtained by healthy subjects and relevant values for patients with CVA. For healthy subjects the 95% interval of confidence (IC<sub>95%</sub>) is reported, whereas for patients, the  $P$  value of comparison with healthy subjects' values is reported.  $R^2$  and  $a_1$  are adimensional coefficients, whereas  $a_0$  is measured in degrees.

Mean $\pm$ SD (IC <sub>95%</sub> )	Hip	Knee	Ankle
Mean $\pm$ SD (IC <sub>95%</sub> ) of healthy subjects			
$R^2$	0.99 $\pm$ 0.01 (0.98; 0.99)	0.97 $\pm$ 0.02 (0.96; 0.98)	0.89 $\pm$ 0.06 (0.86; 0.92)
$a_1$	1 $\pm$ 0.09 (0.96; 1.04)	1 $\pm$ 0.08 (0.96; 1.04)	1 $\pm$ 0.13 (0.94; 1.06)
$a_0$	0 $\pm$ 7.48 (-3.79; 3.79)	0 $\pm$ 7.51 (-3.80; 3.80)	0 $\pm$ 4.12 (-2.09; 2.09)
Mean $\pm$ SD ( $P$ of $t$ -test) of subjects with impairment			
$R^2$	0.90 $\pm$ 0.07 ( $P < 0.001$ )	0.75 $\pm$ 0.20 ( $P < 0.001$ )	0.40 $\pm$ 0.24 ( $P < 0.001$ )
$a_1$	0.77 $\pm$ 0.18 ( $P < 0.001$ )	0.70 $\pm$ 0.29 ( $P < 0.001$ )	0.42 $\pm$ 0.24 ( $P < 0.001$ )
$a_0$	6.72 $\pm$ 11.21 ( $P = 0.040$ )	0.17 $\pm$ 9.69 ( $P = 0.953$ )	-0.09 $\pm$ 6.40 ( $P = 0.961$ )

healthy subjects. In detail, for patients  $R^2$  resulted to be significantly lower at all the three joint levels, the movement amplitude resulted to be significantly reduced for all the three joints ( $a_1 \leq 0.70$ ), and the hip resulted to be hyperflexed ( $a_0 = 6.7^\circ$ ).

3.3. *Sensitivity and Specificity.* The correctness of classification in physiological versus pathological patterns obtained by LFM prediction was 79.4%, 76.5%, and 94.1%, respectively, for hip, knee, and ankle data (in mean 83.3%). It means that only 2 patients out of 34 (5.9%) were classified as without any impairment. For healthy subjects, among all the three joints, only one false positive was found: a knee pattern of one healthy subject was classified as pathological. Hence, the specificity resulted to be of 93.3% (true negative rate: 14/15) and sensitivity resulted to be of 94.1% (true positive rate: 32/34).

3.4. *Reliability.* The reliability of parameters computed for healthy subjects and patients was assessed by means of the intraclass correlation coefficient computation. The mean ICC values for patients are shown in Table 3 (mean value: 0.91  $\pm$  0.07). The reliability was lower for healthy subjects (mean value: 0.69  $\pm$  0.21). A similar trend was observed for the walking speed; the value of ICC was higher in patients (0.91) than in healthy subjects (0.88).

TABLE 3: Analysis of reliability: results of intraclass correlation coefficients.

ICC	Hip	Knee	Ankle
Subjects with impairment			
$R^2$	0.80	0.84	0.84
$a_1$	0.92	0.95	0.88
$a_0$	0.99	0.97	0.97

### 4. Discussion

The aim of our study was to present and validate a linear fit method for assessing the similarity between curves relative to gait datasets. This assessment is usually the basis of GA, both for clinical and research purposes.

The results obtained in this study on synthetic data showed that the values with conventionally used parameters, such as MD,  $RI_{ROM}$ , and RMSE, can be misleading. Conversely,  $a_0$  and  $a_1$ , two of the LFM parameters, can be used as representative of offset and amplitude difference, respectively, without the following problems affecting the values of MD and  $RI_{ROM}$ . For example MD, which is generally adopted to assess the presence of a vertical shift, can be potentially affected by changes in amplitude (as evident for  $Y_2$  and  $Y_3$ ). The problem of vertical shift is particularly important, for example, when tests were repeated and GA-markers need to be replaced (potentially introducing an offset): this shift is the most important factor in reducing the reliability of repeated measures and it should be properly assessed [27, 30]. On the other hand, the use of ROM (and hence  $RI_{ROM}$  and other related indices) can have some disadvantages: (1) it only compares the differences between the maximum and the minimum of the curves, independently from the data distribution; (2) the mean ROM of  $n$  curves can be very different from the ROM of the relevant mean curve, and it can affect the assessment of amplitude similarity. The recovery of a functional ROM is an important outcome measure in rehabilitation. Differently from ROM,  $a_1$  takes into account the amplitude of the gait pattern in respect of the physiological pattern along the entire gait cycle. The RMSE has the problem that its values were similar over the different conditions (such as synthetic data  $Y_1, Y_3, Y_4$ ) and hence its physiological meaning is difficult to be argued. More clear is the meaning of Pearson correlation coefficient  $R$  (despite being improperly used in many studies as an indicator of agreement of two datasets) [31, 32]. However the same information obtained with  $R$  can be obtained in LFM by the  $R^2$  value, which provides a measure of the shape similarity of two curves with a clear mathematical meaning (the percentage of variance of the dataset under analysis explained by a linear transformation of the reference dataset) and a clear physiological meaning (the pattern similarity despite possible amplitude differences or presence of a shift).  $R^2$  resulted in an index for summarising the waveform similarity, and it can potentially quantify the efficacy of a treatment in relationship to functional and structural recovery as indicated by ICF [33].

The LFM have some advantages clearly reported and that could be summarised as follows: (1) LFM takes into account

all data point distributions (resulting in less dependence on single peak-values than ROM and other similar parameters); (2) LFM is simple to be applied; (3) this simplicity implies a clear meaning of its few (three) parameters; (4) its linearity implies that mean parameter values of  $a_0$  and  $a_1$  are equivalent to parameter values of the mean curve; (5) this linearity also allows for using powerful parametric statistics; (6) because  $a_1$ ,  $a_0$ , and  $R^2$  were computed at the same time, they independently assess curve differences in terms of amplitude, offset, and similarity, respectively.

We have validated the LFM on kinematic data, but this validation is clearly extensible to kinetic or even electromyographic data because they are all usually time normalized and reported in terms of gait cycle. Our validation showed LFM is appropriate to discriminate between patients and healthy subjects, showing a good sensitivity in identifying pathological gaits and a good specificity (only one false classification out of 15 in the pathological cluster definition). The reliability of LFM parameters was found high, especially for patients and even for the ankle which was characterized by low  $R^2$  values. The lower ICC found for healthy subjects is not surprising. In fact, in healthy populations the intersubjects variability is similar to intrasubject variability, resulting in a low ratio between intra- and intersubject variability, as it was already highlighted [10].

In this study, we compared datasets obtained using a stereophotogrammetric system, according to the conventional gait analysis. In the last decade, alternative approaches have been developed based on the use of wearable sensors (including accelerometers, gyroscopes, and magnetic units) [2, 34–37]. In this study, we did not test datasets obtained using these devices; however, it is conceivably that LFM could be useful for comparing the results of these new approaches with those obtained with the conventional gold standard (i.e., the stereophotogrammetric system).

The proposed LFM have some limitations which should be considered and which can bound its fields of application: (1) the values of  $a_0$  and  $a_1$  lose meaning when the linear relationship between  $P_a$  and  $P_{ref}$  is poor (low value of  $R^2$ ); (2) there is the need for identifying one of the two dataset as the reference one ( $P_{ref}$ ); (3) a bias can be introduced by the presence of a phase shift. First, LFM relies on the hypothesis that two gait patterns related to a specific joint are usually characterized by a similar waveform, given the intrinsic biomechanical constraints of the musculoskeletal system. In this respect, it is necessary to define when the hypothesis of linear relationship decays. Despite  $R^2$  being high even for nonlinear transformations, such as in synthetic data  $Y_4$  or for hip and knee joints in real data, it should be taken into account that for  $R^2 < 0.50$  (i.e., less than 50% of  $P_a$  variance matched by  $P_{ref}$  variance, corresponding to a  $R < 0.70$ ) the relationship between  $P_a$  and  $P_{ref}$  can be only partially described by means of a linear transformation and hence the values of  $a_1$  and  $a_0$  should be carefully handled. Second, the results of LFM depend on which dataset, between  $P_a$  and  $P_{ref}$ , is chosen as reference. This problem is common also to other methods. For LFM, the optimization of the fitting is determined by minimizing vertical distances

between points obtained from plotting  $P_a$  versus  $P_{ref}$  and fitting a line (see Figure 1). This characteristic may represent a potential limitation of the method, although the existence of a clearly defined  $P_{ref}$  is a common circumstance in most clinical applications (e.g., in comparing the pathological gait patterns to normative data). However, when the reference data set chosen is questionable, such as symmetry assessment in healthy subjects [17], LFM could still be used, but for applying a fit minimizing the orthogonal distances between points and fitting a line. Finally, we would highlight that when a phase shift between  $P_a$  and  $P_{ref}$  is present (i.e., a shift along the horizontal gait cycle axis), this results in a reduction of shape similarity. Since gait curves are usually time normalized and expressed as a percent of the gait cycle this aspect should not be critical in most of the cases, but if a phase shift is present we suggest to previously perform a cross-correlation for quantifying the horizontal shift and then apply the LFM to the realigned curves.

In conclusion, the strengths and attractiveness of the proposed linear fit method are its easy mathematical implementation, the use of a few parameters, their straightforward physical interpretation, and their evident clinical meaning.

## 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

# Inter- and Intrarater Reliability of Modified Lateral Scapular Slide Test in Healthy Athletic Men

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*Objective.* The reliability of lateral scapular slide test (LSST) at 90 degrees of abduction is controversial; therefore, in order to achieve more reliability it may be necessary to make changes in this particular position. *Methods.* Modified lateral scapular slide test (MLSST) was done on thirty male basketball players with two examiners in one session and for the retest with one examiner in the next week. The test was done in 7 positions: arm relaxed at the side (P1), 90 degrees of abduction (P2), 90 degrees of scaption without having a weight in hands (P3), 90 degrees of scaption with having 3 different weights (1, 2, and 4 kg) in hands (P4, P5, and P6, resp.), and 180 degrees of scaption without having a weight in hands (P7). *Results.* In P1 and P6, the ICC scores indicated the highest level of intrarater reliability. In P2, the ICC scores showed a fair level of intrarater reliability, as the minimum reliability. The maximum and minimum interrater reliability were P1 and P4, respectively. *Conclusion.* Scaption with loading, as a functional position in the overhead athletes, is a reliable positioning and may be replaced with the third position of the traditional LSST.

## 1. Introduction

Alteration of scapular position during shoulder motions is commonly associated with injuries that create clinical dysfunction of the shoulder. In overhead sports in which demands placed on the shoulder are extremely high, abnormal scapular kinematics is more commonly involved [1].

The lateral scapular slide test (LSST) developed by Kibler [2] is a more available and clinical method of examining the scapular positioning; therefore, further efforts to improve the reliability of this test are more valuable and cause the test to be more clinical.

The reliability of LSST has been examined in many previous studies but the results were not satisfying and have shown controversial results at 90 degrees of abduction, especially [3–7]. Shadmehr et al. reported poor ICC scores (0.63) in this position for interrater reliability in patients with different shoulder dysfunctions [5]. On the other hand, da Costa et al. showed that, at 90 degrees of elevation in scapular plane, interrater reliability was fair (ICC = 0.74) and intrarater reliability was good (ICC = 0.85) [6]. Struyf et al. determined

poor interrater reliability for 90 degrees of abduction with 1 kg loading (ICC = 0.63) in musicians [7].

With respect to the fair reliability in functional position of scaption, it seems that, with adding the loading to this position, its functional role is better determined and this modification may lead to more satisfying results than Struyf's study that added the loading to the abduction position. Scapular positioning is hypothesized to bear a direct relationship with muscle performance in basketball players who do overhead throwing. Additionally, kinematics alternations in scapular motion have been linked to shoulder pathologies [1]. As a result, basketball players are suitable for evaluating the effects of loading on the scapular kinematics.

Therefore, the main goal of the present study was to determine the reliability of LSST in scapular plane with various loadings in basketball players.

## 2. Materials and Methods

This was a cross-sectional, prospective, and repeated measures study of thirty athletes with two examiners in two

sessions. Thirty healthy male basketball players, between 20 and 31 years old, were recruited from three basketball teams. Prior to participating in this study, all athletes signed an informed consent form. This study was approved by the ethics committee of Tehran University of Medical Sciences (TUMS).

The inclusion criteria for the participants were the ability to actively perform 90 and 180 degrees of scaption, abduction, and full internal rotation of the shoulder, the age of 20–40 years, and a minimum of 2 years of sports experiences or 3 to 4 exercise sessions per week in basketball. Athletes were excluded if they had the following problems: previous shoulder surgery, overuse injuries, a history of systemic diseases, and neuromuscular dysfunctions.

A digital Vernier caliper with an accuracy of 0.01 mm (Mitutoyo Company, Japan), a goniometer with extendable arm (Lafayette Instrument Company, USA), and 1, 2, and 4 kg dumbbell-like weights were used in these experiments. Demographic information including age, mass, height, and body mass index (BMI) was measured and related questions were answered. To determine the scaption plane, the basketball players stood the front of the wall corner on a reference plate with signs for foot stands and a reference line on it. To determine the 90 and 180 degrees of shoulder abduction, we set different parts of a goniometer as follows: a fixed extendable arm was set along with external body line; shaft of the goniometer was under the external part of acromion with a distance of 2 cm and the mobile arm parallel to athletes' arm axis in 90 or 180 degrees. To measure the scaption angle, we set the fixed goniometer arm on the body axis in frontal plane and we put the axis of goniometer on the acromion tip and the mobile arm parallel to athlete arm axis [8]. Since there may be some differences in participant's upper extremity length, we extended the extendable mobile arm of goniometer up the wall and marked crossing points by markers on it. This was done at 90 and 180 degrees of scaption for both upper extremities separately. All athletes were asked to keep upper extremities compatible with these markers. These markers were positioned on the wall in a way in which they made 40 degrees with frontal body plane to guide athletes along with scapula plane [6].

In the first position of the test (P1), the upper limbs were hanging beside the body and the examiner measured the least linear distance between T7 spinous process and inferior angle of scapula in 0.01 mm order using digital caliper (Mitutoyo Corporation, Japan) (Figure 1(a)). In the second position of the test (P2), the upper limbs were at 90 degrees of abduction with internal rotation of arms and the examiner found T7 spinous process and scapula inferior angle by touching; then, he measured the least linear distance between them using caliper in both sides. In the third position of the test (P3), the athlete was asked to keep his upper limbs at 90 degrees of scaption with internal rotation of shoulder (along with mentioned markers) without having a weight in his hands. Then 1, 2, and 4 kg dumbbell-like weights were given to athletes in the fourth (P4), fifth (P5), and sixth (P6) positions of the test, and they were asked to keep their upper limbs at 90 degrees of scaption with internal rotation of shoulder (along with mentioned markers) while keeping weights. In these positions the examiner measured the mentioned distance

TABLE 1: Mean, standard deviation (SD), and range of demographic data of participants ( $n = 30$ ).

Demographic data	Mean (SD)	Range
Age (years)	22.53 (3.72)	20–31
Height (cm)	187.33 (9.81)	170–210
Mass (kg)	84.17 (16.29)	54–130
BMI* (kg/m <sup>2</sup> )	23.83 (3.04)	18.68–30.26
Sports experience (years)	8 (5.5)	2–20

\* BMI: body mass index.

again (Figures 1(b) and 1(c)). In the seventh position of the test (P7), the athletes were asked to keep their trunk fixed and move up their arms along with markers put on the wall, without having weights in hands and as far as possible. In this condition the examiner measured the mentioned scapular distances using caliper (Figure 1(d)).

All of the above measurements were done by two randomly selected expert examiners with M.S. degree in 5-minute intervals (interrater). Before beginning the tests, examiners were qualified to find the landmarks and use caliper in measurements in sufficient time and in sports club. There was a 30-second break between two consecutive tests. Each examiner separately recorded the results in separate sheets without talking to other examiners about the test and its results. One of the examiners repeated the tests after one week exactly in the previous test times (intrarater).

Intraclass correlation coefficient (ICC) ([2, 1] two-way random effects model) and 95% confidence interval (95% CI) for the ICC were used to analyze the reliability of MLSST. A 1-sample Kolmogorov-Smirnov goodness-of-fit test was done to determine normal distribution ( $P > 0.05$ ; data not shown).

Standard error of measurement (SEM) equals the square root of the mean square of the error [9]. The ICCs were classified as follows: <0.69, poor correlation; 0.70–0.79, fair correlation; 0.80–0.89, good correlation; 0.90–1.00, high correlation [5]. All data were analyzed using SPSS version 19.

### 3. Results

In the present study, inter- and intrarater reliabilities of MLSST were determined in healthy basketball players. Table 1 shows the demographic characteristics of the participants involved in this study.

Intrarater and interrater ICCs (single and average measures), 95% CI, and SEM are presented in Table 2. In P1 and P6 positions, the ICC scores indicated a high level of reliability for intrarater (0.94 and 0.90, resp.), as the maximum reliability. In P2 position, the ICC scores showed a fair level of reliability for intrarater (0.79), as the minimum reliability. The maximum and minimum interrater reliability were P1 (0.77) and P4 (0.54), respectively.

The highest and the lowest intrarater SEM were 0.74 cm and 0.32 cm for P2 and P1 positions, respectively. The results showed that the amount of errors in measurements of the two examiners (interrater) was higher than intrarater. The lowest

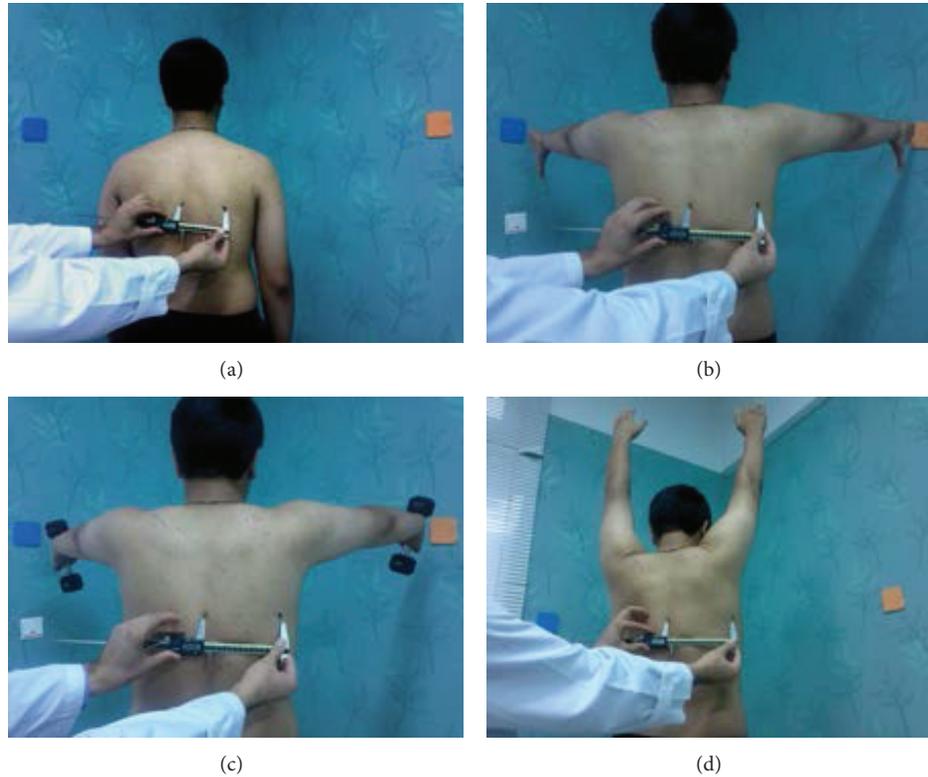


FIGURE 1: The measurement of the distance between the spinous process of T7 and inferior angle of scapula in (a) neutral position, (b) unloaded scaption, (c) loaded scaption, and (d) full scaption.

TABLE 2: Mean and standard deviation (SD) of measurement scores, intraclass correlation coefficient [ICC; 95% confidence interval (CI)], and standard error of measurement (SEM) for intrarater and interrater reliability.

Position	Intrarater reliability			Interrater reliability		
	Single measurement ICC (95% CI)	Average measurement ICC (95% CI)	SEM (cm)	Single measurement ICC (95% CI)	Average measurement ICC (95% CI)	SEM (cm)
P1	0.94 (0.90–0.96)	0.97 (0.95–0.98)	0.32	0.77 (0.65–0.86)	0.87 (0.79–0.92)	0.71
P2	0.79 (0.67–0.87)	0.88 (0.80–0.93)	0.74	0.63 (0.45–0.76)	0.77 (0.62–0.86)	1.08
P3	0.82 (0.71–0.89)	0.90 (0.83–0.94)	0.59	0.73 (0.59–0.83)	0.85 (0.74–0.91)	0.82
P4	0.86 (0.78–0.91)	0.92 (0.87–0.95)	0.45	0.54 (0.33–0.69)	0.70 (0.50–0.82)	0.97
P5	0.89 (0.82–0.93)	0.94 (0.90–0.96)	0.45	0.67 (0.50–0.79)	0.80 (0.67–0.88)	0.78
P6	0.90 (0.84–0.94)	0.95 (0.91–0.97)	0.41	0.64 (0.47–0.77)	0.78 (0.64–0.87)	0.78
P7	0.87 (0.78–0.92)	0.93 (0.88–0.96)	0.37	0.56 (0.35–0.71)	0.71 (0.52–0.83)	0.91

Note: arm is relaxed at the side (P1), 90 degrees of abduction (P2), 90 degrees of scaption without having a weight in hands (P3), 90 degrees of scaption with having 3 different weights (1, 2, and 4 kg) in hands (P4, P5, and P6, resp.), and 180 degrees of scaption without having a weight in hands (P7).

error belonged to P1 (0.71 cm) and the highest amount of errors happened in P2 (1.08 cm).

#### 4. Discussion

The purpose of this study was to evaluate the reliability of the MLSST in healthy overhead sportsmen. Our findings showed that the first position (P1) had the highest level of intrarater reliability. Previous studies confirmed our results [5–7]. Since in this position it is easy to touch inferior angle

of scapula, furthermore, scapula remains in a static position and therefore it is easier to determine its location.

We found that, at 90 degrees of scaption (P3), there is good intrarater and fair interrater reliability. da Costa et al. in similar raters showed results similar to our study's [6]. The above findings may be due to the fact that scaption is the functional and true physiological movement of the shoulder abduction [10]. In this plane the glenohumeral capsule is not twisted and therefore the humeral movement is less restricted compared to the frontal plane [11]. As a result, in most of daily

activities and sports, scaption is a dominant and comfortable position.

There is a natural scapulothoracic rhythm and muscle timing between glenohumeral abduction and scapulothoracic upward rotation accompanied by 20 degrees of scapular posterior tilt [12]. Chu et al. also showed that the range of posterior tilt of scapula is lower in the scaption position than in the abduction position [12]. By increasing the posterior tilt of the scapula, its inferior angle moves closer to the thorax and obtains a deeper position. Therefore, it is more difficult to touch the inferior angle of the scapula as a key point of LSST. It seems easier to touch the inferior angle of the scapula in scaption and this may be a reason for improving the reliability in this position.

In the second modification, we asked athletes to handle 1, 2, and 4 kg loads during 90 degrees of scaption. Our study showed, that by increasing the loads (1, 2, and 4 kg), to unloaded scaption position (ICC = 0.82), the ICC scores for intrarater reliability increased as well (0.86, 0.89, and 0.90, resp.). Our participants were basketball players whose dominant task is throwing a heavy ball, so with applying verified loads in scaption we could perform the LSST in a more functional position for these athletes. We believe that applying the loads in scaption recruits more motor units from stabilizing muscles of the scapula and leads to a higher coordination in surrounding muscles that have a major contributing role in the scapular mobility and stability.

Struyf et al. in their study applied 1 kg load to the healthy musicians at 90 degrees of abduction and showed poor reliability with this modification, which is in contrast with our results [7]. It can be attributed to different sampling, because the overhead players may tolerate applied loads in a more stable position and hence show more reliability in the test compared to the musicians.

Good reliability for intrarater in P7 showed that functional conditions can be very useful for musculoskeletal assessments. In this position, inferior angle of scapula moves to lateral position more than other positions and it has better capability to palpate in the inferior portion of the axilla. Thus, it is not unexpected that this position has good reliability. In addition, it is easier to hold the status relative to scaption (with and without weights) and abduction position for the athletes. Therefore, it causes minimal SEM in this position (SEM = 0.38 cm). Unlike the good reliability for intrarater in P7, fair interrater reliability (ICC = 0.71) shows that in above position the agreement between the raters is low.

Our results determined that the interrater reliability has lower scores compared to the intrarater reliability in all positions. Regarding the fact that the LSST includes two objective and subjective parts, it is noticeable that the results of the test are strongly dependent on the rater's experience and accuracy of bony landmarks determination. It may explain the lower scores of interrater reliability of the LSST. It is recommended in intermittent clinical settings to conduct the test by one examiner.

The interrater standard error of measurement (SEM), which indicates absolute reliability, was lower in this study

in comparison with da Costa et al.'s study [6]. Better precision of caliper than palpation meter may explain apparent improvement in reliability. In the present study we showed a descending trend of SEM from P2 to P6 positions. This means that, by increasing the load, the error of measurement was decreased. On the other hand, it seems that SEM may be affected by samples characteristics. In the present study all participants ( $n = 30$ ) were male with the same sport activity, while in Costa's study, the participants were from both genders (15 males and 15 females).

In this study, each position was measured only once by each of the raters and then they were recorded in questionnaires; therefore, we reported and analyzed single measures of lower and upper bound of 95% CI for ICCs limits. Table 2 shows that an average of several measurements improves the final reliability results and can be used in clinical application.

In this study for intrarater reliability there was one-week interval between the two tests which resolved memory effect in raters. A week is a good interval to avoid significant changes in shoulder posture of athletes while it has no significant clinical effect on the tests.

With respect to the limitations of this study, we can refer to the fact that a 30-minute session was held to explain the test condition and procedures to raters while in other studies such as McKenna et al.'s [13] the training session lasted for 4.5 hours and in Nijs et al.'s [14] it was 2 hours. Although both raters in our study had adequate experience in traditional test, longer sessions for familiarization of the raters could improve interrater reliability results.

Contradictory information about the effect of prior experience of examiners has been reported. For example, in Odom et al.'s study, despite a 4- to 7-year clinical experience of examiners, the reliability of the test was poor [15]. It is in contrast with Nijs et al.'s study that junior examiners obtained high interrater reliability [14].

## 5. Conclusion

In general, this study showed that applying the loads in scaption position of the LSST may improve the reliability of the test in sport men. Scaption with loading, as a functional position in the overhead athletes, is a reliable positioning and may be replaced with the third position of the traditional LSST. However, this study was a pilot and preliminary research on healthy athletes, and it investigated the effect of various loadings on MLSSST reliability in them and it is necessary to do complementary studies for the results to be useful for patients. Also, future studies with different subjects may benefit from loading in scaption. It is recommended that these positions be investigated in another athletic group using weights (especially with 2 and 4 kg weights) to engage more muscles.

## 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

# Design and Reliability of a Novel Heel Rise Test Measuring Device for Plantarflexion Endurance

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**Background.** Plantarflexion results from the combined action of the soleus and gastrocnemius muscles in the calf. The heel rise test is commonly used to test calf muscle endurance, function, and performance by a wide variety of professionals; however, no uniform description of the test is available. This paper aims to document the construction and reliability of a novel heel rise test device and measurement protocol that is suitable for the needs of most individuals. **Methods.** This device was constructed from compact and lightweight materials and is fully adjustable, enabling the testing of a wide variety of individuals. It is easy to assemble and disassemble, ensuring that it is portable for use in different settings. **Findings.** We tested reliability on 40 participants, finding excellent interrater reliability (ICC<sub>2,1</sub> 0.97, 95% CI: 0.94 to 0.98). Limits of agreement were less than two repetitions in 90% of cases and the Bland-Altman plot showed no bias. **Interpretation.** We have designed a novel, standardized, simple, and reliable device and measurement protocol for the heel rise test which can be used by researchers and clinicians in a variety of settings.

## 1. Introduction

Plantarflexion is achieved through the combined action of the soleus and gastrocnemius muscles, located in the calf [1, 2]. The gastrocnemius and soleus muscles share the Achilles tendon, the thickest and strongest human tendon, and are also known as the triceps surae [2]. The triceps surae is responsible for 80% of plantarflexion force [1–3]. Combined, the triceps surae stabilises the foot in weight bearing and produces a plantarflexion moment at toe-off, essential for forward momentum during gait [1, 2]. Therefore, sufficient plantarflexion strength and endurance are essential for basic mobility, such as walking and running [2, 4].

The heel rise test, also commonly described as heel raise test or calf rise test, was first developed in the 1940s [4, 5] to assess the function of the calf muscle-tendon unit and is now widely used by clinicians and researchers across disciplines. The test has commonly been used in neurology, cardiology, gerontology, orthopaedics, and sports medicine

[5]. Plantarflexion strength and endurance are often impaired after lower limb injury, especially after injury to the Achilles tendon [1, 6]. The heel rise test is commonly used not only during the initial assessment of these injuries, but also during rehabilitation to quantify treatment outcomes [5]. The test is often described as a test of calf muscle endurance, strength, fatigue, function, and performance [5, 7, 8]. It involves standing on one leg and rising and lowering the body by lifting the heel off the ground and then lowering it while maintaining a straight knee. The task is repeated without a break until the participant cannot complete it correctly or complains of pain or fatigue in the calf muscles [1, 4, 9]. The number of heel rises a participant can achieve is then recorded. The research literature commonly recommends 25 heel rises as the target norm of clinical performance for healthy subjects; however, much higher and lower values ranging from 7 to 48 have also been suggested in both research literature and musculoskeletal textbooks [5].

A systematic review by Hébert-Losier et al. [5] investigated the test variables and concluded that although this test is widely used in various health professions as a recommended assessment and rehabilitation tool, there is no uniform description of the test. The wide range of reported normative values in the literature may reflect the lack of standardization of the test. This further emphasises the need that the development of a standardized, reliable heel rise test is important for researchers and clinicians alike [1].

Research studies using the heel rise test commonly use customised devices developed to standardize the test and monitor test variables; however, these devices were often extremely complex, prohibiting use in everyday clinical practice or inadequately controlled confounding variables. In one study [10], a device was constructed that consisted of a box with an incline, a weight belt attached to the waist of the participant, and a linear encoder attached to the heel of the shoe to measure plantarflexion endurance. The linear encoder, connected to a computer measuring system, measured the time and length of the heel displacement of the heel rise. A second study [11] used a device with a light beam attached to vertical rods at a fixed height of 5 cm above the heel. The device emitted an audible click when the participant's heel passed the 5 cm height when rising onto the toes to raise the body. However, no feedback was provided to participants about the actual height of each heel rise and therefore a maximum or standardized heel rise was not necessarily achieved. A further disadvantage of this device was that the height was not adjustable and therefore could not allow individualized testing. Additionally, the device required an electrical current to function. A third study [12] used a simple, portable device that positioned a rod horizontally across the foot for participants to touch with the anterior aspect of the arch of the foot when the heel was raised. Some adjustability was allowed by four preset holes in the device in which the rod was placed to enable selection of four different heights. Although the device allowed for some ability to individualize the test, adjustability was incomplete. Furthermore, the safety of placing the rod at the front of the foot is questionable: in the event that an individual lost balance during the test, the rod could prevent stepping off the device with ease.

Clinically, the heel rise test is often employed for assessment and rehabilitation purposes without using a device at all. This may possibly be due to the complexity of devices currently available. However, a standardized device and protocol that is suitable for all individuals is essential to monitor and replicate the heel rise test with consistent outcomes [1, 5]. When the aim is to use the device in clinical settings, it should be simple, cheap, reliable, and clinically accessible [5]. A universally accepted standardization of the heel rise test and consequently a device that allows for standardized, reliable, and individualized evaluation protocols is not currently available. The Ankle Measure for Endurance and Strength (AMES) device (IP Australia; innovation patent application number AU2012101251) and measurement protocol was created to provide the platform for such results. This paper aims to document the construction and reliability of the device.

## 2. Methods

**2.1. Construction of the Device.** The construction of the AMES device is shown in Figures 1 and 2. We used a 44.5 cm × 40.5 cm ( $L \times W$ ) wooden platform as the base. To the bottom we glued two small wooden blocks each of 31 cm × 4.2 cm ( $L \times W$ ) to lift the platform slightly off the ground and to allow fixation of the other parts to the platform.

On top of the platform, two medium sized "L"-brackets were placed parallel to each other on both sides, spaced 23.1 cm apart and 12.5 cm from the back of the platform. The setting of the brackets as illustrated in Figures 1 and 2 has proven to fit the foot length and width of all tested individuals. However, the brackets can be moved forwards, backwards, or wider apart to accommodate individuals' foot length and width. The L-brackets were secured onto the platform with two small "G"-clamps of 12 × 8 cm ( $L \times W$ ).

A 12 mm thick elastic band, on which the participant stands, was placed horizontally between the L-shape brackets and was held in place by two small spring clamps. The elastic band was 30 cm long and cut from a 4-meter strip. The L-shape brackets, the elastic band, the G-clamps, and the spring clamps are adjustable so that the device is able to fit the needs of a wide variety of individuals. The spring clamp facilitates the sliding of the elastic band up and down the L-brackets to cater for each individual's maximum heel rise height. Adjustment of the spring clamps and the elastic band was performed while the individual was standing on the platform and performing a heel rise. The platform, the small wooden blocks, the L-shape brackets, the G-clamps, the spring clamps, and the elastic band were all constructed from compact and lightweight materials and attached to each other but all were removable to facilitate easy repeated assembly and disassembly of the device. The total price of the current device and all its components was approximately \$25.00 USD.

**2.2. Testing Set-Up Protocol.** Before performing the test, the device was adjusted to the individual's maximum heel rise as follows.

- (1) The participant placed the heel, barefoot, on the elastic band between the L-brackets with the individual's foot pointing to the front of the platform (Figure 3).
- (2) The participant performed a maximum heel rise, with extended knee, with the nontesting leg flexed and suspended in the air and the fingertips of one hand on the wall for balance (Figure 4).
- (3) The examiner adjusted the elastic band by sliding the spring clamps up or down until the elastic band was just clear of the heel and in a horizontal position. The participant lowered their heel back onto the platform.
- (4) The participant performed another heel rise to confirm their maximum heel rise; for example, the participant cleared the elastic band on each heel rise during the test.
- (5) Lastly, the test was conducted with the participant rising and lowering their heel to the beat of a metronome

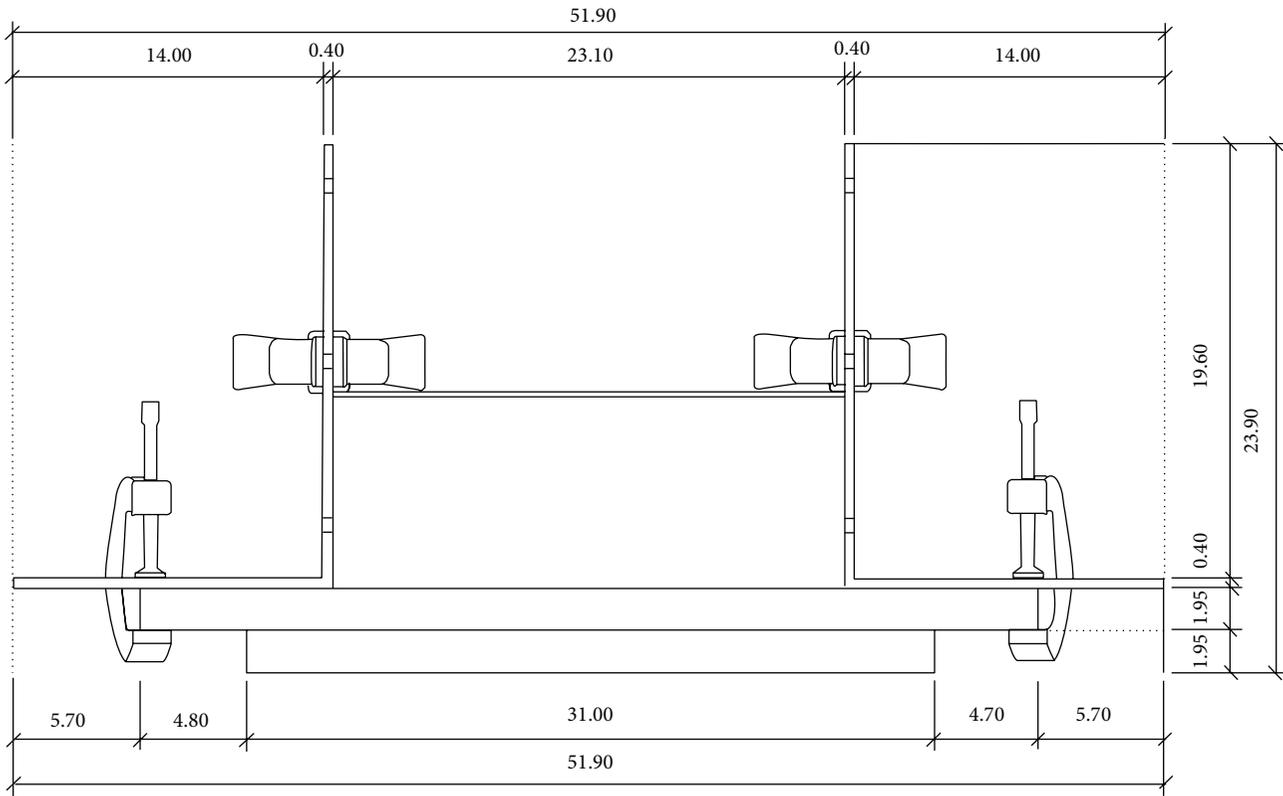


FIGURE 1: Front view of the Ankle Measure for Endurance and Strength (AMES) device.

until the participant can no longer perform a heel rise or fails to perform a technically correct heel rise (maximum heel raise with extended knee, clearing the elastic band, and only fingertips of one hand on the wall for balance) on two consecutive occasions.

**2.3. Reliability of the Device.** Following the design of the device, we tested it for reliability.

**2.3.1. Participants.** The study was performed at the University of Sydney in the Arthritis and Musculoskeletal Research Lab. We recruited participants through advertisements on university noticeboards. A total of 40 participants who met the inclusion criteria for the study were enrolled. Inclusion criteria were as follows: (1) healthy adults over 18 years and (2) no current ankle injury or chronic ankle pain. The only exclusion criterion was any present condition, such as vestibular problems or current lower limb injury that would affect balance ability. Participants were asked if they had been involved in excessive exercise during the 72 hours prior to the test occasion, other than their regular exercise. Ethics approval was obtained from the Human Research Ethics Committee at the University of Sydney (Protocol number 07-2011/13973) and all participants gave their written informed consent prior to commencement of the study.

**2.3.2. Randomization.** Participants performed one trial and the order of test leg was randomly assigned using a web-based randomization program; <http://www.randomizer.org/>.

**2.3.3. Blinding.** The test was performed once and observed by two blinded raters (A.I/A.O). A screen separated the raters from each other to minimise bias. For reliability testing purposes, an independent examiner (A.S), shielded from the raters by a second screen, was present to provide the participants with feedback and correct any errors of technique that occurred during the trial. This feedback was given without verbal cues so that the raters were not alerted to potential errors in performance and thus minimise confounding. The independent examiner would provide a tap on the independent examiner's own ankle indicating participants should rise higher on the next heel rise, a tap on the knee indicated participants should keep their knee straight during the next heel rise, and a tap on the hand indicated they were leaning too much into the wall. Participants were instructed to keep eye contact with the independent examiner throughout the test.

**2.3.4. Methods.** All participants completed a questionnaire including demographic information and medical history related to their ankle and knee to ensure they had no underlying ankle or knee problems.

Participants performed the single leg heel rises barefoot while maintaining an extended knee of the test leg throughout the trial with the nontest leg flexed and suspended in the air. Participants performed a maximum heel rise [1, 5], clearing the elastic band, and then lowered their heel to the platform at the beat of the metronome. As used in previous research by Haber et al. [12], we chose a rate of 46 beats/min (23

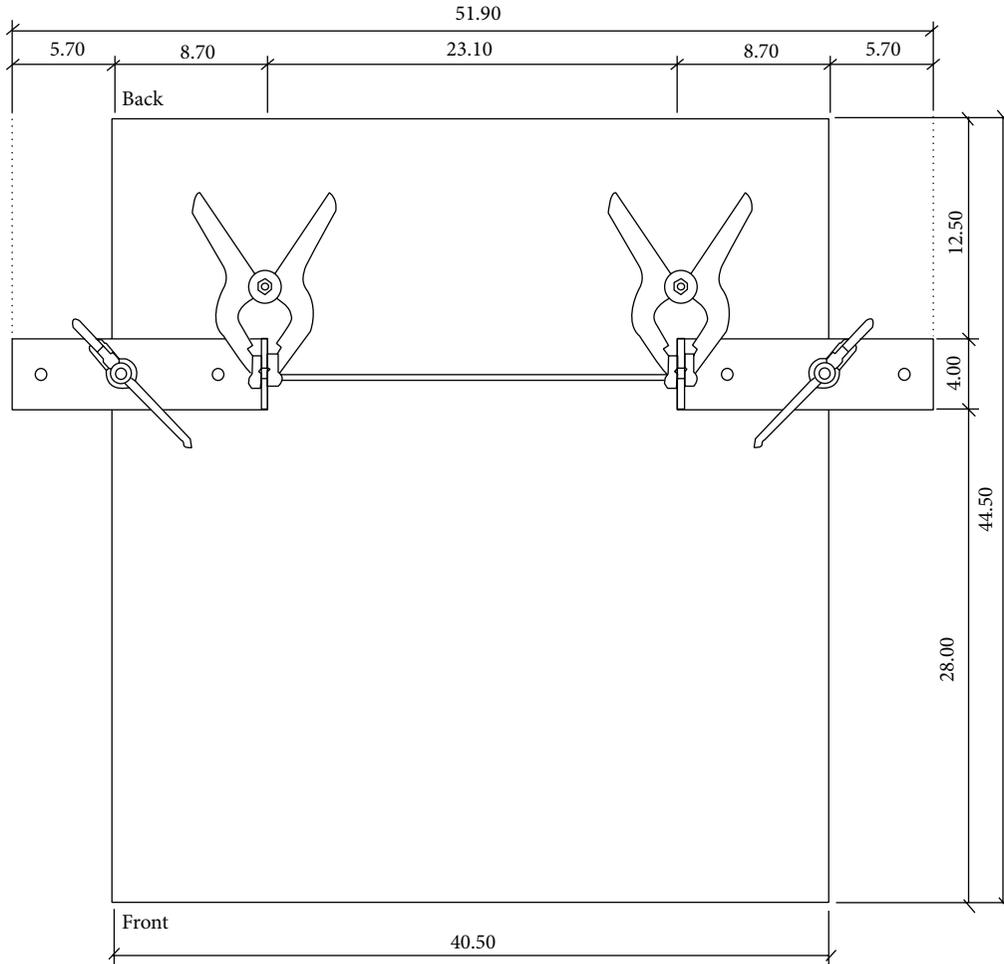


FIGURE 2: Top view of the AMES device.



FIGURE 3: Testing set-up (step 1).

heel rises/min) as set by a metronome. On the sound of the first beat of the metronome, the participant lifted the heel and on the following beat lowered the heel. Participants continued the heel rises and were encouraged to do so by the independent examiner until they could no longer perform a technically correct heel rise. For the duration of the trial, participants were allowed to place the fingertips of one hand on the wall for balance.

The raters would stop counting when the participant on 2 consecutive occasions

- (i) could no longer achieve the maximal rise (clear the elastic band) and/or
- (ii) placed too much weight on the wall (i.e., hip flexion) and/or
- (iii) flexed their knee during the movement and/or
- (iv) missed a beat of the metronome and/or
- (v) wished to stop.

One rater (A.O) adjusted the device for each participant to ensure that the same method was used for all participants and to eliminate variability in the setting of the device that could affect the results. The elastic band, 30 cm long, was replaced after every 5 participants to ensure similar elasticity for each participant.

**2.3.5. Data Analysis.** Data analysis was performed using IBM SPSS statistics 19.0. Data were tested for normality using the Kolmogorov-Smirnov with Lilliefors significance correction and data are reported accordingly. Interrater reliability was

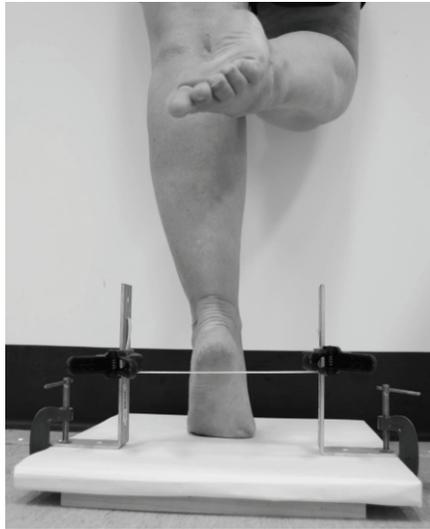


FIGURE 4: Testing set-up (steps 2–5).

TABLE 1: Participant characteristics (N = 40).

Gender		
Female : male		19 : 21
Age (yrs)		
Mean (SD)		24 (6.2)
Height (cm)		
Mean (SD)		174 (12.3)
Weight (kg)		
Mean (SD)		68 (9.3)
Involved in regular exercise <sup>‡</sup>		
Yes : no		5 : 34
History of foot/ankle injury		
Yes : no		23 : 17
History of calf injury		
Yes : no		1 : 39
History of knee injury		
Yes : no		10 : 30

<sup>‡</sup>Missing data for one participant.

calculated by intraclass correlation coefficient (ICC) and limits of agreement. ICCs were calculated using the two-way random effects model (ICC<sub>2,1</sub>) with 95% confidence intervals. Standard error of measurement (SEM) was calculated using the method described by Portney and Watkins [13]; for example,  $SEM = SD \sqrt{1 - ICC}$ , where SD is the standard deviation of the set of observed test scores. A Bland-Altman plot was constructed to determine if there was bias in the measures.

**2.3.6. Results.** We enrolled 40 participants in the study and all participants completed testing. Their demographic variables are presented in Table 1. The mean age was 24 years and 32 participants were aged between 21 and 23 years. Twenty-one participants reported a history of foot and ankle injury, one reported a history of calf injury, and ten reported a history of knee injury. Of the foot and ankle injuries, 17 were

TABLE 2: Limits of agreement (N = 40).

Difference between raters	Frequency	Valid percent	Cumulative percent
0	14	35.0	35.0
1	14	35.0	70.0
2	8	20.0	90.0
3	1	2.5	92.5
4	1	2.5	95.0
14	2	5.0	100
Total	40	100	

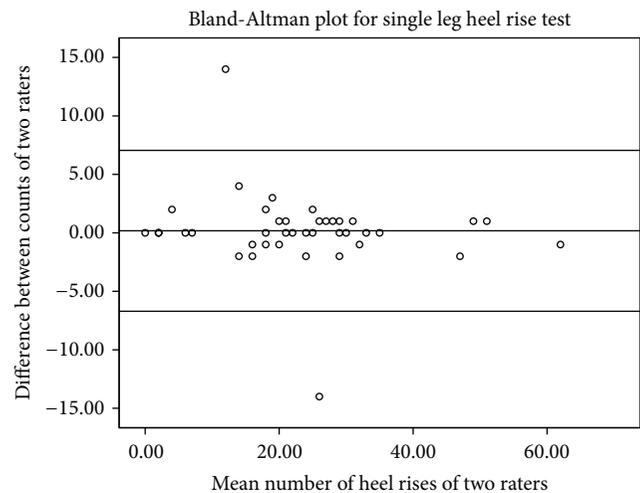


FIGURE 5: Bland-Altman plot.

due to ankle sprain. The mean number of heel rises in this healthy population was 23 (SD 13.3) repetitions. Within our cohort, four individuals completed more than 45 repetitions and most individuals completed fewer than 25 repetitions (63%). There was no noticeable difference between males and females. Most trials required approximately 2-3 minutes to complete and no adverse events were reported as a result of the heel rise test or the device.

Excellent interrater reliability was found between the two trials (ICC<sub>2,1</sub> 0.97, 95% CI: 0.94 to 0.98). The mean difference between the trials was 0.15 repetitions, the standard error of measurement was 2.4 repetitions (SEM% = 10.4%), and the limits of agreement were less than two repetitions in 90% of the cases (Table 2). The Bland-Altman plot (Figure 5) showed no bias.

### 3. Discussion

The heel rise test is a common test of function of the calf muscle-tendon unit and is used as a test to assess calf muscle endurance and strength [5, 7, 8]. However, until now, there was no standardized device or method to measure calf muscle endurance. The device described in this paper aimed to address the limitations of previous devices, such as the use of computers and linear encoders [10], nonadjustable heights

[11, 12], light beams [11], electrical currents [11], and safety of the device [12].

The current study found that, compared to other devices, the current device is safe, cheap, easy to use, and portable. The height of the device is fully adjustable, the elastic band is placed underneath the heel, and the device does not require additional computers or an electrical current to function. The device has excellent interrater reliability and is cheap, portable, and easy and quick to use. The heel rise test can be used in a broad range of settings and has previously been employed during the initial assessment following injury to evaluate treatment progress [5] and to determine return to sport readiness following ankle syndesmosis injury [14, 15] and in research studies investigating risk factors for ankle injury [16]. The device described in this paper may be used as a standardized tool for these purposes in a broad range of clinical populations.

Previous literature reported a wide range of norms using the heel rise test in healthy populations [5] possibly due to the lack of standardization of the test. In a young and healthy population, using the current standardized device and protocol, the mean number of maximum heel rises performed was 23 (SD 13.3). This value may be utilised in the clinic and future research studies as a clinical reference. Although research suggested that the number of heel rises performed varies with age and sex [17], we did not find any notable differences between genders in our study. We were unable to investigate the influence of age on the number of heel rises performed with the current device as the majority of our participants were aged between 21 and 23. Future, larger studies may use this standardized device and protocol to establish norms across the life span.

The current device may be further developed to improve the applicability of the device. Firstly, the current device has no fixation at the front of the foot, which may cause the foot to slide forward during the testing and possibly influence the maximum heel rise height a participant can achieve. Although there were no adverse events due to the device or the tests, the L-shaped brackets may be replaced with curved brackets to minimize risk of injury due to the sharp edges. A limitation to the current reliability testing was the use of an independent examiner to provide nonverbal cues. In daily practice, verbal cues are given by the clinician. It is unknown if this method has influenced the results in any way; however, due to the research design we were unable to change this. Finally, the intrarater reliability has not yet been established, but may be conducted to investigate reproducibility of the test.

#### 4. Conclusion

The heel rise test is widely used for clinical assessment and subsequent decision-making regarding rehabilitation progression by a variety of health professionals; however, to date there has been no standardized heel rise test device that is reliable, portable, and easy to use. We have constructed and tested a novel, standardized, and simple device with a standardized measurement protocol for the heel rise test and demonstrated its excellent reliability.

#### Disclosure

This research received no specific grant from any funding agency in the public, commercial, or not-for-profit sectors.

#### Conflict of Interests

Claire E. Hiller, Adam Imer, Aldrin Ocsing, Joshua Burns, and Kathryn M. Refshauge declare that there is no conflict of interests regarding the publication of this paper. Amy D. Sman filed an innovation patent application for the device described in this paper with IP Australia. This received application no. AU2012101251 on September 13, 2012.

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## Research Article

# Reliability in the Parameterization of the Functional Reach Test in Elderly Stroke Patients: A Pilot Study

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**Background.** Postural instability is one of the major complications found in stroke survivors. Parameterising the functional reach test (FRT) could be useful in clinical practice and basic research. **Objectives.** To analyse the reliability, sensitivity, and specificity in the FRT parameterisation using inertial sensors for recording kinematic variables in patients who have suffered a stroke. **Design.** Cross-sectional study. While performing FRT, two inertial sensors were placed on the patient's back (lumbar and trunk). **Participants.** Five subjects over 65 who suffer from a stroke. **Measurements.** FRT measures, lumbosacral/thoracic maximum angular displacement, maximum time of lumbosacral/thoracic angular displacement, time return initial position, and total time. Speed and acceleration of the movements were calculated indirectly. **Results.** FRT measure is  $12.75 \pm 2.06$  cm. Intrasubject reliability values range from 0.829 (time to return initial position (lumbar sensor)) to 0.891 (lumbosacral maximum angular displacement). Intersubject reliability values range from 0.821 (time to return initial position (lumbar sensor)) to 0.883 (lumbosacral maximum angular displacement). FRT's reliability was 0.987 (0.983–0.992) and 0.983 (0.979–0.989) intersubject and intrasubject, respectively. **Conclusion.** The main conclusion could be that the inertial sensors are a tool with excellent reliability and validity in the parameterization of the FRT in people who have had a stroke.

## 1. Background

Stroke is the third leading cause of death and the leading cause of long-term neurological disability in the world [1, 2]. In Europe there are 250 strokes per 100,000 people each year, and this trend is worsening with time [2, 3]. Patients who survive a stroke often suffer a severe disability that causes major limitations in activities of daily living [2].

Postural instability is one of the major complications found in people who survive a stroke [2]; between 50% and 70% of patients who return to their homes from hospital or a rehabilitation centre experience falls [4]. In addition, a high percentage of patients experience greater difficulty to stand up, postural exaggeration, constant rebalancing in the sagittal and frontal plane, reduction of ability to be supported on the affected limb, and, therefore, an increased risk of falling [2, 5].

The instrumentalisation or parameterisation of functional test is to analyse the development of them acquiring parameters that can be used in clinical practice and basic research [6]. The use of standardised instruments measuring the health status of patients has been promoted in all fields of medicine applied to help establish and implement effective treatment strategies [7]. Because of their portability, reliability, and size, inertial sensors are instruments able to acquire kinematic variables of any gesture or movement [8].

In basic research, several studies have used inertial sensors to analyse the different kinematic variables that the gait can be decomposed [6, 8–11]. In clinical practice, this instrument has been used as a feedback tool for improving the sway on balance and ambulation tests [8].

The functional reach test (FRT) [12–16] is a clinically accepted tool to measure the semistatic balance of a subject

because of its simplicity, reliability, economy, and portability. It is based on analysing the limits of stability in the absence of external shocks, assessing the maximum displacement, intentionally, which can reach a subject without losing balance. Thus, it integrates biomechanics, postural control, and subjective perceptions and correlates results of the greater chance of falling. This tool has been used to analyse the balance in patients suffering from Parkinson's disease, physical frailty, vestibular dysfunction, and stroke [17].

There are no studies in which FRT was instrumentalised by the inertial sensors in patients who have had a stroke.

The aim of this study is to analyze the reliability and validity of the parameterisation of FRT by using inertial sensors to record kinematic variables in subjects who have had a stroke. Our hypothesis is that the IS will be reliable instruments for kinematic study of the FRT.

## 2. Method

**2.1. Design and Participants.** In the cross-sectional study, participants met the following inclusion criteria: stroke verified as defined by the World Health Organization [18], independence walking for 10 m without the need to use physical support or the support of an auxiliary person, at a velocity equal to or less than 0.8 m/s, and the ability to remain standing with or without assistance for more than 30 seconds. People excluded from the study had experienced the following exclusion criteria: less than 65 years, with cardiovascular, respiratory, orthopedic, or severe metabolic problem, limitations in ambulation, serious problems of communication or understanding, history of a secondary neuronal pathology, and not giving informed consent.

Ethical approval for the study was granted by the ethics committee of the Faculty of Health Sciences, University of Málaga. This study was conducted in Accordance with Ethical Principles for Medical Research Involving Human Subjects (Helsinki declaration 2008).

Before performing the functional reach test, the information sheet and the informed consent were presented to each participant, in which the course of the study was explained. They were informed too about the voluntary nature thereof and the facility to leave the study at the moment they wanted, as well as the protection of their personal data according to the Organic Law of Protection of Data Personal 19/55.

**2.2. Inertial Sensors.** The inertial sensors used in this study were the model InertiaCube3™ InterSense Inc. (Bedford, MA, USA) with a sampling frequency of 180 Hz.

The InertiaCube3 is a sensor based on microelectromechanical systems (MEMS) technology without involving casters, which could generate noise, inertia forces, and mechanical failures. The InertiaCube measures nine simultaneously physical properties, that is to say, angular rates, linear accelerations, and magnetic field components along the three axes (Yaw, Pitch, and Roll). Miniature vibrating elements are

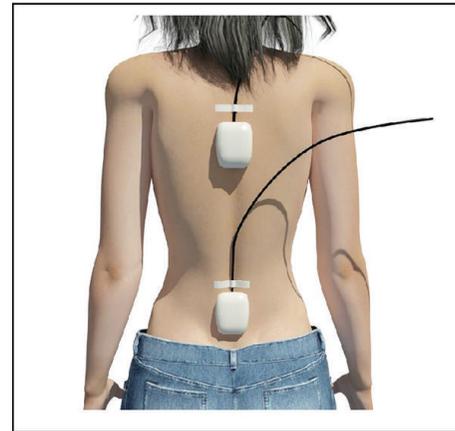


FIGURE 1: Position of the inertial sensors on the back of patients.

used for measuring all the components of the angular velocity and linear accelerations.

**2.3. Functional Reach Test (FRT).** The subjects were standing, parallel to a wall, close to but not touching, and with their feet open to shoulder height. The shoulders were positioned at 90 degrees of flexion, with elbows and hands extended. In this position, on a yardstick, the assessor recorded, at the third metacarpal head, the starting position. The subject held this position for three seconds. Then, without moving their feet off the ground, the participant performed hip flexion by moving their trunk forward and reaching as far as they could without taking a step. At this point, the assessor located the position of the third metacarpal. Subsequently, the participant returned to the starting position and remained still for a further three seconds to clearly differentiate the end of the movement. The difference in centimeters between the first and second mark during the functional reach test was FRT value.

The FRT was performed three times, but the average of the last two was considered the FRT measure.

Before starting, it was explained to the participants the movement execution. The subjects could take all the tests considered necessary for better understanding of the test, which has a reliability of 0.81 [19].

Two inertial sensors were placed, one in the centre of mass and the other in the trunk (Figure 1), which made a cinematic record during test execution. Registration of the kinematic variables of test development was carried out throughout the test over the initial and final three seconds. This served the subject to reach the starting position and the researcher as a reference to analyze the data. The analysis was performed with performance that had greater distance in the FRT.

The sensor was placed so that the origin of the coordinates (X, Y, Z) (0, 0, 0) was positioned in the posteroinferior left corner (Figure 2).

After completion of data collection, a blinded investigator performed offline extraction of variables from each of the graphs generated following completion of each test.

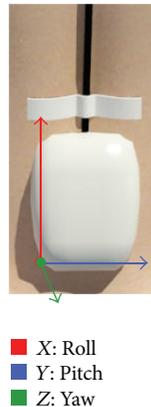


FIGURE 2: Origin of the coordinates ( $X, Y, Z$ ) in the inertial sensor.

**2.4. Outcome Measures.** The outcomes measures extracted from FRT or Duncan test were as follows: *FRT distance*: the distance in centimeters that the subject is able to achieve during the realization of the FRT; *maximum angular lumbosacral/thoracic displacement FRT*: the angular variation on the pitch in the subject during the performance of FRT axis; this amplitude is considered from the time the test begins until peaking imbalance before returning to the starting position; *time maximum angular lumbosacral/thoracic displacement FRT*: the time that the subject takes to reach the peak during running the FRT; *time return starting position*: the time that the subject takes to return to a starting position from reaching the peak; *total time FRT*: the time the subject takes from start to perform the FRT until he returns to his starting position. All variables listed above were extracted from the registry of the inertial sensor in the pitch axis.

Subsequently, using the extracted data, the following variables were calculated: *average speed FRT*: average rate at which the subject performs all the FRT; *maximum angular lumbosacral/thoracic displacement speed FRT*: the average velocity at which the subject reaches the peak from the start of carrying out the FRT; *starting to return position speed*: the average rate at which the subject performs the movement back to the starting position to the maximum peak; *average acceleration FRT*: the mean acceleration at which the subject carries out the FRT; *maximum angular lumbosacral/thoracic displacement average acceleration FRT*: the average acceleration at which the subject develops the test from the beginning until he reaches the peak; *acceleration average return starting position FRT*: the average acceleration of the subject from reaching the peak until the return to his starting position.

In addition, the *mean and standard deviation of  $X, Y,$  and  $Z$*  were calculated in the highest, lowest, and average speed and acceleration in both sensors, just as the *mean and standard deviation in the resultants* of displacement and resultants of minimum and maximum speed and acceleration. Resultant is calculated previously by finding the square root

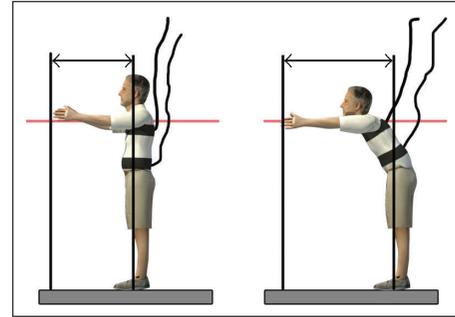


FIGURE 3: Performing of FRT.

of the sum of the squares of the three axes in movement, the maximum and minimum of speed and acceleration of FRT.

The outcome analyzed was the highest value obtained during the performance of the three repetitions of the test.

**2.5. Procedure.** Before beginning, the test was explained in detail to participants and the participants signed the informed consent. Sociodemographic data and anthropometric measures of each subject were collected. To improve the description of the sample, participants completed the Barthel Index (BI), the Stroke Impact Scale-16 (SIS-16), and the Canadian Neurological Scale (CNS). The reliability of these tools is Kappa = 0.93 [20, 21], Kappa = 0.76 [22], and ICC = 0.70 to 0.92 [23], respectively.

Functional reach test (FRT) or Duncan's test (Duncan 1990) was performed. During execution, the subjects carried two inertial sensors, one was placed at the level of  $L_5-S_1$  (lumbar) and the other in  $T_7$  (trunk) (Figure 3). Two researchers monitored the implementation of the test and performed the analysis of the results independently. Under the supervision of individual researchers, the test was performed three times, considering the average of the last two repetitions the measure of the FRT.

After analyzing the data obtained in the kinematic registration by inertial sensors, a number of direct and indirect variables were obtained. Direct variables obtained were time and displacement between each of the points of the three intervals. And the indirect variables, calculated thereafter, were the speed, acceleration, and the resultant.

**2.6. Data Analysis.** After completing the sample, a descriptive analysis was made, which included anthropometric measurements and the results of various self-administered questionnaires specifically designed for patients with neurological affectations. A descriptive analysis of all kinematic outcomes recorded by the two inertial sensors (trunk and lumbar) was developed and the average range achieved in the FRT.

After performing the normality of the variables by Kolmogorov-Smirnov (KS) test, the results were compared, records between trunk and lumbar, both directly measured outcomes (time and displacement) and outcomes obtained

TABLE 1: Descriptive values of participants.

	Minimum	Maximum	Mean	SD
Age	69	92	76.7	
CNS	8.0	9.0	8.500	0.4082
BI	85	100	92.50	6.455
SIS_16	57	75	67.00	7.832
<i>N</i> valid (according to the list)	4			

CNS: Canadian Neurological Scale; BI: Barthel Index; SIS-16: Stroke Impact Scale-16.

indirectly (velocity, acceleration, and resultant). For parametric outcomes, Student's *t*-test was used, and for the nonparametric, Wilcoxon's test was used. The index of significance was established in less or equal to  $P = 0.05$  values.

Reliability measures were calculated by analysing the internal consistency (intraclass correlation coefficients) were calculated for intrarater and interrater reliability) of the measures with 95% confidence interval of each outcome variable. The reliability was calculated in the functional reach test and the outcomes measured by the IS (time and displacement). The reliability of the indirect variables was not calculated (velocity and resultant acceleration), because its value is determined by the reliability of the direct measures. The levels of reliability were excellent ( $ICC > 0.80$ ), good ( $0.80 > ICC > 0.60$ ), moderate ( $0.60 > ICC > 0.40$ ), and poor reliability ( $ICC < 0.40$ ) [24].

The Statistical Package for the Social Sciences (SPSS) (version 17.0 for Windows, Illinois, USA) was used to represent the statistical analysis.

### 3. Results

Table 1 shows the anthropometric and demographic data of the participants, as well as the values of the various specific tests that each participant completed as well as the values of the various specific tests which were intended to identify the degree of involvement of the patient as a result of the stroke.

Table 2 presents the description of the kinematic variables of the FRT based on their placement in the centre of mass and thorax, distance functional reach test, and the number of participants. Three intervals of movement based on the following points were considered: beginning of the test, maximum angular displacement, and end of the test. The outcomes calculated in each of these intervals were time, displacement, velocity, and acceleration. It can be seen in Table 2 the maximum, minimum, mean, and standard deviation of each outcome.

Table 3 shows the resultant displacement, resultant in the maximum and minimum velocity, and the resultant in maximum and minimum acceleration in the FRT, and minimum of speed and acceleration. All the outcomes previously mentioned have been presented as mean and standard deviation of the sum of the participants relating to *X*, *Y*, *Z* of each of the sensors and the difference between them.

Table 4 presents the intrasubject and intersubject reliability of the outcomes measured directly in the parameterisation of the FRT. Intrasubject reliability values observed in the use of inertial sensors are all located above 0.820, ranging from 0.829 (time B.C lumbar area) to 0.891 (A.B displacement of the trunk). Likewise, the observed intersubject values range from 0.821 (time B.C lumbar area) to 0.883 (B.C trunk displacement). On the other hand, the reliability of the FRT was 0.987 (0.983–0.992) and 0.983 (0.979–0.989) intersubject and intrasubject, respectively.

### 4. Discussion

After analyzing the data obtained, it shows how the inertial sensors are a reliable, specific tool for the parameterization of a functional reach test in a sample of stroke patients who suffer from problems of imbalance, and we can say that the aim of this study was achieved. Furthermore, based on the results, the hypothesis set out at the beginning is confirmed.

No study was found that uses inertial sensors to parameterize the FRT. However, these instruments have been used themselves for the kinematic analysis of other tests [6, 25–31], and these were static [27, 31], semistatic [28, 29], or dynamic [6, 25, 26].

Reliability levels observed in the present study could be categorized as excellent [24] in base of the results that the intraobserver reliability ranges between 0.829 and 0.878 and the interobserver between 0.821 and 0.883 (Table 4). These results are in accordance with all the studies consulted [6, 25, 27–31], except the study of Lugade et al. [26], which showed reliability levels over 0.9.

These results are consistent even if we consider some details of registration, such as the position of the sensor, where the values of obtained reliability ( $ICC: 0.835–0.877$  (trunk) and  $0.829–0.878$  (lumbar)) are comparable with other studies that share the sensor location, as in the study of Kavanagh et al. [25], who analyzed the reliability of IS to analyze the progress at different speeds (slow, determined by the participant, and fast), placing, among others, IS in trunk and lumbar, which achieved a reliability of 0.83–0.93 (trunk) and 0.78–0.92 (lumbar) during performance on the speed determined by the participant.

Considering the observer test, an intraobserver reliability of 0.829–0.878 and interobserver of 0.821–0.883 (Table 4) were noted, which are similar to those reported by Kavanagh et al. [25]:  $ICC$  values (95% IC) of 0.84–0.91 (intraobserver) and from 0.85 to 0.93 (interobserver).

The reliability of runtime testing, total or different partials, demonstrated excellent intrasubject reliability with  $ICC$  values (95% IC) of 0.863–0.877 (trunk) and 0.829–0.867 (lumbar) (Table 4). These values are comparable with those presented by Duffy et al. [21] in the other semistatic test (sit to stand), where the reliability values were 0.89 (0.78–0.94), 0.83 (0.67, 0.92), and 0.8 (0.61, 0.9) for the total time of the test, stand to sit time, and sit to stand time, respectively. In addition, as regards the reliability of the time keeps in the inter-observer analysis with values ranging between 0.821 and 0.858, respectively.

TABLE 2: Description of the kinematic variables of FRT depending on the placement of the sensor.

	Minimum	Maximum	Mean	SD	
Functional reach test distance	11	15	12.75	2.06	
Trunk	Time A_B (s)	7.05	9.97	8.71	1.5
	Displacement A_B (°)	4.73	20.55	13	7.94
	Speed A_B (°/s)	0.67	2.06	1.49	5.29
	Acceleration A_B (°/s <sup>2</sup> )	0.10	0.21	0.17	3.53
	Time B_C (s)	2.09	12.98	6.96	5.53
	Displacement B_C (°)	5.16	12.88	9.8	4.1
	Speed B_C (°/s)	2.47	0.99	1.41	0.74
	Acceleration B_C (°/s <sup>2</sup> )	1.18	0.08	0.20	0.13
	Time A_C (s)	12.06	20.03	15.68	4.03
	Displacement A_C (°)	6.2	20.58	13.5	7.19
	Speed A_C (°/s)	0.51	1.03	0.86	1.78
	Acceleration A_C (°/s <sup>2</sup> )	0.04	0.05	0.05	0.44
Lumbar	Time A_B (s)	5.55	12.15	8.56	2.93
	Displacement A_B (°)	4.61	11.1	7.49	3.25
	Speed A_B (°/s)	0.83	0.91	0.88	1.11
	Acceleration A_B (°/s <sup>2</sup> )	0.15	0.08	0.10	0.38
	Time B_C (s)	2.38	13.37	8.13	6.019
	Displacement B_C (°)	6.62	14.93	9.74	3.68
	Speed B_C (°/s)	2.78	1.12	1.20	0.01
	Acceleration B_C (°/s <sup>2</sup> )	1.17	0.08	0.15	0.02
	Time A_C (s)	11.98	20.32	16.7	3.7
	Displacement A_C (°)	9.22	22.2	14.98	6.49
	Speed A_C (°/s)	0.77	1.09	0.89	1.75
	Acceleration A_C (°/s <sup>2</sup> )	0.06	0.05	0.05	0.47
N valid (according to the list)		4			

A: beginning of the FRT; B: maximum angular displacement; C: end of the FRT.

In analyzing the data from each inertial sensors, it can be observed how the registered values of each sensor are very broad with respect to the standard deviations presented (Tables 2 and 3), in both the trunk and the lumbar. Moreover, the different registration observed between the lumbar and the trunk sensor (Table 3) confirms that the inertial sensors, in addition to being sensitive, are tools with high specificity. These results are consistent with other studies that have also found inertial sensors as instruments with high sensitivity and specificity [29, 31].

On the other hand, in analyzing the reliability of the measures of the functional reach, it is observed, with regard to people who have suffered a stroke, that the reliability levels are greater than 0.98 (ICC: 0.987 (0.983–0.992) and 0.983 (0.979–0.989) for intra- and interobserver). These reliability levels are not consistent with those observed in previous studies, where FRT reliability levels were 0.86 [15] and 0.64–0.74 [32]. The difference between levels of reliability can be because, in the present study, participants were stroke victims, which determine the imbalance in the functional reach (12.75 (11–15) cm), thus limiting, in turn, the variability

of the measuring and improving the reliability. However, in other studies consulted, the study subjects are patients suffering from Parkinson's disease [32] or are healthy older women [15]. These participants achieved values in FRT in excess of those obtained in the present study: 33.54 ( $\pm 7.36$ ) [32] and 17.1 ( $\pm 6.7$ ) [15] FRT.

Age also appears to be a negative determinant of the results obtained in the FRT. Several studies on stroke patients have been published and the results of functional reach are not comparable with those observed in the present study [12, 13], since in both cases the mean values observed in these studies are double (24.6–25.6 cm [13] and 28.0 cm [14]) those presented in Table 2 (12.75 cm  $\pm$  2.06). The difference may reside, as we indicated earlier, in the average age of participants, 56.3–56.8 years [12] and 55.9–56.3 years [14], respectively. However, when participants have a similar age, the results observed are consistent with the present study. The values presented by Palsbo et al. [33] and DeWaard et al. [15]—values in the FRT of 2.7–17.0 cm [33] and 17.1 (8.9–26.0) cm [15]—are similar to those obtained in the present study (2.06  $\pm$  12.75 cm); the mean age in each of the studies

TABLE 3: Mean and standard deviation of the records of each of the sensors and differences between them.

	Trunk			Lumbar			Mean difference		
	X	Y	Z	X	Y	Z	X	Y	Z
Resultant displacement		<b>33.87</b> (±6.71)			<b>36.45</b> (±14.01)			<b>1.86</b> (±23.64)	
Speed mean	<b>1.72</b> (±0.21)	<b>26.37</b> (±7.20)	<b>23.30</b> (±9.54)	<b>1.74</b> (±0.75)	<b>24.25</b> (±7.98)	<b>20.10</b> (±5.59)	<b>0.01</b> (±0.91)	<b>-2.32</b> (±2.83)	<b>-2.31</b> (±13.18)
Maximum speed	<b>-0.60</b> (±0.74)	<b>10.27</b> (±4.15)	<b>9.77</b> (±1.52)	<b>1.77</b> (±1.18)	<b>10.53</b> (±6.50)	<b>8.38</b> (±1.27)	<b>2.00*</b> (±0.75)	<b>0.64</b> (±5.13)	<b>-1.32</b> (±2.40)
Minimum speed	<b>-2.32</b> (±0.92)	<b>-16.10</b> (±3.10)	<b>-13.53</b> (±8.08)	<b>0.03</b> (±1.56)	<b>-13.72</b> (±4.99)	<b>-11.72</b> (±4.39)	<b>2.00</b> (±1.54)	<b>2.97</b> (±5.72)	<b>0.99</b> (±10.80)
Maximum resultant speed		<b>14.34</b> (±3.67)			<b>13.86</b> (±5.89)			<b>-0.12</b> (±5.66)	
Minimum resultant speed		<b>21.74</b> (±6.19)			<b>18.53</b> (±4.74)			<b>-3.05</b> (±3.29)	
Mean acceleration	<b>2.17</b> (±1.28)	<b>3.27</b> (±1.48)	<b>6.53</b> (±1.32)	<b>1.27</b> (±2.51)	<b>-0.54</b> (±4.50)	<b>4.75</b> (±1.01)	<b>-0.24</b> (±1.50)	<b>-3.96</b> (±4.53)	<b>-1.71</b> (±1.28)
Maximum acceleration	<b>-0.81</b> (±1.44)	<b>-2.90</b> (±3.07)	<b>95.40</b> (±8.54)	<b>-0.00</b> (±0.86)	<b>-2.16</b> (±1.43)	<b>92.11</b> (±4.75)	<b>1.04</b> (±1.95)	<b>0.81</b> (±4.57)	<b>-5.30</b> (±5.85)
Minimum acceleration	<b>-2.98</b> (±2.34)	<b>-6.17</b> (±3.47)	<b>88.88</b> (±9.58)	<b>-2.41</b> (±0.90)	<b>-1.62</b> (±4.54)	<b>87.36</b> (±4.66)	<b>0.66</b> (±2.66)	<b>4.77</b> (±3.46)	<b>-3.59</b> (±7.02)
Maximum resultant acceleration		<b>95.50</b> (±8.41)			<b>92.15</b> (±4.76)			<b>-5.36</b> (±5.70)	
Minimum resultant acceleration		<b>89.22</b> (±9.36)			<b>87.50</b> (±4.66)			<b>-3.77</b> (±6.69)	

Significance level: \*  $P < 0.05$ .

TABLE 4: Intraobserver and interobserver reliability of variables measured directly during functional reach test.

Variable	SEM (stand. error measu.)	Intraobserver				Interobserver			
		ICC	IC (95%)		ICC	IC (95%)			
			Min.	Max.		Min.	Max.		
Trunk	Time	A_B	<b>0.867</b>	<b>0.855</b>	0.833	0.872	<b>0.851</b>	0.828	0.869
		B_C	<b>4.582</b>	<b>0.835</b>	0.822	0.852	<b>0.831</b>	0.824	0.848
		A_C	<b>3.194</b>	<b>0.847</b>	0.839	0.868	<b>0.840</b>	0.839	0.868
	Displacement	A_B	<b>2.364</b>	<b>0.891</b>	0.879	0.913	<b>0.883</b>	0.877	0.913
		B_C	<b>2.329</b>	<b>0.863</b>	0.843	0.878	<b>0.858</b>	0.845	0.871
		A_C	<b>4.153</b>	<b>0.877</b>	0.861	0.895	<b>0.870</b>	0.859	0.888
Lumbar	Time	A_B	<b>1.463</b>	<b>0.867</b>	0.844	0.880	<b>0.858</b>	0.841	0.879
		B_C	<b>1.624</b>	<b>0.829</b>	0.806	0.855	<b>0.821</b>	0.804	0.852
		A_C	<b>3.011</b>	<b>0.851</b>	0.837	0.869	<b>0.839</b>	0.832	0.860
	Displacement	A_B	<b>1.840</b>	<b>0.878</b>	0.850	0.896	<b>0.875</b>	0.852	0.893
		B_C	<b>1.851</b>	<b>0.868</b>	0.849	0.883	<b>0.863</b>	0.846	0.870
		A_C	<b>1.738</b>	<b>0.872</b>	0.853	0.889	<b>0.868</b>	0.850	0.877
Functional reach test			<b>0.987</b>	0.983	0.992	<b>0.983</b>	0.979	0.989	

was 80.8 (66–90) years [15] and 64 [33], compared with 76.7 years in the present study.

The present study has strength in observing that the parameterization of FRT allows obtaining reliable and valid kinematic measures with a high potential for research in

the clinical field, either in the assessment or the monitoring of different types of patients. However, it also has some weaknesses such as the lack of a control group or restriction on the right side as the affected side of the patient. In addition, the results presented in this study are those obtained in a pilot

trial; however, it is necessary to expand the sample to obtain the results of sensitivity and specificity of inertial sensors in the parameterisation of FRT.

## 5. Conclusions

The main conclusion that can be reached is that the inertial sensors are a tool with excellent reliability, validity, sensitivity, and specificity in the parameterisation of the functional reach test in individuals who have had a stroke.

## Conflict of Interests

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

## Authors' Contribution

Antonio Ignacio Cuesta-Vargas and Manuel González-Sánchez have made contributions to conception of this study. Jose Antonio Merchán-Baeza and Manuel González-Sánchez participated in the acquisition of data. Jose Antonio Merchán-Baeza, Manuel González-Sánchez, and Antonio Ignacio Cuesta-Vargas participated in the analysis and interpretation of data and were involved in drafting the paper or revising it critically for important intellectual content. All the authors have given final approval of the version to be published.

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