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EDITORS' LETTER

This volume 3, number 2 gathers a set of articles based on the most outstanding research on accessibility and disability issues that was presented in the International Conference on NeuroRehabilitation 2012 (ICNR).

The articles' research present in this number is centred on the analysis and/or rehabilitation of body impairment most due to brain injury and neurological disorders.

JACCES thanks the collaboration of the ICNR members and the research authors and reviewers that have collaborated in making possible that issue.

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ASSESSMENT AND TRAINING IN HOME-BASED TELEREHABILITATION OF ARM MOBILITY IMPAIRMENT

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Abstract: The aging population and limited healthcare capacities call for a change in how rehabilitation care is provided. There is a need to provide more autonomous and scalable care that can be more easily transferred out of the clinic and into home environments. One important barrier to this objective is achieving reliable assessment of motor performance using low-cost technology. Toward this end, an assessment framework and methodology is proposed. The framework uses 4 sequential games to measure aspects of range of motion, range of force, control of motion, and control of force. Parameters derived from the range of motion task are used to define motion requirements in all subsequent assessment games, while parameters derived from the range of force task are used to define subsequent lifting force requirements. A 12-week usability study was conducted in which 9 patients completed the clinical testing phase and 6 therapists and 7 patients completed the questionnaire. Feedback from the questionnaire shows the system is easy to use and integrates well in the clinical setting. The most commonly requested modifications were the inclusion of more games and the incorporation of hand training. Some initial position and force data are shown for one subject and discussion on implications for mobility assessment using the developed device are provided.

Keywords: Arm rehabilitation, Mobility assessment, Reach training, Stroke, Rehabilitation, Home-based telerehabilitation.

Introduction

Home-based telerehabilitation is a growing field that has much to offer healthcare. In many developed countries, the population is aging; people are living longer; and the prevalence of stroke continues to increase dramatically with age. These factors add to a large and growing stroke population, while healthcare resources remain rather stagnant. As a result, there is a need to provide more autonomous and scalable care that can be more easily transferred out of the clinic and into the home. One of the largest challenges in this task is providing a means of low-cost quantitative assessment that can provide clinical relevance for therapists. The assessment must allow transparent supervision and the ability to make informed revisions of prescribed training plans during post-acute stroke care.

Stroke is the most common source of long-term disability in Spain and throughout developed countries worldwide. European statistics as a whole report that nearly 1 million people experience a first or recurrent stroke each year (Hesse et al. 2005). Improved medical treatment during acute stroke care has resulted in lower rates of mortality, and yet residual arm impairments persist long-term with only 14-16% of the hemiparetic survivors recovering complete or near complete motor function (Nakayama et al., 1994).

A variety of methods are used in post-stroke rehabilitation including constraint-induced movement therapy and progressive resistance training, as well as techniques aimed at patients with less mobility such as bilateral movement training, and mirror therapy (Oujamaa et al., 2009; Fasoli et al., 2004; Stevens & Stoykov, 2004). For patients with more mobility, constraint-induced therapy is a widely-used approach aimed at combating learned non-use of the impaired limb, but has the limitation that patients must have a minimum level of movement and control in order to use it (Dobkin, 2005) and therefore may only be applicable in as few as 10 percent of patients (Grotta, 2004). Progressive resistance training is another widely-used approach and one which can be supported through web-based interfaces and robotic technologies.

Performing progressive resistance training exercises have been shown to increase both strength and function in a number of studies (Patten et al., 2004). Patten and colleagues summarize nine studies that show evidence that training as little as 3-4 times per week for 6-12 weeks is enough to yield functional improvements. Dobkin (2005) reports significantly better outcomes in task-oriented practice for patients who are able to engage in 16 or more additional hours per week as opposed to those who only spend a few additional hours. Although this supports the "more is better" approach, it has further been suggested that the process and quality of care are likely to be as important as total hours of therapy (Quinn et al., 2009). Patients improve more, for example, when they actively participate in training tasks rather than play a passive role.

Still, the amount of professionally-supported rehabilitation training provided to the average patient falls short of the ideal. A Dutch report published in 2008 (Peerenboom et al., 2008) reported that the average treatment time for stroke patients in skilled nursing facilities was about 4.5 hours per week. Only about half of this time, just over 2 hours per week, was spent in physical therapy.

In an era where rehabilitation services are diminishing under the weight of growing demands and fewer therapists, robotically assisted rehabilitation and home-based rehabilitation have become a major focus of much research. Robotics offer precision and repeatability of movements, quantitative measures, and data that can be used for assessment of movement quality. For these and other reasons, robot mediated therapy for upper limb rehabilitation continues gaining momentum as a very promising technique. Results of clinical trials with robots such as the MIT-Manus (R&D prototype) and the InMotion3.0 (commercialized version of the MIT-Manus) have demonstrated that robotically assisted rehabilitation is safe, accepted by patients, and comparable with conventional therapy (Krebs et al., 1999; Lo et al., 2010). The MIT-Manus, under development since the late 1980's and later commercialized as the InMotion3.0, was one of the earliest systems for robot assisted rehabilitation. Like many research prototypes, its primary focus was on assessment and training in the clinical setting. Only in more recent years have

several groups begun to launch systems for home use to address the real unmet needs in the healthcare system.

Home-based telerehabilitation offers a way of increasing duration and intensity of post-stroke training. Unfortunately, most platforms for home-based rehabilitation are developed with a specific set of rehabilitation tools or devices in mind, and therefore have limited extendibility to other tools and devices. Another criticism that can be made is that most telerehabilitation software is developed from an engineering perspective with minimal requirements derived from the wide spectrum of stakeholders. Because of this, combined with the inherent difficulty of altering existing policies and practices in medicine, progress in clinically-supported telerehabilitation technologies has been slow.

Despite the pace, home-based technologies are making significant advances as the need becomes more recognized. Some of the more notable advances in the history of game-based telerehabilitation have been made by Cogan et al. (1977) with the introduction of Pong to the world of rehabilitation, Reinkinsmeyer et al. (2001) with Java Therapy, Ellsworth and colleagues (Johnson & Winters, 2004) with TheraJoy, Feng and Winters (2005) with UniTherapy, and Lum et al. (2005) with AutoCITE. Following these and other preliminary research studies, a large increase in telerehabilitation efforts have been seen, particularly over the last 3 years. Recent commercial players in the field include Telefonica ([20]), MediTouch ([21]), HomeTelemed ([22]), Tyromotion ([23]), Hocoma ([24]), and others ([25-30]) (see Appendix 1). Even with the players involved, commercial success is limited; new technologies are still needed that can support patients while training at home and simultaneously lessen the load on the therapist. The solution lies in the development of a system that can be easily integrated with current practice and that supports a smooth and early transition of patients to the home environment, with a special focus on minimizing initial and recurrent costs.

In this work, an assessment methodology for a new home-based telerehabilitation system for post-stroke arm rehabilitation is presented. Together with a set of games for mobility assessment and training, results of

a usability study with the ArmAssist system are presented, and some preliminary assessment data and discussion are provided.

Background and Previous Work

Telerehabilitation System Overview

A telerehabilitation system called the ArmAssist (Figure 1) has been under development at TECNALIA for the past five years. It combines a portable device for arm support with web-based therapy management software and a set of games for assessment and training. The training concept is based on well-known research on gravity-induced discoordination in the shoulder and elbow and its effect on the active range of motion of the arm [Beer, Dewald, & Rymer, 2000; Dewald & Bear, 2001; Bear et al., 2004; Ellis et al., 2005].

The combined system is designed to allow the initiation of arm training in the clinical setting, under the direct supervision of a therapist, and the continuation of training at home, thereby increasing both duration and intensity of training. The system components and functionality have been previously described in publications (Zabaleta et al., 2011; Perry et al., 2012; Rodriguez-de-Pablo et al., 2012) and a short summary can be found in Appendix 2.

Telerehabilitation and Assessment Software

In addition to the hardware to support arm reach training, a modular telerehabilitation platform was developed. It was designed to support the phases of therapy planning, execution, and assessment. Details of the functionalities supported in the platform are further described in [Perry et al., 2011a; Arcas Ruiz-Ruano et al., 2012].

In the development of game interfaces, a distinction was made between games for assessment and games for training. Assessment games were short tasks (1-2 minutes) that involved a targeted movement with defined parameters, while training games spanned longer timeframes (5-15 minutes) and provided more entertaining or challenging environments to fill the majority of training time and maintain user engagement. The training games were developed following fundamental training methodologies recommended

by the therapists including movements that should be encouraged and discouraged. Encouraged movements were those that worked against the abnormal muscle synergies in the upper limb (Brunnström, 1970), forcing the user to train reach extension movements that were difficult to control because of abnormal patterns of muscular contraction. The movements generally require simultaneous abduction of the shoulder and extension of the elbow. Arm movements that require the opposite movement, (i.e., flexion of the elbow with adduction of the shoulder), should be supported against gravity and/or allowed to move freely, if not assisted. This is recommended in order to minimize the potential of eliciting undesired spastic reflexes. Assessment games were designed to provide the clinician with an objective assessment of the range of movement, vertical support force, and the patient's ability to perform trajectories and supported reach extension movements. Training games, on the other hand, were composed of more complex tasks and exercises. They included integrated cognitive and motor components of higher complexities in order to better motivate and engage the subject; they incorporated aspects such as: problem resolution with jigsaw puzzles, card games, and sorting tasks; memory recall tasks with classic memory games; and language skills with word completion tasks. Further details about the Telerehabilitation platform and games developed for training can be found in Appendix 3.

Figure 1. The ArmAssist telerehabilitation system is composed of the ArmAssist base module, extended arm reach support, a web-based telerehabilitation (TR) platform, and a 21-inch touchscreen display.
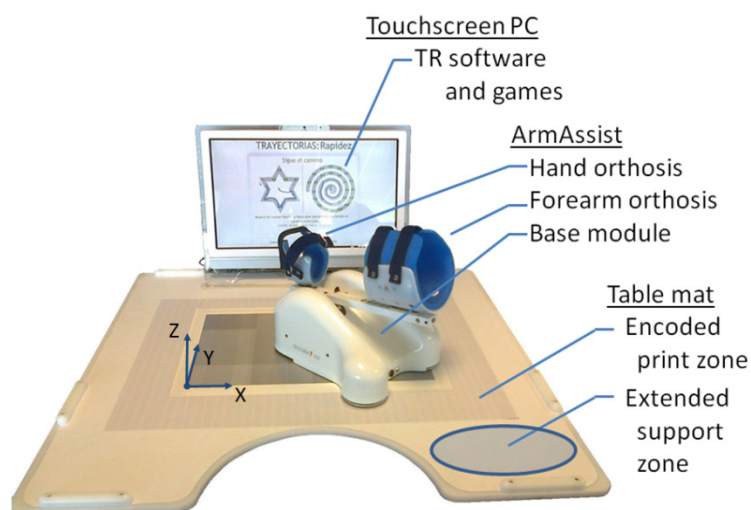
Initial levels of the assessment games are shown in Figure 2. The set of assessment games were designed to measure: 1) Range of motion, involving multi-directional reach extension from a central point (Fig. 2a); 2) Range of Force, involving support arm weight in the vertical direction (Fig. 2b); 3) Control of motion, involving a trajectory following task (Fig. 2c); and 4) Control of force, involving a sustained vertical support force while performing a reach extension task (Fig. 2d). Each game is described in detail in Appendix 4.

Figure 2. ArmAssist games for assessment: (a) range-of-motion, (b) range-of-force, (c) control-of-motion, and (d) control-of-force.

(a)

(b)

(c)

(d)
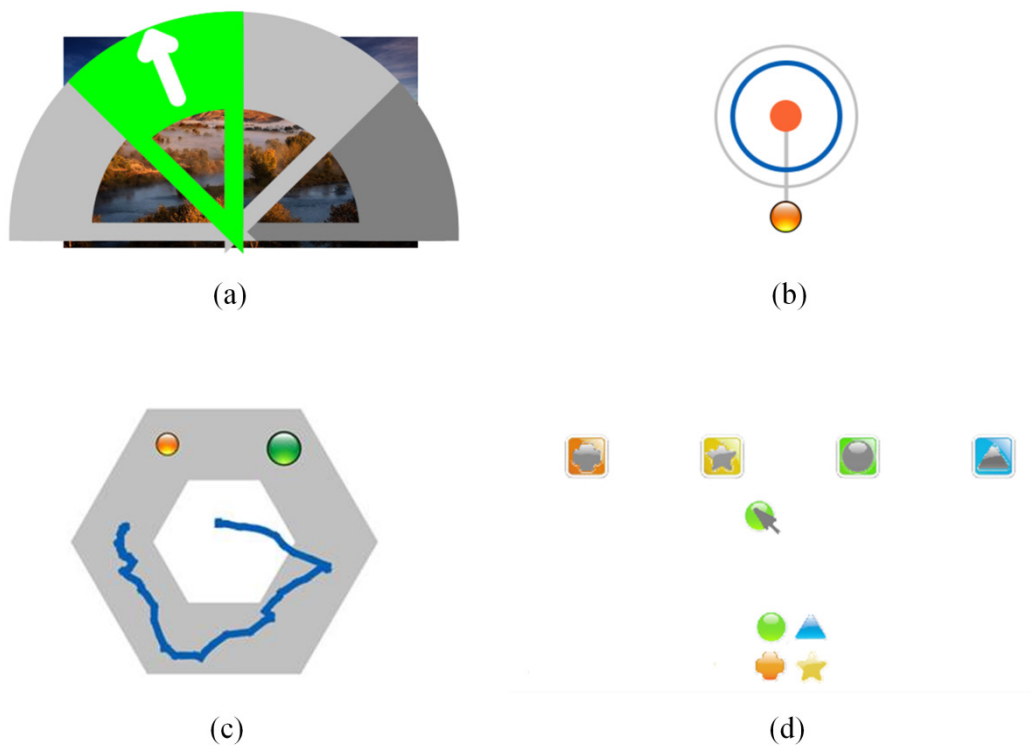
Methods

Usability Testing Protocol

Usability testing of the passive (non-motorized) ArmAssist prototype was carried out at two rehabilitation centers. A 12-week clinical pilot test was conducted according to the timeline shown in Figure 3. The protocol involved a period of 3-4 weeks training in the clinic with direct supervision from the therapists, an 8-9 week training at home, and a transition period in

between during which the system was setup in the patient's home. The transition period allowed for differences between various in-patient stays and coordination with local research personnel to support the setup process. The target amount of ArmAssist training during the study was 30 minutes per day, 5 days per week.

The evaluation of patients' progress using standard clinical scales were planned to take place at fixed stages of the process: on admission, on discharge from in-patient training, on discharge from home training, and 3 months after the home training discharge. In-patient training started when therapists decided that each patient had sufficient trunk and shoulder stability to use the ArmAssist device. The measures for this were not standardized between the centers. The time that each patient spent at the hospital varied depending on his/her condition. Performing the home-training phase at the patient home was not possible in some cases, due to the nature and duration of in-patients at one of the centers. As a result, the degree of supervision during the "home training" phase varied with each center and therapist, and was not strictly enforced.

Feedback was collected from seven patients and six therapists through written questionnaires while other patients contributed only through recorded movement data of 2D position and vertical force. Questionnaire feedback was collected voluntarily from patients and therapists via a series of structured interviews and Likert-based evaluation questions. The administered questionnaires presented 16 questions to each patient and 19 questions to each therapist. Questions included aspects of system features, system usability, and recommendations from the user to improve the system. The specific questions can be found in Appendix 5.1. 2D planar position and vertical support/lifting force data during assessment tasks were recorded throughout the clinical and home training phases for later analysis. Of the participating patients, nine completed the clinical testing phase. Range of motion and range of force data from two of these patients is provided in the results section.

Mobility Assessment Games and Measures

General performance indicators were stored for all games in each session. During assessment games, full force and trajectory information were also stored in order to allow a more detailed post-processing analysis. In case of lost network coverage, the platform was equipped with an offline training and data storage mode so that data could be stored locally on the hard-drive and synchronized periodically with a central server.

Strict overall times and intermediate countdowns in the case of inactivity were employed in all the assessment games to ensure that assessments were carried out efficiently.

Figure 3. Testing timeline for usability evaluation of the ArmAssist telerehabilitation system.
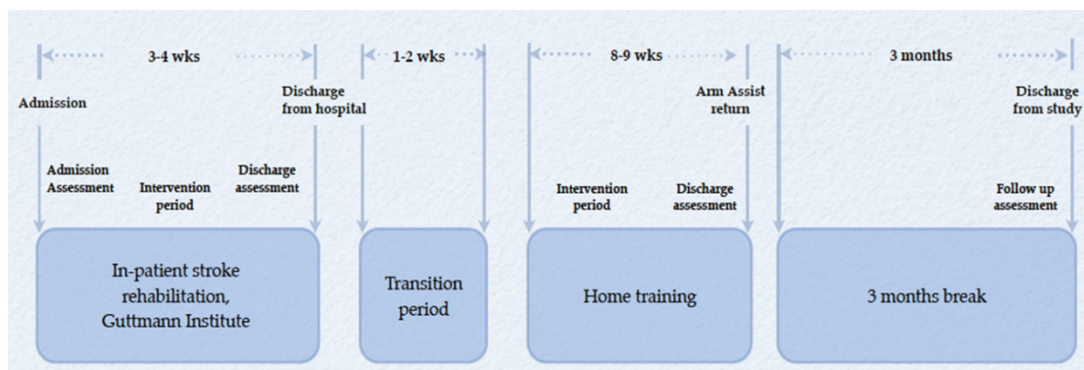
Level Structure

The game level structure implemented had three level aspects: motion level (L_{ROM}), force level (L_{ROF}), and task level (L_{TG}). The motion level, L_{ROM}, was set by the range-of-motion assessment game and altered the range of motion of all successive games (assessment and training) for the session. For this reason, the range-of-motion assessment was always performed first. The force level, L_{ROF}, set by the range-of-force assessment game was also used to alter all successive games. The range-of-motion game did not involve a force level assignment as the objective was to measure range of movement in the fully-supported condition.

There were 5 difficulty levels for each level component (i.e., L_{ROM}, L_{ROF}, L_{TG}) to allow adjustment of the assessment environment to best fit the patient´s

capabilities. Each game was scored based on a combination of evaluated features. In this experiment, the game levels were adapted automatically based on performance. The motion and force levels were adjusted by the range-of-motion and range-of-force games, respectively, and the task level was adapted based on the previous performance(s) in each respective game. The adaptation method adopted was the following: a game score of 100 percent or two consecutive scores of at least 80 percent prompted a level increase. The game levels involved in defining the difficulty of each task are illustrated in Figure 4a. Examples of the initial and final game levels for the assessment games are shown in Figure 4b

Figure 4. (a) Sequential relationships of assessment measures and level structure. (b) Assessment game level increases from initial level to final difficulty level for each of the assessment games. Note that the force level in the range of force game (bottom-right) can be adapted without changing the visual layout of the task.
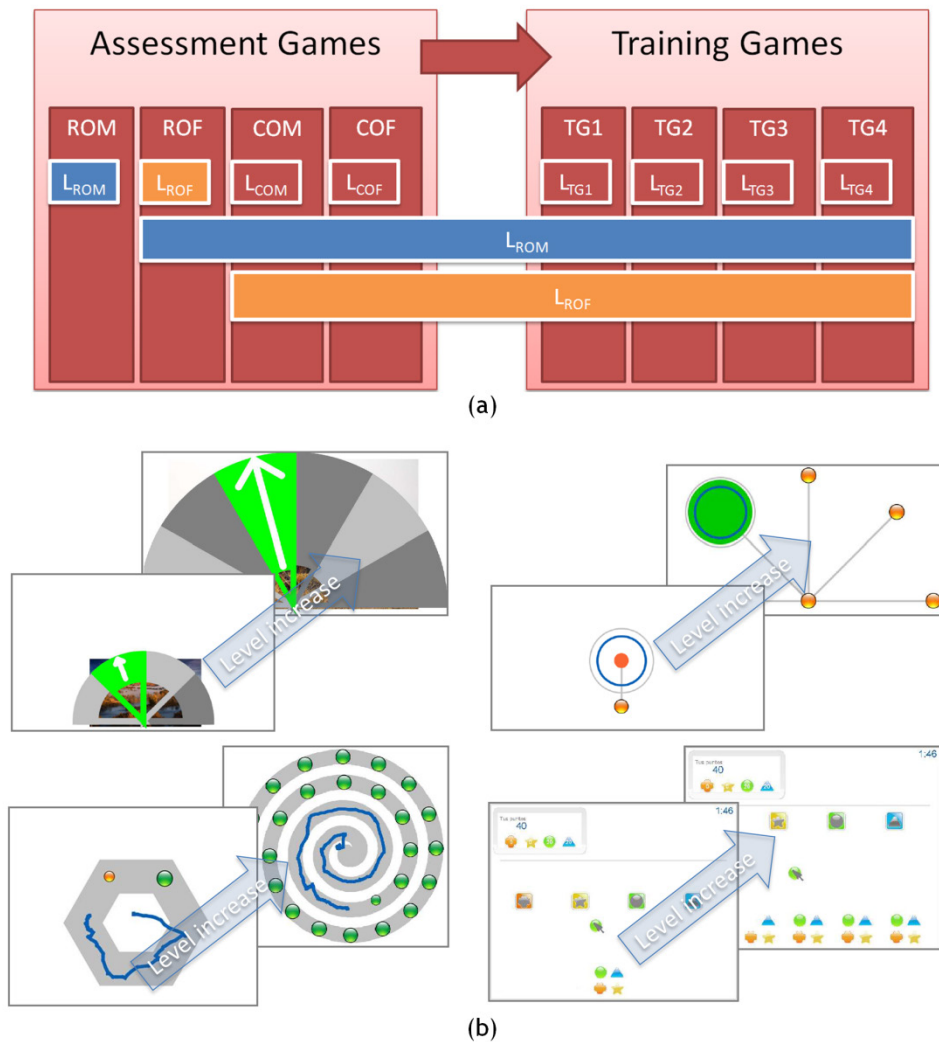
Results

Usability Feedback

Six therapists and seven patients provided feedback through the evaluation questionnaire. The feedback gathered from patients and clinicians was overall very positive. The system was found easy to use, and was generally well accepted. The games provided a clear increase in motivation when patients started using the system and the therapists felt the tasks were well aligned with the techniques they typically used for training. Through the questionnaire, therapists expressed that they felt the system would be useful for the kinds of patients that they see, that the patients would benefit from the training, and that the training would produce an improvement in the patient condition. One of the criticisms common to both therapists and patients concerned the need for a wider selection of training games and for an increased number of levels within the same game. This and other feedback related to the usage and user perspectives of the telerehabilitation platform and games are being integrated into the system to improve its features and usability. Further details about the questions and responses for both patient and therapist questionnaires can be found in Appendix 5.

Pilot Assessment Data

A qualitative evaluation of the data progressions reveals characteristics of the reach movements related to range of motion, directional control, smoothness, and limb support capacity. It should be acknowledged that to show these characteristic trends quantitatively along with their respective magnitudes and significance, further analyses and computation of metrics are needed. For the purpose of illustration and discussion, initial results of range of motion and range of force assessments from two stroke subjects are presented in Figures 5 and 6. The subjects were training at two different rehabilitation centers in Spain. Although 9 subjects completed the clinical training phase, only two subjects from one center continued the protocol at home due to various reasons unrelated with system performance. Due to the

nature of one center, it was not possible to support a truly home-based training, and so when possible, patients continued training in the clinic under reduced supervision.

Figure 5. Pilot result of planar movement and vertical support force for stroke Subject A. Polar plots of range of motion ((a)-(d)) and range of force ((e)-(h)) assessment show movement data in the horizontal plane, and boxplots of arm support ((i)-(l)) show force applied to the device at the target locations illustrated in polar plots (e)-(h). Color coding in subplots (e)-(l) show correspondence between locations and magnitudes of vertical force data. Data shown in each column were recorded during the same session (session dates provided at the top of each column).
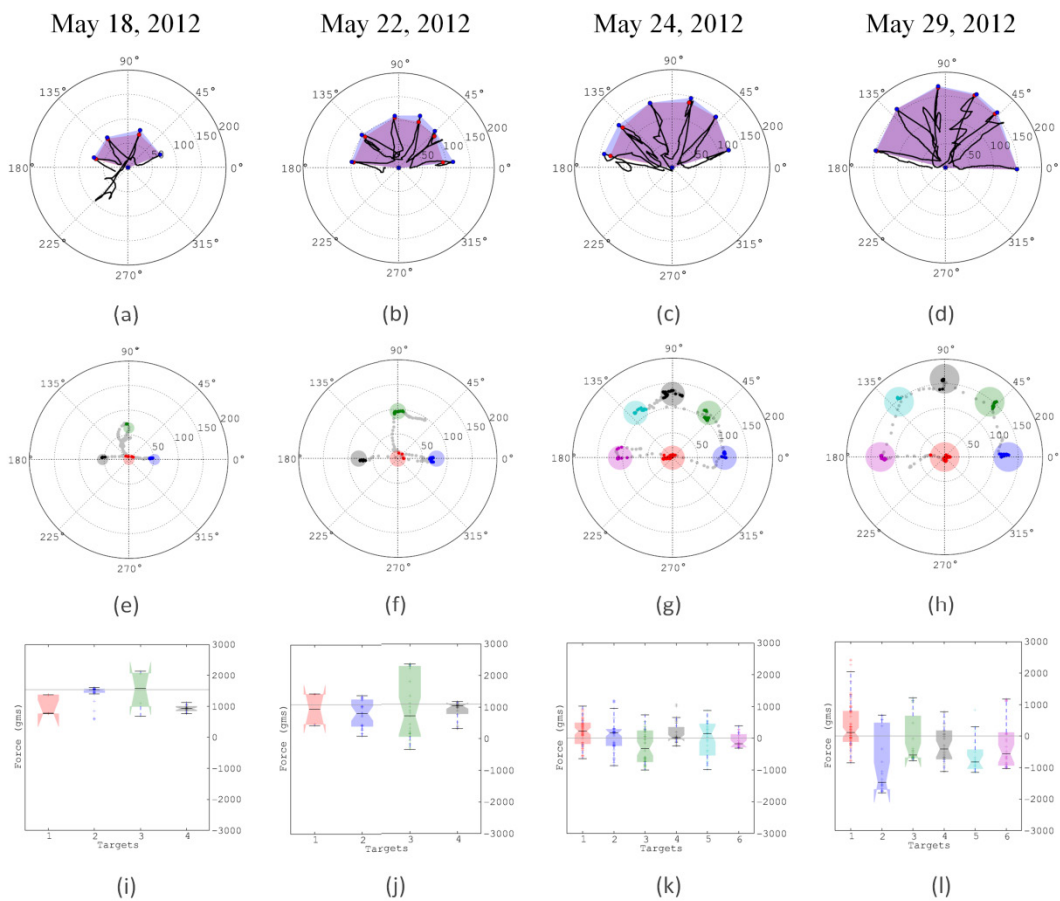
In both Figures 5 and 6, subplots (a)-(d) show four polar plots of planar movement trajectories and work areas during the range of motion assessment. Subplots (e)-(h) show polar plots of planar movement trajectories and locations of selected force measurements near the targets. Colored data points in subplots (e)-(h) show the locations of the force data that have been included in the boxplots of subplots (i)-(l). Subplots (i)-(l) show boxplots of the vertical force measures that are achieved near the targets where the velocity is low (i.e., less than 10 percent of the peak velocity). Grey data points in subplots (e)-(h) indicate trajectory points where the velocity was higher than 10 percent of the peak velocity or where the distance to the target was more than 25 percent of the average distance between central and peripheral targets. This area is shown in subplots e-h as colored areas surrounding each target location. These were not visible during the assessment game. Data shown in each column were recorded during the same session and the session date is provided at the top of each column.

In the range of motion assessment data for Subject A, shown in Figure 5, both target work area (red shaded region) and performed work area (blue shaded region) increase over the sessions. The planar location of vertical force assessments in Figures 5(e)-(h) (colored circles) progressively shift more distal while the target vertical force threshold (grey line, Figures 5(i)-(l)) which shows the target level of unloading (i.e., reduced resting weight on the device) lowers on the graph, and the patient's ability to unload the arm is maintained or increased at progressively more distal targets.

Similar trends can be seen in the data of Subject B, shown in Figure 6. Although the work areas in Figures 6(c) and 6(d) were nearly the same, improvements in directional control and force measures are seen. More direct trajectory paths and smaller groupings of endpoint positions during the lift tasks indicate the patient exhibits a higher level of control. Comparing vertical unloading forces in Figures 6(i) and 6(l), although the magnitudes and variations are similar, the progression shows that an improvement in sustained support in extended reach positions was achieved.

In Figure 6(g), two targets on the left were not reached within the allotted time and the assessment algorithm moved the targets 50 percent closer to the central target where they were then successfully reached.

During the usability study, the actual time spent in assessment and training by the patients fell short of the desired 30 minutes per day, averaging 14 minutes and ranging from 6.7 to 40.8 minutes per day.

Figure 6. Pilot results of planar movement and vertical support force for stroke Subject B. See Figure 5 caption for subplot details.
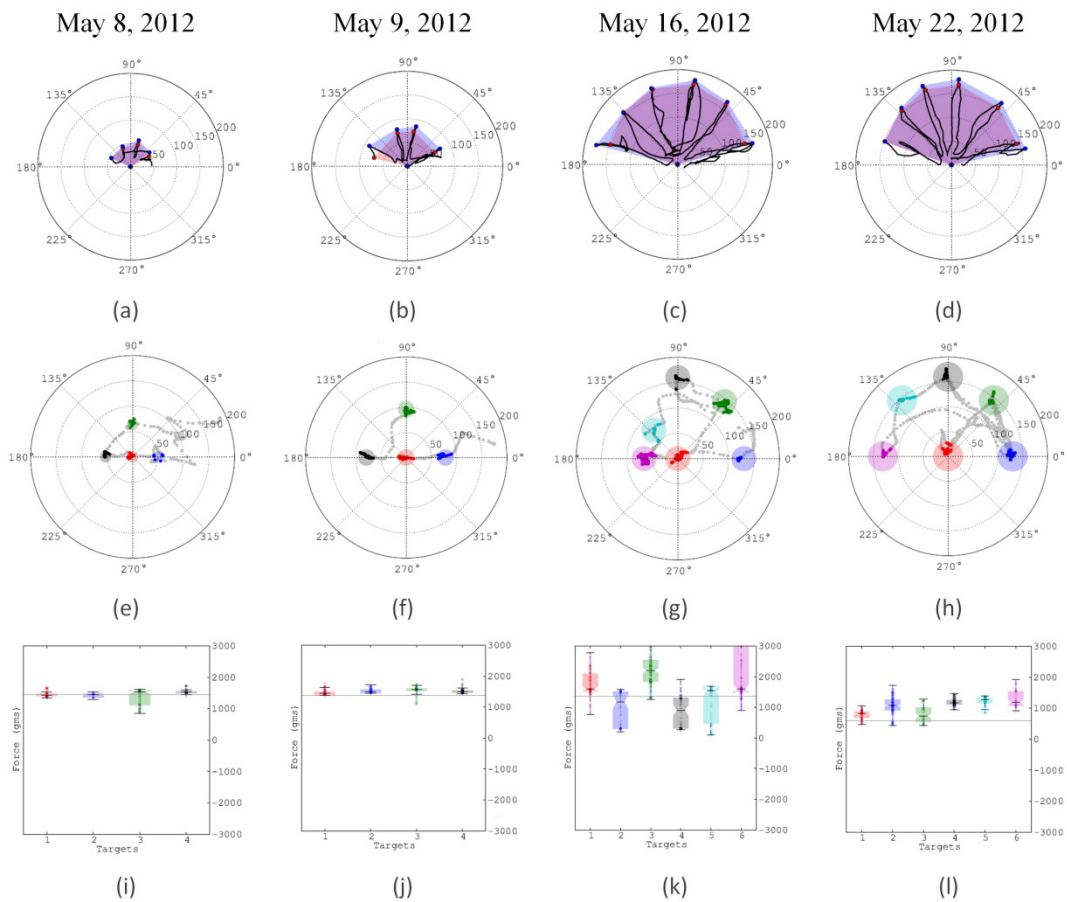
Discussion

Training Motivation

Although patients were clearly motivated at the start by the game interaction and feedback, the set of games and levels available played a significant factor in decreasing motivation as the patients trained for longer durations. When asked at the beginning of training about patient motivation using the system, therapists were nearly unanimous in their belief that the system increased motivation. When asked at the end of the study whether the patients were motivated to train longer with the system, the responses were a bit more neutral but maintained a clear tendency toward agreement that the system increased motivation (see Appendix 5.2). At the same time, therapists and patients consistently made requests for a wider variety of games, both to increase the number and expand into new genres. As a result, although the strength of motivation reported was more neutral after the 2-3 months of training than at the start, the results indicate that motivation is increased by the system and that the system´s maximum potential for motivation was not reached.

Assessment Metrics

In this paper, various assessment games and training adaptation methods have been proposed and qualitative observations on assessment data have been made, but little focus has been placed on metrics. These qualitative observations can theoretically be confirmed with the computation of quantitative metrics, providing a more objective evaluation of patient mobility performance. Although the selection and comparison of optimal mobility metrics has not been presented here, this work is in progress and will be the focus of a future publication. It should be noted that the optimal methods and metrics to use for mobility assessment is a current and ongoing debate (Lambercy et al., 2012). For range of motion, the metrics of interest were those that represented the extent of extension movements away from the torso, and therefore can be represented by an array of linear measures, or as a single measure of area. For control of motion, the metric of interest

was the smoothness with which positional changes are achieved to reach a known target that requires extension of the arm. For range of force, the metric of interest is the maximal level of self-support against gravity that the user can achieve. This does not imply a movement of the arm or an active application of force by the device, but rather the passive measurement of the weight of the user´s arm resting on the device. This was done at known and predetermined locations within the user´s active range of motion. For control of force, the metric of interest was smoothness of the force signal as well as the error with respect to the desired force.

In the measure of range, an important element to monitor is posture in order to ensure a proper measure of movement. If compensatory movements can be avoided, the important aspect in range of movements is not *when*, but *whether* a target can be reached. The element of time will be accounted for in the later measures of control, as a lack of control will naturally lead to sub-optimal movement trajectories that require a larger execution time. Postural changes during compensatory movement are a common occurrence during arm rehabilitation. Although a greater level of compensation may be allowed during training, for a proper assessment, postural compensation must be handled either through the use of physical or mechanical restraints or visual monitoring and corrective feedback.

Game Levels

Five game levels were defined by the number and radius of sectors in the 2D planar workspace with the goal of: a) allowing patient-specific adaptation in order to maximize visual resolution of movement feedback to the user, b) improving the match between ability and task challenge within the assessment, and c) reducing the potential for demotivation as a result of the size or complexity of the displayed, and/or potentially unused, workspace. Through combination, the 5 levels produce up to 25 game levels in the range of force game and up to 125 levels in games specified by L_{ROM}, L_{ROF}, and L_{TG} levels. The number of levels that will be played by any one patient, however, will vary with their ability level and the speed at which they progress through each individual assessment level. While this could offer benefits for patients whose compensatory movements are prevented or

corrected during assessment, in many clinics this was not the case. Assessments were done with minimal correction on posture allowing patient to advance through the assessment levels at a potentially faster rate than otherwise would be expected. This consequently makes it more difficult to compare patient results between sessions over the duration of training until level 5 is reached and the assessment task remains constant. As a result, it is recommendable to present range tasks in a standardized single-level format. The results of the range-of-motion can (and should) then be used to normalize the range-of-force assessment, and the outcome of both range-of-motion and range-of-force should be used to normalize the control-of-motion and control-of-force assessments, and all successive training games to the patient. This automatic game normalization should be carried out with the purpose of optimizing the presentation of tasks in the appropriate position and force workspaces in order to both challenge and motivate the patient to advance.

Training Time

It was noted during the study that the actual time spent in training varied substantially between patients and sessions. This is thought to be due in large part to the method in which training assignments were made and the nature of the game lengths to be highly dependent on the ability and speed of the user. Game durations were not fixed, nor were estimated lengths provided to the therapists to know approximately how much training time they had assigned to the patient. These results lead to a need for a better control on the length of assigned training tasks, for example, by repeating assigned games until a predefined time period has expired.

Future work

Building from the initial pilot testing results showing qualitative improvement with training, further work on comparison of alternative assessment metrics will help to provide clinicians with objective measures on which to base training revisions. Optimization of these metrics, and comparison to standardized clinical scales in order to best characterize mobility deficits within the stroke population, will provide valuable

measures of patient progress that can be performed independently and monitored remotely. For the patient, improved quantitative characterization of deficits is expected to improve automatic level adaptation within the training games in order to optimally match the user´s ability with the task challenge. Proper adaptation of the task challenge such that the risk of boredom (being under-challenged) and frustration (being over-challenged) are minimized, is expected to increase aspects of user engagement and motivation.

Conclusion

The telerehabilitation system and training adaptation structure described in this paper has been developed and evaluated with therapists and patients in both in-clinic and at-home settings in order to maximize usability with the end users. A set of games for mobility assessment and training were developed following therapist recommendations that games should train coordinated movements that go against the abnormal muscle synergies, and avoid movements that can reinforce flexor muscles. Through appropriate game design and selection of task parameters, tasks and task difficulty can be adapted for each training session in telerehabilitation training at home. In their current versions, the range-of-motion, range-of-force, control-of-motion, and control-of-force assessments are performed by uncovering a picture, lifting (or unloading) the arm at one or several positions, tracing a path trajectory, and by picking and placing objects from proximal to distal locations.

It is believed that the quality of arm mobility in planar reach movements can be adequately characterized by measures of planar position and vertical force. Qualitatively, the mobility data measurement recorded by the non-motorized ArmAssist, consisting of 2D position and 1D vertical force, are able to represent improvements in mobility performance over time. The visual progressions with increased training duration show noticeable improvements in both directional control and general smoothness during arm reach and lift tasks. However, demonstration of long-term efficacy of the intervention in

home use requires further study of mobility metrics and stricter adherence to the home testing protocol.

Overall, initial feedback from patients and clinicians has been highly positive. Some decline in motivation and participation was observed with some patients due to the low number of training games and levels developed for the purpose of system evaluation. It is recognized that a high number of training games is needed in order to maintain patient engagement and motivation in long-term high intensity training. These findings promote the need for a larger array of task-specific games and training exercises to both improve the variety of training activities available and to increase the level of sustained user engagement and active participation.

Acknowledgements

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APPENDIX

APPENDIX 1. Growth of telerehabilitation

Over the past 50 years, the number of software systems for home-based rehabilitation has remained relatively small with a sharp increase in related research in the last decade. The promising benefits of rehabilitation systems for home use are attracting universities, research institutions, and companies alike. Companies showing interest in the field are primarily new startups, but examples of well-established companies such as Telefonica, can also be found. A short list of research and commercial activities in Telerehabilitation are illustrated in Table 1.

In a first attempt to produce more enjoyable and motivational games for rehabilitation, Cogan et al. (1977) modified the commercial game Pong into a task interface played with a joystick for rehabilitation of hemiparetic patients. In 2001, Reinkensmeyer et al. (2001) introduced a novel web-based

force feedback telerehabilitation application called "Java Therapy". Building on previous ideas, Ellsworth and colleagues created TheraJoy (Johnson & Winters, 2004), a telerehabilitation environment that uses a modified force-feedback joystick to complete games and tracking tasks created with the custom software UniTherapy (Feng & Winters, 2005), a computer-assisted neurorehabilitation tool for teleassessment and telerehabilitation of arm function. In 2005, Lum et al. performed some novel experiments with the AutoCITE device ([23]) for providing guidance in the performance of various Constraint Induced Movement tasks. The system was used to simulate the effect of telerehabilitation by separating the patient and therapist in different rooms. Results indicated the AutoCITE could be used to provide Constraint Induced Movement therapy with a 75% reduction in Therapist time.

Table 1. Summary list of devices and platforms for telerehabilitation, showing the approximate year of development, whether upper limbs (UL) or lower limbs (LL) are targeted, and the company or research group involved.

Rehab Device Name	Year	Targeted Joints	Company (Location) / Researcher (University)	Compatible TR Platform
*T-WREX**	*2001*	*UL: shoulder, elbow*	*Renkinsmeyer (Univ. California, Irvine)*	*Java Therapy*[14]*
*TheraDrive**	*2004*	*UL: wrist*	*Winters (Marquette U.)*	*TheraJoy* [15]*
AutoCITE	*2005*	*UL: wrist*	*Lum (Catholic University of America)*	*(Simulated platform)* [17, 18]*
(IMU-based device)	*2009*	*LL: hip, knee*	*Telefonica (Spain)*	*RehabiTIC [19]*
HandTutor, ArmTutor, LegTutor	*2010*	*UL: wrist, fingers, UL: shoulder,elbow, LL: hip, knee*	*MediTouch (Natanja, Isreal)*	*MediTutor [20]*
Curictus VRS	*2010*	*UL: shoulder, elbow,wrist*	*Curictus (Gothenburg, Sweden)*	*Curictus Analytics* [24]*

Rehab Device Name	Year	Targeted Joints	Company (Location) / Researcher (University)	Compatible TR Platform
ReJoyce	*2010*	*UL: shoulder, elbow, wrist, fingers*	*HomeTelemed (Edmonton, Alberta)*	*(video-conference software) [21]*
Pablo, Pablo®Plus	*2011*	*UL: shoulder, elbow, wrist*	*Tyromotion (Graz, Austria)*	*(Assessment and Therapy) [22]*
ArmeoBoom	*2011*	*UL: shoulder, elbow*	*Hocoma (Switzerland)*	*Armeocontrol [23]*
*ArmAssist** (Others)*	*2011*	*UL: shoulder, elbow UL*	*Tecnalia (San Sebastian, Spain)*	*TeleREHA** [25]*
(Various devices)	*2012*	*UL / LL*	*Play4Health.com (Palma de Mallorca, Spain)*	*Platform P4H [26]*
(Kinect-based device)	*2012*	*UL / LL*	*Principe Felipe (Valencia, Spain)*	*Neuro@Home* [27]*
Gloreha	*2012*	*UL: fingers*	*Indrogenet (Brescia, Italy)*	*? [28]*
(Kinect-based device)	*2012*	*UL / LL*	*VirtualWare (Basauri, Spain)*	*VirtualRehab [29]*

In the case of Telefonica, a well-known provider of telecommunication services, a platform called RehabiTIC has been developed to guide patient movements based on movements acquired from wearable sensors (Olivares, 2011). Six systems are under development through small startups, and two through more established companies (Hocoma in Switzerland and Tyromotion in Austria). Two systems are based on movements acquired from a Kinect sensor while games are used for guidance and correction ([27], [29]), while the majority use proprietary devices in the training of movement. One of the six startups, Curictus (Gothenburg, Sweden), was acquired by the JSM Group in 2010 and, as a later result of difficulty entering the market, decided to close the project. Subsequently, the code was made available to the public

for others to use and further develop ([24]). The list in Table 1 is not a comprehensive overview of the current telerehabilitation work, but provides insight on the recent rate of growth.

## APPENDIX 2. ArmAssist Hardware for arm reach support

The ArmAssist system is equipped with 5 integrated sensors and 4 degrees of freedom (dof). Together, the sensors and freedoms allow measures of supported movement in a large semi-planar workspace. The hardware is designed for left and right use, employing lateral symmetry in components such as the base module, table mat, and monitor placement. Non-symmetrical components like the orthoses can be rapidly disconnected and replaced with a single button press (Perry et al., 2012).

Wireless Mobile Base Module

The internal structure of the wireless mobile base unit is composed of an aluminum structural frame, an integrated pcb, a 1-dof force sensor, and a quick-connect forearm assembly. The structural frame is an assembly of custom aluminum brackets designed to house three omni-directional wheels, the force sensor, and connect the various other components that make up the assembly. The 1-dof force sensor has been integrated as the connecting element between the structural frame and the forearm assembly, measuring all user interaction forces in the vertical direction. This vertical support measure is a fundamental component of the system and the primary measure that enables progressive load training.

Global Position Detection Table Mat

Figure 7. The global position detection mat is printed with 16 zones (a), each containing a distinct repeated pattern (b).

(a)   (b)
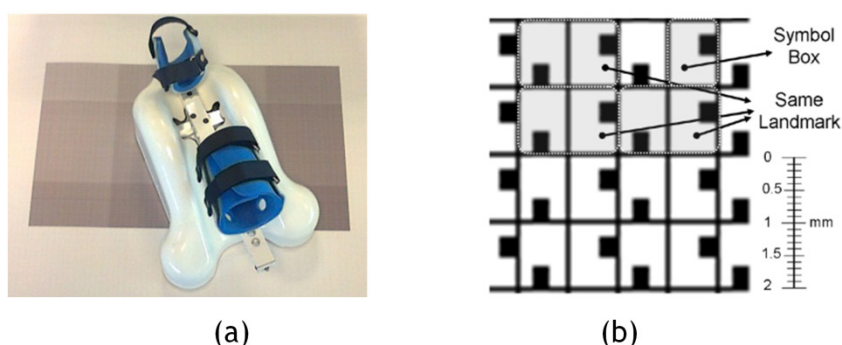
The global position detection mat is composed of a high density polyethylene sheet with a high resolution laminated print mounted to the top. The sheet extends the planar support surface provided by a standard table in the lateral zones. A semi-circular cutout allows patient to sit close to the table while allowing the mat to wrap around the torso, providing support to the mobile base module in the lateral regions when the patient is at rest. Further details of the global position detection mat and the optically-based encoding method utilized has been previously published (Zabaleta et al., 2011).

In summary, the laminated print is encoded with a central grid composed of 16 zones (Fig. 7a) that are each made up of a repeated 2-symbol pattern (Fig. 7b). The pattern is then captured by a camera and sent to the PC for image processing and the position within the current zone is estimated. The accuracy of the estimation can be affected by image print quality and camera resolution. For laser print quality and an ADNS-3080 mouse sensor (Avago Technologies, 30x30 pixel count), the system planar (x-y) resolution that can be achieved is on the order of +/-1 cm in position and +/-2.6 degrees in orientation (Perry et al., 2012).

APPENDIX 3. TeleREHA platform and training game development

TeleREHA Online vs. Offline Platforms

It was evident during early system testing that gaining access to a reliable internet signal in many hospital settings is currently unrealistic. As a result, both online and offline versions of the software were developed. For the online version, Java was used as the primary language for the system server development. The Spring framework was used in a 3-layer model, including Hibernate with MySQL as the database layer, and Primefaces for the presentation layer. For the video communications, a red5 media server was used. In the case of the offline platform, a 2-layer model using Java Server Pages and XML as storage repository was used, avoiding the installation of a locally-running database. Ajax was used for results synchronization with the centralized server. In both online and offline platforms, Tomcat 6.0 was used for the application server.

In the development of serious games for assessment and training, the same technology was used in both TeleREHA online and offline versions. 2D games, described in (Rodriguez-de-Pablo et al., 2012), were developed with Java 2D, whereas 3D games were built with Java Monkey Engine 2. Games were deployed on the web browser using Java Web Start, and results were stored in XML files for transfer to the centralized server.

Serious Game Development

Throughout the game design and implementation process, ergonomic and user interface design standards were closely observed. Design criteria considered included aspects that were deemed necessary for proper administration of game-based training such as consistency between games, suitability toward visual or cognitive impairments, clarity of instructions and feedback, and robustness. A detailed description of the aspects of serious games that were considered fundamental are described in (Rodriguez-de-Pablo et al., 2012).

Four games for assessment and five games for training were developed. Assessment games measured multi-directional range of motion from a central point, vertical force support capacity, trajectory-following ability, and controlled lift and reach ability, involving a combined control of planar movements with a simultaneous vertical support force. Difficulty levels within the assessment games were configured with varying rules. Parameters related to range were increased linearly, both for force and motion targets within each sector, whereas increasing difficulty parameters related to control were less defined. The control of motion game, for example, increased in workspace according to the range of motion assessment, but simultaneously increased nonlinearly in complexity by adding additional targets to fill the workspace.

A first implementation of the lift and reach control-of-force assessment game, called Drag and Drop, was developed and later determined to involve too much time and cognitive involvement for a rapid assessment. The drag and drop game has since been removed as an assessment game and instead added to the set of training games, and a new lift and reach game has been

developed to better assess the user´s level of force control in a fast and simple manner.

Games for training included Memory, a Puzzle, Solitaire, Word Completion, and the Drag and Drop game. The complexity of each game was designed to be adaptable over a range of difficulty levels to better match the ability of the user.

APPENDIX 4. Assessment game descriptions

The Discover the Picture assessment game (Fig. 2a) evaluated the range of movement in different directions of the transverse plane. In the game, a picture was uncovered by erasing the sectors with a reach extension movement of the arm. The direction of the movement and the sector in which the movement should be made was indicated by a white arrow on a green background. The user had to make a controlled movement without excessive velocity in order for the range to be counted. Lateral deviations from the sector prompted the user to return to the sector at the last value of range achieved before an additional range of movement could be obtained. Five game levels were defined by the number and radius of sectors.

The Range of Vertical Force game (Fig. 2b) assessed the arm support/lifting capacity in different positions of the plane by placing the cursor over a circular target and lifting the arm. As the arm was unloaded from the device, the size of the target was increased in proportion to the lifting force in order to reach the diameter of a peripheral ring which indicated the target unloading level. The different levels were configured by the number of target positions and the percentage of arm weight to be lifted to get the maximum score at each location.

The Trajectory game (Fig. 2c) monitored the ability to perform a controlled movement along a trajectory signaled by a discrete path of circular targets. The various levels were defined by the number of circular targets, the trajectory difficulty (hexagon, star, or spiral) and the path width. In the first level, the user must trace the path of a hexagon, whereas in more advanced

levels the user traced a spiral (clockwise for right-arm patients, counter-clockwise for left).

The Force Control game (Fig. 2d) involved a sustained support force while performing an extension reach movement. A set of objects are located proximally on the screen and a set of targets with similar color and shape to the object to which they corresponded were located distally on the screen. The user was instructed to partially unload the weight of the arm from the device and maintain it in order to lift the objects from proximal locations and carry them to the distal targets. Once the target was reached while still carrying the object, the object was removed from the display.

APPENDIX 5. Questionnaires and questionnaire results

APPENDIX 5.1 Questionnaires

Six therapists and seven patients answered the evaluation questionnaire administered (16 questions for the patients, 19 questions for the therapists). The questions were over the same topics for both groups except for 3 additional questions (12, 18, and 19) on the therapist questionnaire related to the appropriateness of the system for patients and the feedback of patient results; these questions were omitted from the patient questionnaires. For each statement in the questionnaire, subjects were asked to show their level of agreement or disagreement according to a standard 7-point Likert scale measuring the level of agreement (1-strongly disagree, 7-strongly agree).

Statements included in the questionnaires that required Likert-based valuations were the following (*note that* questions 12, 18, and 19 were submitted only to the therapists*, and that* question 13 was submitted only to the patients):

Q1. It has been easy to learn how to use the system, both the hardware and the software.

Q2. I think I will often need the support of a technical person to be able to use this system.

Q3.   Using this system, I need to spend a lot of time in non-training activities (system setup, login, game selection/loading, etc.).

Q4.   I can remember with no problem how to use the system effectively every time I work with it.

Q5.   It took a long time to be able to use the system without problems.

Q6.   I think that I will benefit from using this system / I think that patients will benefit from using the system.

Q7.   Using this system I am motivated to train longer / I think the system will motivate the patient to train longer.

Q8.   I think that this system is uncomfortable to use.

Q9.   I enjoyed training with this system / I think the patients enjoy training with the system.

Q10.   I would recommend other people to use this system.

Q11.   I think that the system must be improved.

Q12.   I think that a tele-rehabilitation system would not be beneficial for the kind of patients I treat.

Q13.   I had internet connection problems while using the system.

Q14.   I feel uncomfortable using a system like this, because I have no experience in using a pc.

Q15.   I don't think using this system will make any change to my condition / I don't think using this system will make any change to the patient's condition.

Q16.   I feel that the games are inadequate for the training.

Q17.   I am familiar with this kind of technology.

Q18.   The outcome results of the training are sufficient and clearly presented.

Q19.   I don't need to use the outcome feedback of the system to see if the patient has improved.

In addition to the likert-based scoring on the statements above, patients and therapists were asked to comment on the aspects they liked most and least about the system, and also to leave any additional comments they had to improve the experience with the system.

APPENDIX 5.2 Questionnaire Results
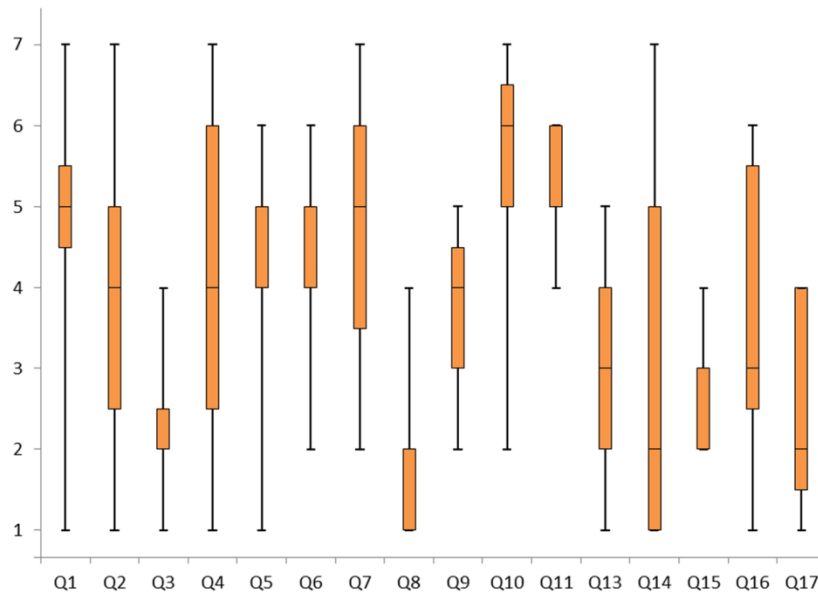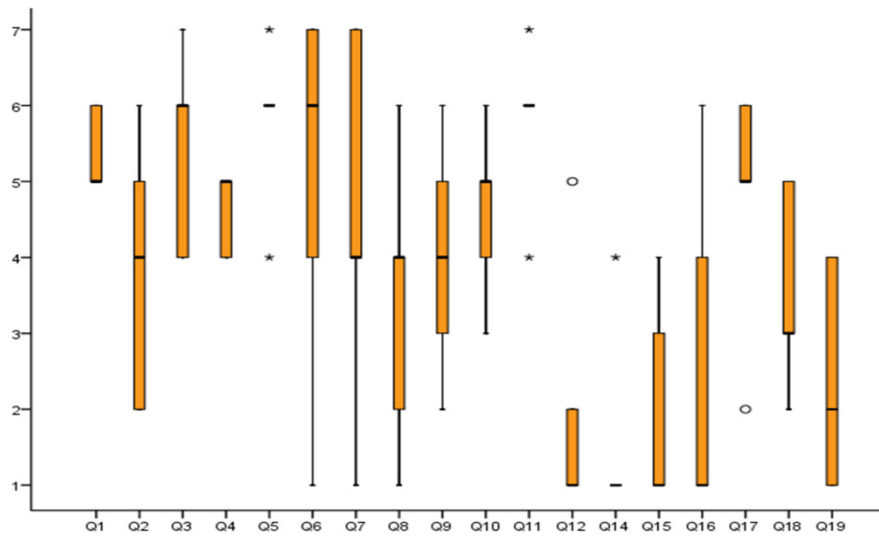
Figure 8. Patient Questionnaire Results

Figure 9. Therapist Questionnaire Results

THE EFFECTS OF NINTENDO WII® ON THE POSTURAL CONTROL OF PATIENTS AFFECTED BY ACQUIRED BRAIN INJURY: A PILOT STUDY

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Abstract: Scientific literature demonstrates that *postural control* after suffering a brain injury can actually relate to its functional prognosis.

Postural control is a result of complex interactions of different body systems that co-operate in order to control the position of the body in the space and is determined by the functional task as well as by the environment in which it is developed. The use in rehabilitation of Nintendo's Wii® gives some results on motor functions.

Objective: Analyse the effects of the Nintendo Wii® console on *postural control* during the execution of an everyday life task consisting of getting up and walking three meters.

Range: Quasi-experimental study of a test-retest type. Not random sample of patients (n=12) affected by brain injury evaluated for the afore mentioned task with the Timed Get Up and Go Test (TGUG). Intervention and results on experimental group (n=6). Comparison of variables with respect to control group (n=6).

Results: Significant results have been obtained (p=0,007<0,05) at the TGUG of the experimental group.

*Keywords:* *postural control*, Nintendo Wii, ABI.

Introduction

Postural control arises from the interaction between a subject, a task and the environment in which it is carried out, with the purpose of maintaining the subject stable and spacially oriented. It must ensure the control of the position of the body against gravity in all the actions that are accomplished (Shumway-Cook & Woollacott 2000; 2006).

Body orientation is defined as the ability to maintain an appropriate relationship between body segments, and between the body and the environment, while performing a particular task, maintaining a vertical orientation of the body.

In the process of maintaining stability (the ability to maintain the center of body mass (COM) between the limits of the support surface), we use multiple sensory references, including gravity (vestibular system), the support surface (somatosensory system) and the relationship of our body with objects in the environment (visual system) (Shumway-Cook & Woollacott, 2000; 2006).

Postural control requires the integration of the musculoskeletal and neural systems. The muscleskeletal components include aspects such as range of movement, muscular properties and biochemical relationships among the corporal segments involved. The neuronal components, that are essential for *posture control*, cover motor processes that include synergic neuromuscular responses, sensory processes that include vision, vestibular and somatosensory systems, sensory strategies that organize these multiple inputs, important internal representations to go from the sensation to the action, and processes of superior levels that are essential for adaptative and anticipatory aspects of *posture control* (cognitive aspects) (Shumway-Cook & Woollacott, 2006; Gallahue & Ozmun, 2006; Martín Sanz et al., 2004; Bertoti, 2004) as shown on the diagram below. These factors will depend expressly on the required task and the posture.

Figure 1. Processes necessary for postural control.
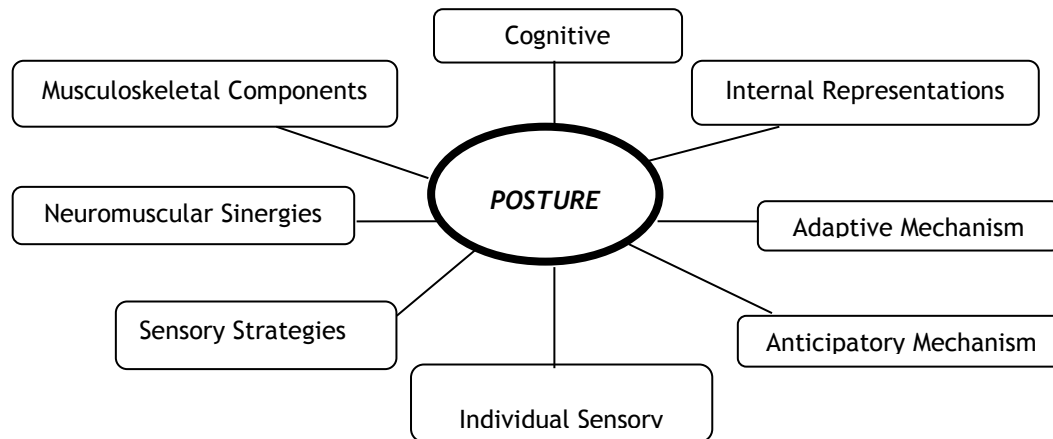
Adaptative postural adjustments imply the modification of sensory and motor systems in response to the changes on the environment and the tasks' demands. The anticipatory aspects of *posture control,* pre adjusts the sensory and motor systems for postural demands based on previous learning and experience. Other cognitive aspects that affect *posture control* include processes like attention, motivation and intention [(]Shumway-Cook & Woollacott, 2000; 2006).

The cycle for *postural control* consists firstly on the *input* of *sensory information,* needed to make a functional movement; it is necessary to previously analyze the data from three systems: vestibular, visual and somatosensory; in order to *assimilate this information* and receive a *respond*. When all these systems are intact, individuals show an adaptable *postural control* and are capable of finding stability and orientation goals in any context.

Standing position is characterized by small spontaneous balancing and postural oscillation (Shumway-Cook & Woollacott, 2006; Kandel, 2000), in which various factors contribute to stabilize this situation, factors such as: *body alignment*, that minimizes the gravitational forces effects with less energy waste; *muscle tone*, the muscles' resistance while being stretched and where the stretch reflexes play a feedback role during the balance position, while aiding balance control in standing position; and the *postural tone*, which is increased by the antigravity muscle activity when standing in order to control the gravitational force. In scientific literature, the postural

tone acquires an important relevance, as main mechanism in the body's support against gravity. Many investigators have suggested that postural tone is the most important element for normal stability control in the erect posture (Shumway-Cook & Woollacott 2000; 2006).

In fact, researchers suggest that *postural control* includes an active sensory process that calculates the position of the body in space and predicts where it is going and what actions will be necessary to control that movement. Taking into consideration the previous experience, the object of action and the context in which it is carried out (Geurts, 1996).

An effective *postural control* requires more than the capacity to generate and apply forces that control the body's position in space. In order to know how and when to apply force, the Central Nervous System (CNS) must have an exact image of where the body is and if it is standing still or in movement (Geurts, 1996).

These factors or necessary systems to maintain an adequate *postural control* are altered in patients that have suffered an Acquired Brain Injury (ABI). "ABI is an injury produced in the brain structure as a sudden damage in people born without any kind of brain injury, who later on suffer brain injury as a consequence from an accident or illness" (Feigin, 2010; Strong, 2007; Langlois, 2006). The etiology of a sudden brain injury may be vascular (cerebrovascular accident -CVA-), traumatic (traumatic brain injury –TBI-), tumorous, by infection or anoxia (Feigin, 2010).

As for the rehabilitation in these types of patients, it is considered that after the first six months, the possibility of spontaneous recovery is very limited, therefore we can assert that improvements from this period on, will be the consequence of therapeutic act (Feigin, 2010; Strong, 2007; Langlois, 2006).

In 2008, Nintendo Wii® was the last generation of consoles in the videogame domain to include innovative characteristics that justified its position among the top ranked in sales in the US. Since it is a low cost and recreational solution, its particular form of interaction has raised interest among health professionals to be used on the field of rehabilitation. All these professionals agree in that it allows participants to forget they are working, taking their

limits further. Moreover, the patients' interaction with the console also allows them to reinforce the motivational aspect and facilitates adhesion to the treatment which is an important point that influences on *postural control*.

Nowadays, the implementation of virtual therapies as a complement to the treatment of acquired brain injury (ABI) in motor and cognitive limitations, is being widely accepted in neurorehabilitation (Laver, 2011; 2012). A current revision on the subject (Laver, 2012) has obtained significant results on this kind of approach of improvement in these patients.

The object of this study is to assess the effects of a neurological rehabilitation treatment combined with Nintendo's Wii®, using Wii Fit and Wii Balance Board (WBB), on some relevant aspects of *posture control* on patients that have suffered from ABI.

On the present study, the task will consist in getting up, walk three metres, go back to the start and return to a sitting position.

Methodology

Design of the experiment

A quasi-experimental study was carried out in a test-retest format, with the objective of validating the effects of Nintendo Wii® on the *posture control* of a group of patients with ABI. The experimental and control groups were tested in parallel.

Participants were assigned in a proportion of 1:1 in both study groups, according to the subjects' schedule and the availability of the virtual hall. Patients from the experimental group couldn't be slanted in favor of the intervention.

The game system Nintendo Wii® launched in 2005, introduced a new virtual reality style through a wireless control that interacts with the player while detecting movement and its avatar performance on the video.

The WBB contains four transducers used to evaluate the distribution of the load and the resulting movements from the center of the mass. Originally designed as a videogame device, the WBB is mainly used in combination with associated software and a console. Due to its capacity to provide instant information and the possibility to improve the motivational level (Ramchandani, 2008), this system is already part of the rehabilitation programs of neurological patients with balance disorders.

The information on the screen produces a positive reinforcement and gives feedback for the patient, facilitating the execution and the progress of the task. (http://www.nintendo.com/wii/what).

The data was gathered at Centro LESCER, in Madrid, from November 2010 to January 2011.

Participants

This study was held at the Centro Lescer in Madrid, with the participation of 12 patients who have suffered from ABI and who are following individual rehabilitation programs according to their needs and involving different areas like: physiotherapy, occupational therapy, neuropsychology and speech therapy.

A not random sample was chosen: 6 patients from ischemic stroke (50%), 3 from hemorrhagic stroke (25%), 1 TBI (8.3%), 1 brain tumor (8.3%) and 1 hypoglucemic coma (8.3%), all of them already diagnosed. A high percentage of patients in this sample are vascular patients, as it occurs in ABI (Feigin, 2010; Strong, 2007; Langlois, 2006).

The 12 participants in the group were 2 women and 10 men. The mean age was of 62.17 ± 11.535, with a minimum of 39 and a maximum of 79 years old. On the other hand, according to the elapsed time since the injury, the group consisted of 4 acute patients (33.3%) and 8 chronic patients (66.7%). The first 6 subjects were included in the experimental group, while the last 6 were in the control group.

The general characteristics of the subjects are exposed in Table 1.

Table 1. General characteristics of the subjects.

Subject nº	Gender	Age	Elapsed time since injury (months)	Diagnosis	Group
Subject 1	Male	59	4 (Acute)	Hemorrhagic Stroke	Experimental
Subject 2	Female	45	40 years (Chronic)	Traumatic brain injury (TBI)	Experimental
Subject 3	Male	39	43 (Chronic)	Tumour	Experimental
Subject 4	Male	65	28 (Chronic)	Coma	Experimental
Subject 5	Male	70	12 (Chronic)	Hemorrhagic Stroke	Experimental
Subject 6	Male	65	4 (Acute)	Hemorrhagic Stroke	Experimental
Subject 7	Male	54	43 (Chronic)	Ischemic Stroke	Control
Subject 8	Female	79	7 (Chronic)	Ischemic Stroke	Control
Subject 9	Male	60	22 (Chronic)	Ischemic Stroke	Control
Subject 10	Male	71	26 (Chronic)	Ischemic Stroke	Control
Subject 11	Male	68	6 (Acute)	Ischemic Stroke	Control
Subject 12	Male	71	6 (Acute)	Ischemic Stroke	Control

Inclusion Criteria

Among the patients with brain injury, only the ones who surpassed or leveled up to the 2nd category of the Functional Ambulation Categories scale (FAC) (Leder, 2008; Holden, 1984) were included, this means, patients who needed continuous or intermittent human support to assist balance or coordination when walking. From these, only patients who are medically stable, alert and who are capable of following simple verbal commands.

Exclusion Criteria

Patients with global aphasia, non collaborating patients like those with conduct problems, medically unstable patients or those to whom risk may be implied.

▪ Study

At the beginning of the study (pre-test), 4 validated tests were administered: FAC (Functional Ambulation Categories scale), Berg Balance Scale (BBS), Barthel Index (BI) and the Timed Get Up and Go test (TGUG). These are reliable, sensitive and validated tests used to assess *postural control*, therefore adequate for this study according to the investigators criteria. Below is a brief description of these scales:

- FAC: categorizes patients according to basic motor skills necessary for functional ambulation. It does not assess endurance. Describes six categories according to the assistance required. Patients should be rated at their most independent level (Leder, 2008; Holden, 1984).

- BBS: evaluation instrument used to identify balance disorders through functional activities such as reaching over, leaning over, transferring objects, standing on one leg, and so on, until completing 14 tests. Balance is the capacity of maintaining the centre of gravity on a support base, generally on a vertical position. With a good balance a patient will be able to sit, get up or walk safely. BBS was developed in the 1990's to assess balance in elderly individuals. It has been proved to be a consistent and reliable tool. Some tasks are classified according to the performances' quality, while others are assessed by the time needed to accomplish the task (Mehrholz, 2007).

- BI: Evaluates a person's ability to independently or dependently perform 10 activities of daily living like feeding oneself, bathing, dressing, grooming oneself, bowel and bladder control, toilet use, bed/armchair transfers, mobility on level surfaces and stairs; and assigns a score according to the time and assistance needed. Designed in 1965 by Mahoney and Barthel to measure the subjects' evolution with neuromuscular and musculoskeletal processes in a hospital for chronic patients in Maryland (Berg, 2004).

- TGUG: is a mobility and locomotor performance test that includes several tasks such as standing up from a sitting position, walking, turning around, standing still and sitting. All the important tasks necessary for a persons' mobility independence are represented in

this test. The score consists of the time taken to complete the test activity (Mahoney & Barthel, 1965).

TGUG can be characterized by several non temporal criteria of postural nature (this is one of the most unknown aspects of this test) (Podsiadlo, 1991; Fife, 2000), that best define the subjects' functional capacity. Patients with balance disorders must adequately coordinate certain trunk adjustments in anteroposterior and lateral scenes (Allum, 2001).

Apart from the four quantitative tests already mentioned, the subjects were assessed every day the Wii® was used, with the consoles' test, for example, the *walking test* (walking 20 steps; gives as a result the lower body members' weight distribution percentage), the *agility test* (moving the center of gravity–represented on the screen- in a determined time), the *statue test* that consists on standing on both legs or the *one leg test*, standing in one leg (that provides the stability percentage while we continue standing on both legs or one, respectively).

All patients have been following a neurological rehabilitation program for several months, since all of them have both motor and cognitive disorders, as consequence of the ABI. This Program consists of two 45 minute daily sessions of physiotherapy, occupational therapy, speech therapy and neuropsychology depending on the patient's intrinsic goals and needs. On the experimental group, one of the weekly sessions of physiotherapy was substituted by the Wii® (45 minutes/week).

Subjects who have participated on the investigation, have worked on the WBB platform executing the different balance games of the console, once a week during 10 weeks, as well as their routinary physiotherapy and occupational therapy. The WBB and Wii Fit allow the patient to practise balance exercises in standing through the centre of gravity monitoring and visual feedback.

You can see the results of each test on the following table.

Table 2. Results of pre-test and post-test (Functional Ambulation Categories scale, Berg Balance Scale, Barthel Index, Timed Get Up and Go test) of the subjects.

TEST	FAC[1]		BBS[2]		BI[3]		TGUG[4]	
Subject nº	Pre	Post	Pre	Post	Pre	Post	Pre	Post
Subject 1	3	4	34/56	47/56	35/100	75/100	23.02	13.85
Subject 2	5	5	52/56	53/56	95/100	95/100	7.72	5.40
Subject 3	5	5	51/56	52/56	100/100	100/100	10.09	7.72
Subject 4	5	5	44/56	48/56	95/100	95/100	20.47	14.32
Subject 5	4	5	43/56	53/56	100/100	100/100	13.97	11.40
Subject 6	5	5	51/56	54/56	100/100	100/100	8.87	6.31
Subject 7	4	4	38/56	38/56	85/100	90/100	27.94	20.50
Subject 8	2	3	13/56	33/56	15/100	55/100	65.54	36.15
Subject 9	4	4	45/56	50/56	95/100	95/100	12.22	9.40
Subject 10	4	4	51/56	52/56	85/100	85/100	20.83	18.37
Subject 11	5	5	50/56	52/56	90/100	90/100	7.40	6.41
Subject 12	4	5	38/56	55/56	85/100	95/100	17.23	12.30

Table 3. Comparison of parameters before the treatment.

Test	Control	Experimental	P - value
FAC[1] Pre	3.83 ± 0.983	4.50 ± 0.837	0.022
BBS[2] Pre	39.17 ± 13.992	45.83 ± 6.97	0.200*
BI[3] Pre	75.83 ± 30.069	87.50 ± 25.836	3.022×10^{-5}
TGUG[4] Pre	25.19 ± 20.99	14.02 ± 6.39	0.39

[1] Functional Ambulation Categories scale

[2] Berg Balance Scale

[3] Barthel Index

[4] Timed Get Up and Go test

Statistical Analysis

The statistical software SPSS v.17.0 was administered. The descriptive data is expressed as average ± standard deviation. The statistical inference was based on the *t of student* and non-parametric tests on two associated samples (*Wilcoxon test*), to compare the initial and final values of every scale in each separate group. Non-parametric tests were also administered on two independent samples (*Mann-Whitney U test*) with the purpose of comparing the scale difference (final value minus initial value for each scale) on the control and experimental group. The *Kolmogorov-Smirnov test* was carried out as test for normality (with the correction of the Lilliefors significance). The level of statistical significance gathered in the study was of 95% ($p \leq 0.05$).

All 12 patients were evaluated; therefore there weren't any lost values.

Results

In view that the TGUG represents the task presented throughout this study, the analysis began by verifying differences in the initial *TGUG*, according to the group they were in, control or experimental. With a confidence index of 95% [9.50 – 29.71], no relevant statistical differences were found among this

two groups, therefore we can't discard the possibility that it may all be down to chance (p=0.39). (Table 3)

The neurological rehabilitation program isolated or associated to the Nintendo Wii® console contributed to substantial improvement while comparing the FAC (functional walk) test results, before and after the treatment on both; the control group, (4.17 ± 0.753 compared to 3.83 ± 0.983; p=0.02) as on the experimental group (4.83 ± 0.408 compared to 4.50 ± 0.837; p= 0.02). The 95% Confidence Index on the pre-post score difference for both groups is of: [(-0.21) – 0.87]. (Table 4)

As for the balance test (BBS): an improvement has been made in both treatment groups, even though it has not been statistically significant in neither (7.5±8.734; p=0.157 >0.05 in the control and 5.33±5.006; p=0.189 >0.05 in the experimental). In the first, the 95% CI of the remainder from the final minus the initial score is of: [(-1.67) – 16.67)], while in the second, the 95% CI is of [0.08 – 10.59]. (Table 5)

Regarding the daily living activities (ADL), the experimental group displayed a significant improvement in the final measurements (94.17 ± 9.704) compared to the 87.50 ± 25.836 from initial measurements, p= 4.6x10-5 <0.05; CI 95% [(-10.47) – 23.80]; while in the control group, (85.00 ± 15.166 compared to 75.83 ± 30.069), p= 0.069 >0.05; CI 95% [(-7.23) – 25.56], even though there was improvement, it was not significant. (Table 6)

Finally, in order to evaluate the task suggested in this study, we will analyze the results that reflect this activity on the TGUG. We noticed that the time was significantly reduced (p= 0.007 <0.05) in participants who used the Nintendo Wii® console (9.83 ± 3.88 compared to 14.02 ± 6.39) as well as in those who only received conventional rehabilitation (17.19 ± 10.70 compared to 25.19 ± 20.99), p= 0.018 <0.05). The 95% CI is of [(-7.1853) - (-1.1947)] and [(-19.25) – 3.24], respectively. (Table 7)

As statistical method to determine clinical significance, it follows the standard error of measurements (SEM) used to calculate the minimum detectable change. This was calculated as the standard deviation by taking the square root of the sample size.

Table 4. Comparing the FAC test results, before and after the treatment on both group.

Test group	Prior	SEM[2]	Post	SEM[1]	Difference and Probability
FAC[1] Control Group	3.83 ± 0.98	0.401	4.17 ± 0.75	0.307	0.33 ± 0.51 p=0.02 p<0.05
FAC Experimental Group	4.50 ± 0.84	0.341	4.83 ± 0.41	0.167	0.33 ± 0.51 p= 0.02 p<0.05

[1] Functional Ambulation Categories scale

[2] Standard error of measurements (SEM)

Table 5. Comparing the Berg Balance Scale test results, before and after the treatment on both group.

Test group	Prior	SEM	Post	SEM	Difference and Probability
BBS[1] Control Group	39.17 ± 13.99	5.712	46.67 ± 8.94	3.649	7.5 ± 8.734 p= 0.157 p>0.05
BBS Experimental Group	45.83 ± 6.97	2.845	51.17 ± 2.93	1.196	5.33 ± 5.006 p= 0.189 p>0.05

[1] Berg Balance Scale

Table 6. Comparing the Barthel Index test results, before and after the treatment on both group.

| Test group	Prior	SEM	Post	SEM	Difference and Probability
BI[1] Control Group	75.83 ± 30.07	12.275	85.00 ± 15.17	6.193	9.166 ± 15.625 p= 0.069 >0.05
BI Experimental Group	87.50 ± 25.84	10.547	94.17 ± 9.70	3.96	6.667 ± 16.329 p= 4.6×10^{-5} <0.05

[1] Barthel Index

Table 7. Comparing the Timed Get Up and Go test results, before and after the treatment on both group.

Test group	Prior	SEM	Post	SEM	Difference and Probability
TGUG[1] Control Group	25.19 ± 20.99	8.569	17.19 ± 10.70	4.368	-8.00 ±10.71 p= 0.018 <0.05
TGUG Experimental Gr.	14.02 ± 6.39	2.608	9.83 ± 3.88	1.584	-4.1900 ±2.85 p= 0.007 <0.05

[1] Timed Get Up and Go test

The graph below shows the resulting values of the difference between the pre and post scores of each test, from both the experimental and the control group:

Figure 2. Resulting values of the difference between the pre and post scores of each test on both group
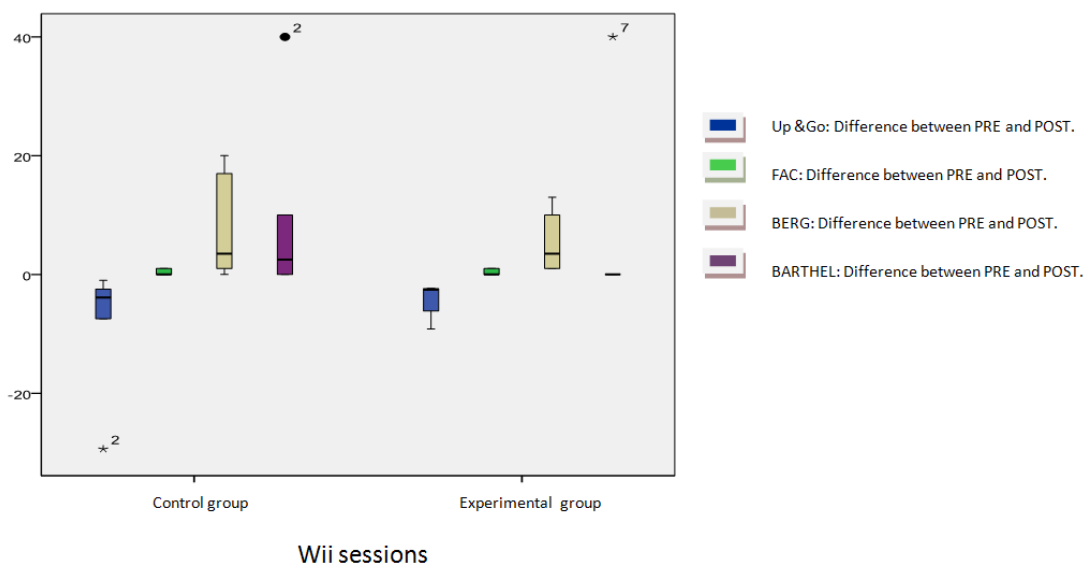
Discussion

Even though there has been a great improvement in all the tests administered to each patient, and as a result an improvement on the performance of the suggested task, not all have been statistically significant. Hence, and although there was a p=0,02 on the *FAC*, the confidence index of

95% holds the value zero, indicating that this improvement on the score is not statistical significant. As well as occurring in the BI scores of the experimental group (p<0.001) once again, we find a 95% CI that holds value zero, making it impossible to reject the invalid hypothesis. This also happens on the *TGUG* from the control group.

However, as it has been also stated in other researches (Gil-Gómez, 2011), the virtual rehabilitation provides significant improvements in *postural control* and balance. Regarding the ADL, the positive significant differences are also demonstrated in a recent virtual reality review (Laver, 2011).

Nevertheless, we found significant results while discarding the possibility that chance might explain the results, on the *TGUG* of subjects who benefitted from the Nintendo Wii® console (p=0.007). We found that when performing the task of standing up, walking three meters, turning around, walking back three meters and sitting down, the time used in this activity will diminish 1.19 and 7.18 seconds, with a confidence of 95%.

Although this research represents a small step, the investigation should ideally continue with a bigger and more homogenous sample to avoid slants. Also, the elapsed time since the injury has varied significantly from one individual to another, influencing to a large extent on the statistical analysis. As for the distribution of patients on both groups, a randomization process must be fulfilled, in order to have the lowest influence on the results.

An important limitation in this study has been the time at hand to collect data (November 2010 – January 2011), and the limited availability to the console (if it was being used by another subject), leaving each individual from the experimental group, with 10 sessions of 45 minutes. Therefore a continuation of the investigation searching long-term results is recommended.

Nevertheless, it would be useful to continue this line of study, by separately analyzing the effects of the different treatments, for example, the console with physiotherapy, or console with occupational therapy, or the Wii® alone.

Future lines should also consider larger samples and random assignment.

Apart from its usage as a feedback and amusement tool, the WBB can also be used by healthcare professionals to compile and analyze balance data, by using techniques and precise measurement results for targeted patients. The WBB is a small fraction of the cost of a dynamic posturography laboratory, it is commercialized in large scale and portable, therefore it has the potential of turning into a key component of tests, given it is possible to demonstrate that liable and valid results can be generated.

Conclusions

The effects of a neurological rehabilitation treatment in addition to Nintendo Wii have been shown to provide benefits to our sample of ABI patients related to *postural control* and balance. This was reflected on a task evaluated through TGUG, demonstrating that the time used to achieve the objective has diminished significantly on the experimental group.

The sensory process of the visual system needed to obtain *postural control*, took place thanks to the feedback displayed on the screen, the patient needed to calculate the body's position in space and accomplish the necessary actions to control the required movements.

Moreover, it could be concluded that the patients with virtual therapy included in their neurorehabilitation program, obtained significant improvements in their ADL and, therefore, in their *functionality*. Although the Barthel Index revealed improvements in the control group, these have not been statistically significant.

Due to the characteristics of this device and the positive results obtained, it seems to be of great utility as a complement to the traditional rehabilitation therapy for patients with ABI.

This highlights the need to create an available low cost, portable system to evaluate balance. The Wii Balance Board (WBB) part of the popular videogame Wii Fit, accomplishes all these criteria.

In short, by improving *postural control* these patients can obtain a higher functionality.

Acknowledgements

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A HOME-BASED NEUROREHABILITATION SYSTEM FOR CHILDREN WITH UPPER EXTREMITY IMPAIRMENTS

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Abstract: The objective of this paper is to introduce a novel low-cost human-computer interface (HCI) system for home-based massed practice for children with upper limb impairment due to brain injury. The proposed system targets motions around the wrist. Successful massed practice, a type of neurorehabilitation, may be of value for children with brain injury because it facilitates impaired limb use. Use of automated, home-based systems could provide a practical means for massed practice. However, the optimal strategy to deliver and monitor home-based massed practice is still unclear. We integrated a motion sensor, video games, and HCI software technologies to create a useful home-based massed practice at targeted joints. The system records joint angle and number of movements using a low-cost custom hand-held sensor. The sensor acts as an input device to play video games. We demonstrated the system's functionality and provided preliminary observations on usage by children with brain injury and typically developing children, including joint motions and muscle activation.

Keywords: massed practice, brain injury, home-based video game system.

Introduction

Patients with motor dysfunction due to musculoskeletal or nervous system injuries need structured repetitive motion practice to gain or regain muscle power or fine or gross motor control (Saposnik & Levin, 2011). Motivation and safe practice at home are key factors for increasing practice of prescribed motions (Wu, Wilcox, Donoghue, Crisco, & Kerman, 2012). Thus, what is needed are a system and a method to safely and reliably deliver massed practice at the home-setting which tailor the needs/capabilities of each individual to achieve the therapeutic benefits and record and track the performance changes.

Cerebral palsy (CP) is the most common pediatric developmental disability (Arneson et al., 2009) corresponding to a high estimated lifetime cost of $921,000 per case ("Economic costs associated with mental retardation, cerebral palsy, hearing loss, and vision impairment--United States, 2003," 2004). Hand function deficits seen in the children with hemiplegic cerebral palsy prevent children from achieving their full potential for healthy and productive lives and independence. Improved hand function can dramatically broaden lifelong opportunities necessary for normal social and motor skill development as well as employment.

Results from studies of constraint-induced movement therapy and robotic therapy demonstrate that massed practice (intensive training within a set period of time) can improve arm function (Charles, Wolf, Schneider, & Gordon, 2006; Lo et al., 2009). However these approaches are expensive and require intensive supervision. As a result these forms of increasing therapy cannot be made available to every child. Home-based therapy coupled with recent advances in electronic technologies could provide children with a low-cost way to carry out massed practice. Current forms of home-based therapy, such as conducting therapist-prescribed range-of-motion exercises and potentially playing video games with commercial controllers (Andrysek et al., 2012; Deutsch, Borbely, Filler, Huhn, & Guarrera-Bowlby, 2008; Hurkmans, van den Berg-Emons, & Stam, 2010; Lange, Flynn, Proffitt, Chang, & Rizzo, 2010) are used to extend therapy into the home setting. However, self-conducted exercise, such as range of motion exercise, has the

disadvantage of compromised compliance due to a lack of motivation in patients (Jurkiewicz, Marzolini, & Oh, 2011). Moreover self-conducted exercise and current forms of video game-play for selective motor control neither take undesirable motions into consideration nor do they tightly constrain training movements. Hence current home-based therapies fail to meet the objective of highly-structured practice (repetitive training of specific joints along with real-time feedback) that is preferred for promoting recovery (Kerr, Cheng, & Jones, 2011; Levin, Kleim, & Wolf, 2009; Taub, Uswatte, & Pidikiti, 1999). However, a number of studies have demonstrated the efficacy of video game-based rehabilitation for a broad range of conditions, mostly for adult clients. For example, video game-based systems have been developed to provide motor rehabilitation to adults who have suffered strokes(Cameirao, Bermudez, Duarte Oller, & Verschure, 2009). Golomb et al used gloves equipped with angle detecting sensors to rehabilitate hand functions of three adolescents with hemiplegic cerebral palsy. They found improvements after 3 months of training (Golomb et al., 2010) lasted for 14 months reported in the case report(Golomb et al., 2011). Video-game based rehabilitation holds great promise as an effective means to extending therapy beyond the clinical setting. Currently, however, there is no system which can target specific wrist joint and monitor the patient's exercise in the home environment. To overcome these deficits we introduce a novel low-cost human-computer interface named the Neuroplasticity-targeted Exercise Trainer (NExT) that will provide highly-structured upper limb rehabilitation at home.

Neuroplasticity-targeted Exercise Trainer

System overview

The NExT system is designed to allow patients to repetitively perform therapeutic exercise of their hemiparetic wrist using video game-play. Figure 1 shows the prototype of the NExT, which consists of a personal computer, a custom hand-held controller, an armrest equipped with five sensors, and a friendly graphic user interface (GUI) that permits patients to play a wide

range of video games (for example, free online arcade games or commercialized rehabilitative games("www.nanogames.com,").

Figure 1. NExT.

Easy to hold lightweight controller

The controller design had to conform to a number of requirements in order to be compatible with the patients' physical limitations. In another study, a modified wheelchair joystick was interfaced with a computer and other electronic device for wheelchair users (Casas et al., 2012), but such a configuration would not meet our needs, as it does not allow for wrist joint exercises performed in free space. The joystick is restricted in its degrees of freedom and must be used with a mounted based. And the mounted based is not aligned with the axis of wrist joint, this off-axis alignment can restrict wrist movement. A lightweight (<30g) and easy-to-grasp controller (Figure 2) is designed especially for children with brain injury who have insufficient grasp strength and are unable to use off-the-shelf game controllors. An egg-shaped controller provides a spherical surface for palmer grasping for children with weak grip strength. This egg-shaped controller detects prescribed joint motions. It consists of two parts: an inertial measurement unit (MinIMU-9, Pololu) and a microcontroller board (Pro Micro 5V/16MHz, Sparkfun). The 3-axis accelerometer (LSM303DLH) and 3-axis gyroscope (L3G4200D) onboard the MinIMU-9 are used to detect and measure rotations about each of the three axes of the controller. The Pro Micro, utilizing the

ATMega 32U4 microcontroller from Atmel, is used to interface with the IMU and relay the linear acceleration and angular velocity measurements to the computer via UART serial communication protocol over one of the computer's USB ports.

Custom Chair with Posture Control Armrest

Figure 2. The handheld controller and its internal components. (a) The controller as a whole. (b) The IMU. (c) The microcontroller board.
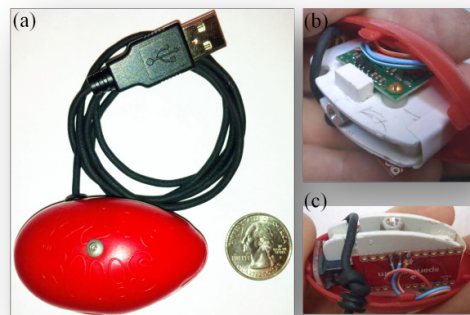

A modified chair is incorporated into the system (Figure 3). The chair has an attached assembly on its right side on which the posture control armrest is positioned. To adjust for variation in size among different patients, the assembly is constructed to permit repositioning of the armrest in all three dimensions. This ensures that the patient can rest their forearm naturally and comfortably.

Figure 3. Chair with armrest assembly.

An essential part of the chair is a digitally equipped armrest (Figure 4). It limits the patient from executing compensatory proximal joint movements while performing his/her exercises. Many patients with motor structure injury make posture changes while performing wrist motions. They have often been observed to rotate their shoulder or elbow instead of their wrist when operating controllers. Importantly the armrest discourages this compensatory movement, ensuring that the affected muscle groups are targeted by the therapy.

Figure 4. Digitally equipped armrest
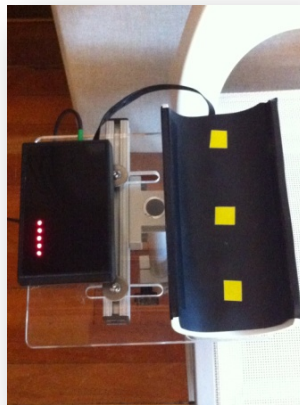
The armrest is constructed out of a section of PVC pipe (measuring the length of a child's forearm) cut longwise to form a u-shaped platform (cylindania). The inner surface of the half-pipe is lined with a soft rubber sheet (2 mm thick) on which the patient can rest their hemiparetic forearm. Fixed underneath the rubber sheet are five digital pushbutton switches spread out linearly across the platform which become depressed when a patient's forearm is placed on top. With the patient's forearm rested on the armrest, the depressed state of the switches indicates that the patient is using their wrist rather than their proximal joints to operate the controller during game-play. The digital electrical states of the switches are serially communicated to the computer over a different USB port by a separate Pro Micro microcontroller board.

Graphical User Interface

The core of the HCI is a java based program that serves as a graphical user interface (GUI). It guides patients through the exercise, provides them with interactive feedback and logs their usage. Through serial communication, the GUI reads the incoming data from the controller and passes that information to the gesture recognition application, the Wekinator, over Open Sound Control protocol. The Wekinator is an open source software package for gesture recognition which classifies the gestures using input features (the acceleration and angular velocity data from the controller generated by the patient's wrist motions) and patches them to any output parameters determined by the therapist (the control signals for game-play). The gesture classification is done by a machine learning process using K-Nearest Neighbor and Support Vector Machine algorithms. Training examples of controller orientations are submitted to the Wekinator to develop a wrist motion-control signal pairing model which will be loaded at the beginning of each exercise session. Table 1 is an example of the corresponding wrist motions (input features) to the control signals (output parameters) for game-play. For a more detailed explanation of the Wekinator, please see the reference.(Fiebrink, 2011; Morris & Bartlett, 2004)
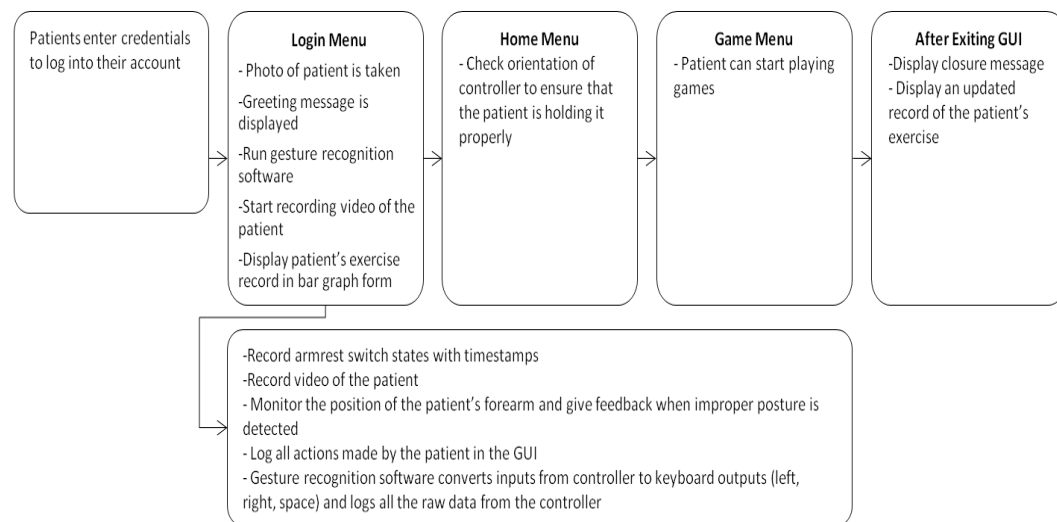
Table 8. control signals v.s. wrist motion (Right Hand)

Control signal	Wrist motion
Move to the Right	Supination
Move to the Left	Pronation
Right + Action	Supination + Wrist Flexion
Left + Action	Pronation + Wrist Extension
Action	Radial Deviation

The GUI is organized into four menus – the login menu, home menu, games menu, and settings menu. Figure 5 illustrates the process of navigating through the GUI. The patient logs into his/her account using his/her

credentials and is then presented with a greeting. Moving on to the home menu, the patient goes through a preparation process. When this is completed, the patient is allowed to start playing games. The settings menu is where the therapist can adjust various GUI settings for the patient. A more detailed explanation of the menus is discussed below.

Figure 5. Flowchart illustrating how the GUI operates.

| Patients enter credentials to log into their account | **Login Menu**
- Photo of patient is taken
-Greeting message is displayed
-Run gesture recognition software
-Start recording video of the patient
-Display patient's exercise record in bar graph form | **Home Menu**
- Check orientation of controller to ensure that the patient is holding it properly | **Game Menu**
- Patient can start playing games | **After Exiting GUI**
-Display closure message
- Display an updated record of the patient's exercise |

-Record armrest switch states with timestamps
-Record video of the patient
- Monitor the position of the patient's forearm and give feedback when improper posture is detected
- Log all actions made by the patient in the GUI
- Gesture recognition software converts inputs from controller to keyboard outputs (left, right, space) and logs all the raw data from the controller

Login and informative greeting

Upon running the GUI, the patient is taken to the login menu (Figure 6) where they are prompted to enter a username and password in order to access their personal account. Once this is accomplished, the GUI takes a photo of the patient for identity verification. As the GUI collects important data related to the patient's exercise, it is important that we know if the patient's sibling(s) log(s) into the patient's account, which would otherwise compromise the legitimacy of the data. An audio greeting is then played depending on the date and time of the previous login, as shown in Figure 7.

Figure 6. Login menu as would appear the first time the patient logs into his/her account. (a) The four menus of the GUI depicted in tab-style. (b) A disclosure informing the patient that pictures and video of the patient will be taken during each exercise session. (c) The greeting message. (d) Tells the patient how to move onto the next menu. (e) A chart depicting how many minutes of game-play the patient has accomplished each day.
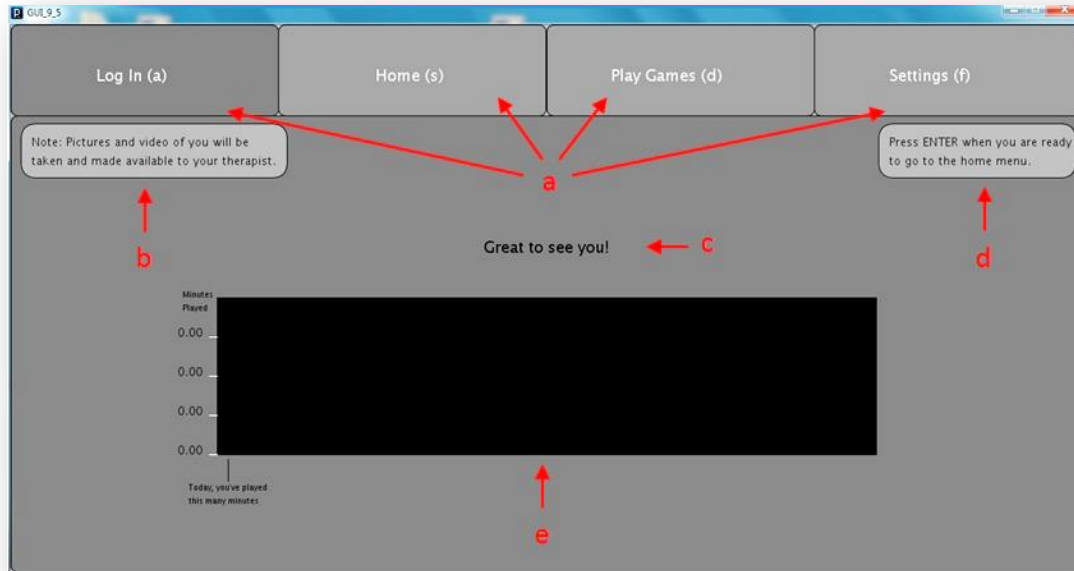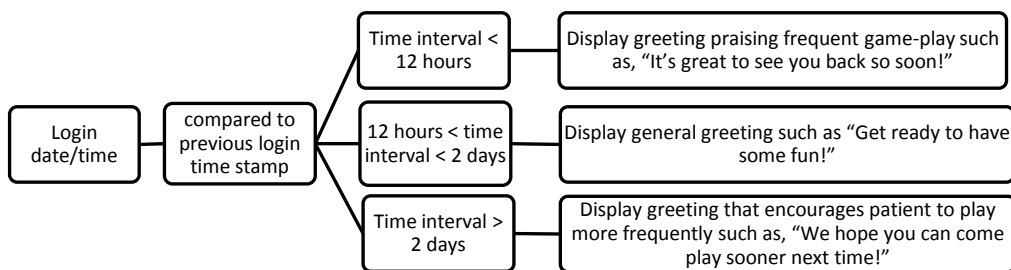

Figure 7. Flow chart illustrating the greeting message selection process.

If no longer than twelve hours has elapsed after the previous login, a congratulatory message is generated to praise the patient for their frequent exercise. If the time between logins is greater than two days, a message that encourages the patient to play more frequently is given. For in-between cases, a general greeting is given. The greetings are presented in this manner in order to promote frequent exercise. Furthermore, once logged in, the GUI runs the Wekinator and loads the appropriate wrist motion-control signal pairing model, thereby allowing the controller to

emulate the left, right, and spacebar keys for game-play. In addition, the login menu displays a chart depicting the total number of minutes the patient has spent playing games each day. Figure 6 shows the play chart as it would appear the first time the patient logs into his/her account. If they were to play for 7 minutes during that exercise session, for example, upon exiting the GUI, the closure screen (which will be discussed later) would display an updated play chart as in Figure 8. Figure 9 shows what might the play chart look like after 21 days into the home-based therapy.   By continually displaying their progress, the chart helps motivate the patients to perform their exercises more frequently and for longer periods of time. Lastly, for offline posture analysis, the GUI starts recording video of the patient at a resolution of 640x480 pixels, 30 frames per second.

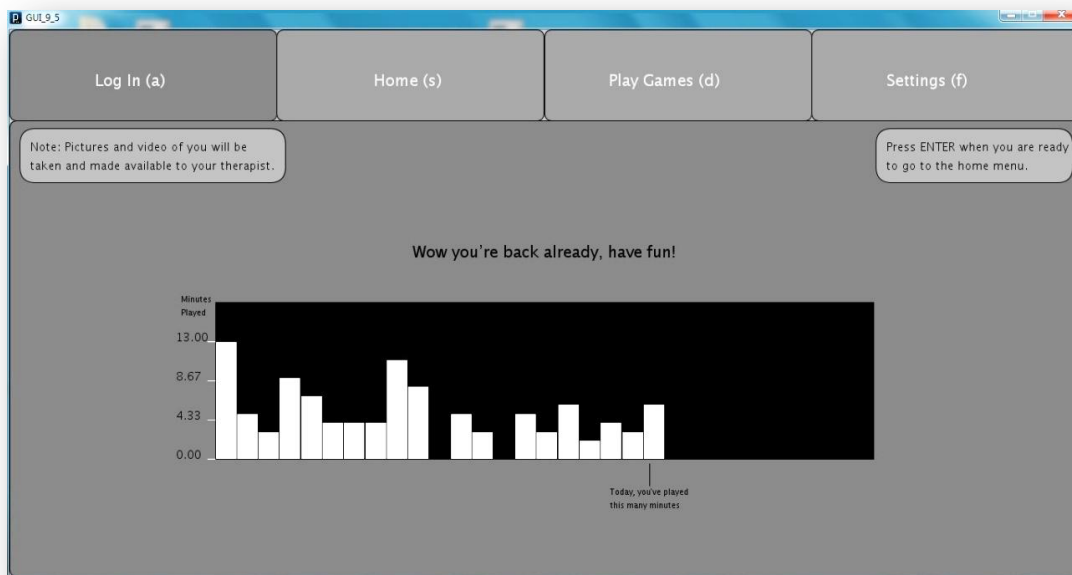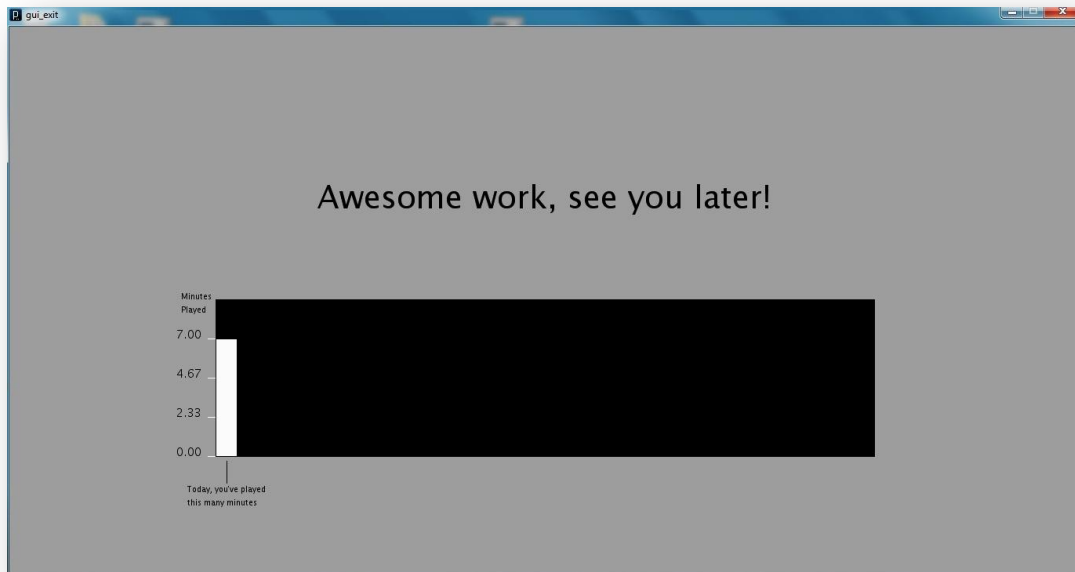*Figure 8. Closure screen showing an updated play chart and a goodbye message.*

Figure 9. Login menu as it might appear 21 days into the home-based therapy

Home menu

Once logged in, the patient is free to navigate to the home menu (Figure 10), in which they are shown an image of themselves holding the controller in the starting position (the starting position is the orientation that corresponds to no control signal). They are then told to hold the controller in the same manner as in the picture for a time period of two seconds. Due to the symmetrical shape of the controller, there exists the possibility of the patient holding the controller upside-down. Therefore, to ensure that this does not happen, the GUI reads the accelerations in the x, y, and z axes of the controller and compares these values to what they should be if the controller is held in the neutral orientation. If the two sets of acceleration values agree and the patient maintains this controller orientation for the required two second time period, they are automatically taken to the game menu. If not, a message is voiced informing the patient that they might be holding the controller upside-down.

Figure 10. Figure 10. Home menu. (a) Picture of patient holding the controller in the neutral orientation. (b) Progress bar showing how much longer the patient must maintain this controller orientation.
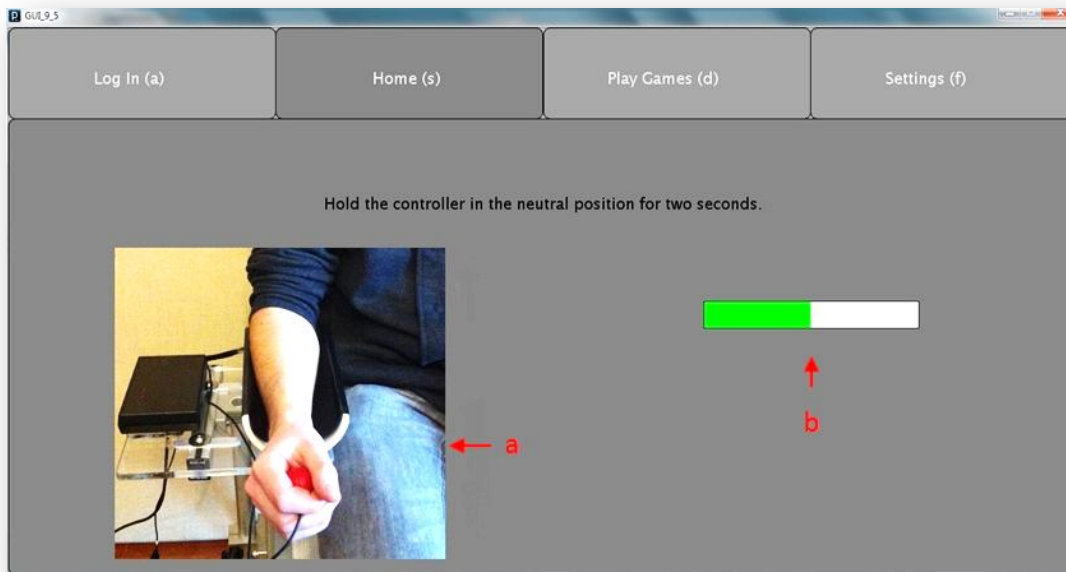
Settings menu

The fourth menu of the GUI, the settings menu (Figure 12), is accessible only by the therapist. In this menu the therapist is able to modify various HCI related settings particular to the patient. One of the settings the therapist may modify is whether a Skype conversation between patient and therapist is automatically initiated upon patient login. During the early stages of the home-based therapy, the therapist may want to observe the patient's posture in realtime and the Skype video conference would make that possible. This menu also allows the therapist to configure various Wekinator settings, including the pairing model the GUI will load once the patient logs in. This is important as it allows the therapist to adjust the therapy as needed. For example, if the patient shows improvement in their wrist range of motion, the therapist may configure the Wekinator to require wider angles of wrist rotation from the patient in order to operate the controller.

Figure 11. Game menu showing different game options.
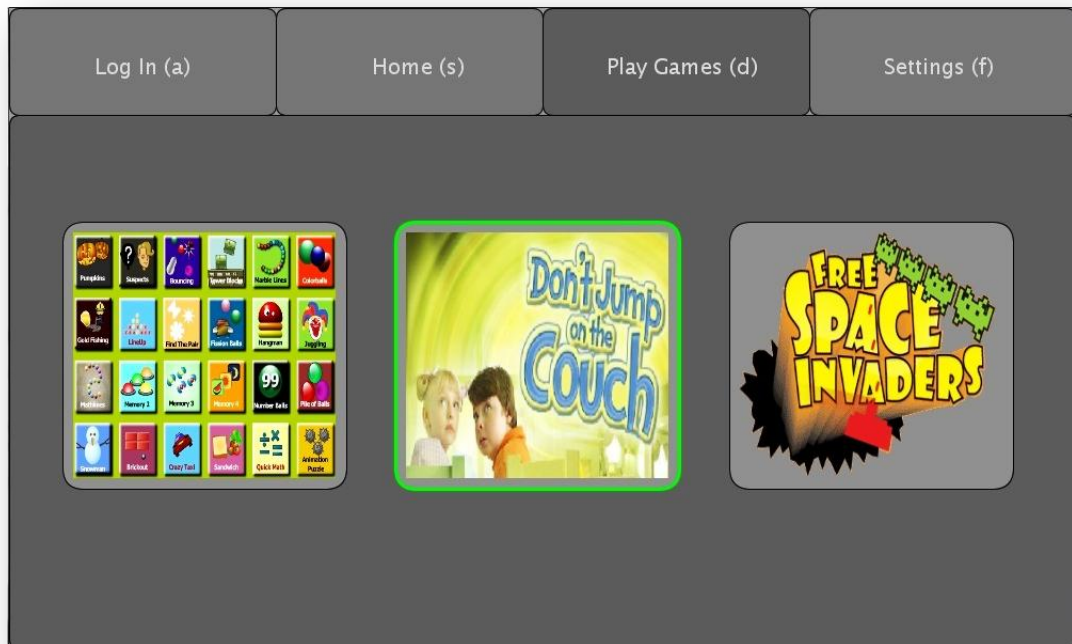
Game menu

Once the controller check is completed, the patient is taken to the game menu (Figure 11) where they have access to a variety of puzzle and arcade games. They are free to play the games at their leisure. As the patient is playing, the GUI monitors the switches in the armrest to detect improper posture. If at least three of the five switches are released a voice message is played telling the patients to place their forearm back on the armrest. After the patient has finished playing, they simply exit the GUI. Upon exiting the GUI, a window is displayed containing a good-bye message and updated game-play chart depicting the patient's progress for the day (Figure 8).

Figure 12. Settings menu. (a) Allows the therapist to enable/disable Skype conversations. (b) Allows the therapist to modify the Wekinator pairing model. (c) The therapist can run the model to test it out.
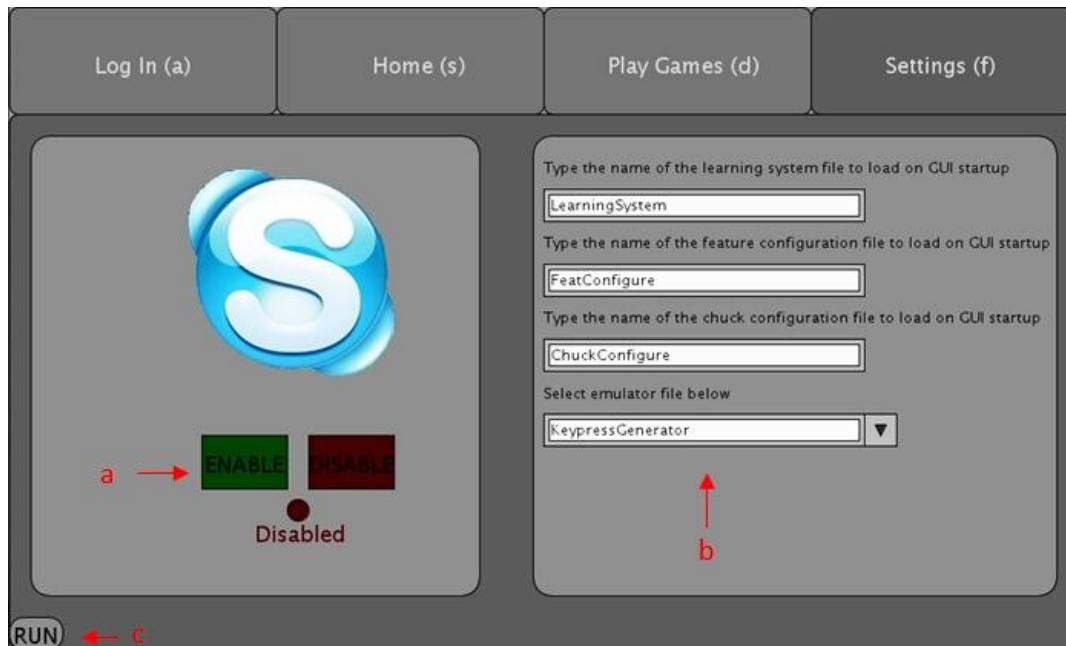
Other functions for meeting the needs of research and science

In addition to the guidance that it provides to the patient, the application records important data for future analysis. One set of data recorded consists of the acceleration and angular velocity output from the controller, their corresponding control signal, followed by a timestamp. This data is recorded continuously throughout the game-play at a rate of 100 Hz. Concurrently, the digital states of the armrest switches are recorded with timestamps at a rate of 100 Hz. Furthermore, the number of minutes the patient spends playing during each exercise trial is recorded when the patient exits the GUI. Apart from the quantitative data, the application also takes and saves a photo of the patient through the computer's webcam upon login for patient identity verification and records video of the patient during game-play for posture analysis by the therapist.

Potential User Requirement

Although the system can be potentially adapted for other limbs and joints, the setup and functions described here specifically target motions involving the wrist. The system includes video games with varied complexities. However most games can be easily comprehended by most children from the age of 7 years old and older with comparable cognitive ability. This study is geared towards young patients who are in need of wrist or arm rehabilitation. The users must also be able to at least follow basic instructions such as "move your hand to the right".

User validation of the system

NExT was developed to promote massed practice in a controlled and reliable manner. To validate the system, we studied the practice patterns and collected user feedback from children with and without hand impairments.

To study practice patterns using NExT, we recorded muscle activity and joint motions while a subject was playing the video games. By doing so, we can understand whether a patient (1) plays games by the prescribed joint motions/muscles using our system and (2) repeats the prescribed motions as frequently as their typically developing peers do. We recorded activities of flexor carpi radialis (FCR) and extensor carpi ulnaris (ECR) by using electrodes of a commercial surface electromyography (EMG) system (Bagnoli Desktop EMG system, Delsys Inc., USA) with a gain of 1000. The EMG signals were sampled at 1000 Hz and sent to a laptop through an analog-to-digital converter (USB-6343, National Instruments Corp. Austin, TX). In addition, two motion sensors (MTw, Xsens, The Netherlands) were placed across the patients' wrists to capture kinetic data of wrist motions and this data was sent to a 2nd laptop running the Xsens commercial software. A trigger pulse was sent to the laptop on which the EMG was recorded in order to synchronize the two sets of data. The two sets of data were combined and visualized through a LabVIEW VI (National Instruments Corp. Austin, TX), and off-line data processing was conducted in Matlab (MathWorks Inc. Natick, MA).

We recruited seven children with hand impairments, who were diagnosed with hemiplegic cerebral palsy (three girls, four boys; mean age 10y 6mo, SD 1y 4mo) and five typically developing children (four girls, one boy; mean age 8y 9mo, SD 1y 4mo) to use the prototype of our developed system. The children with hand impairments were recruited from the Children's Rehabilitation Center of Hasbro Children's Hospital in Rhode Island, USA. The typically developing children were recruited from the siblings of patients and children of staff in the Pediatric Rehabilitation Center.

Inclusion criteria for the patient group include: (1) Manual Ability Classification System (MACS) levels I-II, ability to handle most kinds of objects independently using one or both hands with or without compromised performance quality, (2) able to actively participate and follow instructions. The exclusion criteria: (1) other neuromusculoskeletal problems, such as hand fracture or peripheral nerve injury, (2) wrist and forearm contracture. The typically developing children were free of any neurological and musculoskeletal impairment while participating in this validation study.

All the participants were asked to play six video games for a total duration of 18 minutes. After they played the video games using our system, we inquired the children two questions: (1) Do you like to play video games using our system? (2) Will you practice every day using our system if we put a device at your home? The children answered our question using a 7 point Likert scale. Seven indicates they like it very much or will practice every day and one indicates that they do not like the game at all or they will not use our system at all.

The signals from the controller generated by the participants were captured by our system. The signals were then classified as different joint motions, such as wrist motion, forearm supination, and forearm pronation (Figure 13.) over time.

Figure 13. The amount of movement made by the targeted joints (a) Typically developing child (b) child with CP
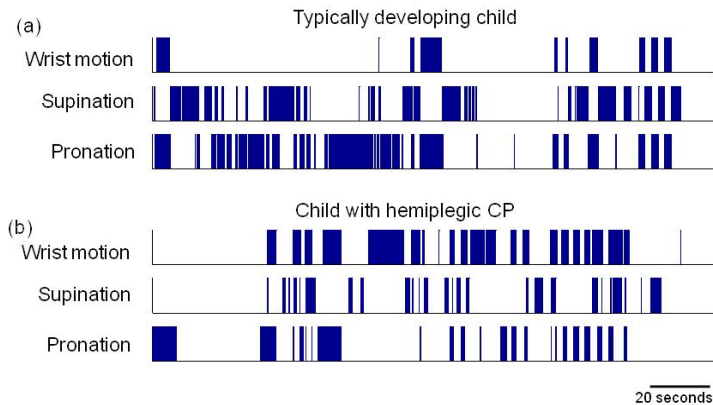

Figure 13 shows an example of the type of data obtained from joint motions measured over a period of 200 seconds of game-play. This example illustrates that a child with brain injury can use this system for repetitive wrist movements which is key to reliable massed practice, although there might be differences from normal usage patterns.

Figure 14 shows representative electromyographic activity and its corresponding joint motions during game-play. The recorded EMG signals were first fed to a notch filter to eliminate 60-Hz line noise and then to a band-pass (10-400 Hz) filter to remove the motion artifacts and high frequency noise. The filtered EMG data were processed in a linear envelope representation (ratification before low pass filtering at 20 Hz).

Figure 14. Muscle activation during video game-play: our training system could successfully elicit the target muscle activation during game-play.
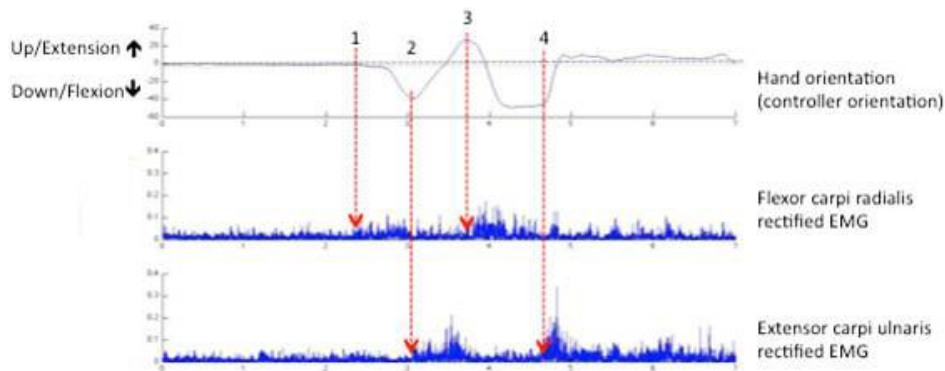
We demonstrated that the system can elicit the desired motions by recording EMG during game-play. With initiation of flexion or extension (number 1, 2, 3 and 4 in the first trace of Figure 14), we detected clear corresponding EMG activation of FCR and ECU. This trial confirms the activation of desired joint motions.

The feedback from children with brain injury or typically developing children shows no difference. Both groups could accept our system as an adaptive way to play video games. Only one child with brain injury responded that he would not use it every day but 1-2 times per week, which lowers the Likert scale score in the study shown in Figure 15.

Figure 15. Children's response to our system and willing to practice at home.
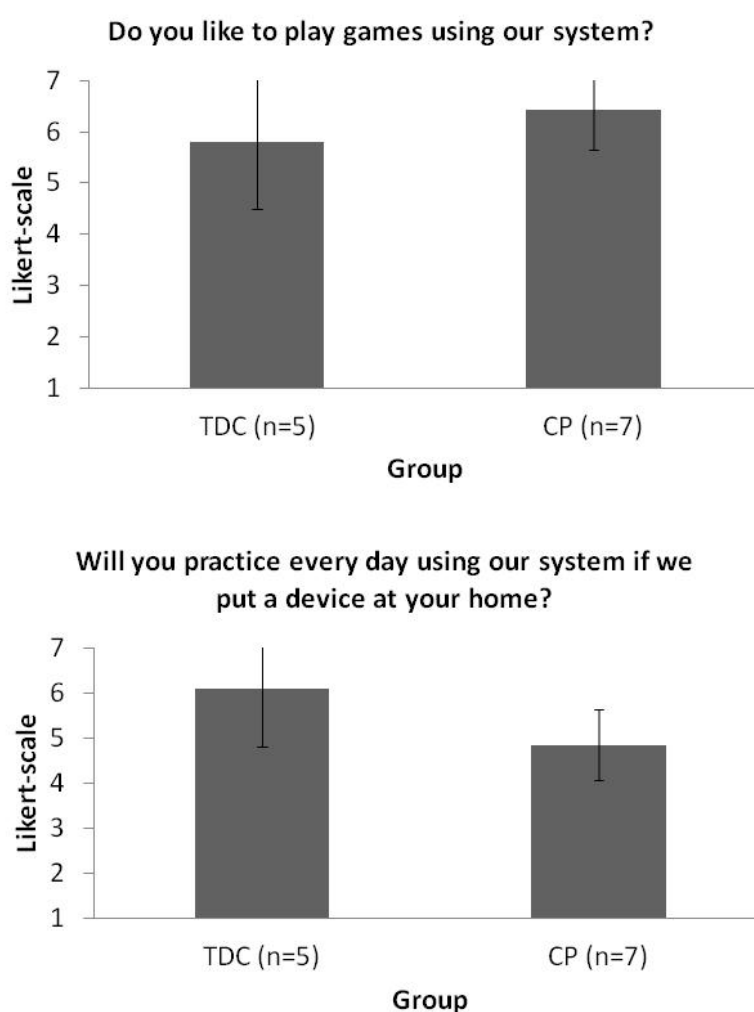
Discussion and Conclusion

Our system addressed the needs of efficient neurorehabilitation while providing a complementary tool for upper extremity training in children with cerebral palsy. Our approach was intended to allow children with brain injury to receive rehabilitation therapy more frequently at convenient times and locations. A concern with a hand-held sensor to play games, such as Nintendo's Wii® Sport, is that the desired game motion could be accomplished by moving a combination of many other joints rather than the target joint, in our case, the wrist. Although commercial active video games (Nintendo's Wii® Sport and Wii® Fit) have been studied on balance, reaching or physical fitness with controversial results(Brichetto, Spallarossa, de Carvalho, & Battaglia, 2013; Deutsch et al., 2008; Graf, Pratt, Hester, & Short, 2009; Hurkmans et al., 2010; Ramstrand & Lygnegard, 2012), these approaches may not be ideal for motor training which targets specific joint motion. Microsoft Kinect™ has potential to increase motor training; however, Kinect™ currently cannot register wrist/forearm rotations with sufficient precision. We are incorporating simple switch methods with the capability of using an additional camera to ensure that the posture of other body parts is maintained and unwanted compensatory, maladaptive movements are limited. The system will help to ensure compliance with therapeutic regimens prescribed by clinicians, while promoting massed practice. In future studies, data collected in the home setting will be used to develop a scientific basis for therapy evaluation, including how the frequency and pattern of activation influence motor improvements. The overall goal of the NExT system is to enhance motor performance in children with brain injury while allowing scientists and clinicians to better understand optimal training and therapy approaches.

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A WIRELESS USER-COMPUTER INTERFACE TO EXPLORE VARIOUS SOURCES OF BIOSIGNALS AND VISUAL BIOFEEDBACK FOR SEVERE MOTOR IMPAIRMENT

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Abstract: Severe speech and motor impairments caused by several neurological disorders can limit communication skills to simple yes/no replies. Variability among patients' physical and social conditions justifies the need of providing multiple sources of signals to access to Augmentative and Alternative Communication (AAC) systems. Our study presents the development of a new user-computer interface that can be controlled by the detection of various sources of biosignals. Wireless sensors are placed on the body and users learn to enhance the control of detected signals by visual biofeedback, on a switch based control approach. Experimental results in four patients with just few residual movements showed that different sensors can be placed in different body locations and detect novel communication channels, according to each person's physiological and social condition. Especially in progressive conditions, this system can be used by therapists to anticipate progression and assess new channels for communication.

Keywords: Human-computer interaction, Assistive Technologies, Augmentative and Alternative Communication, Complex Communication Needs, biosignals control, Amyotrophic Lateral Sclerosis, Locked-in Syndrome.

Introduction

Several neurological conditions, either static or progressive, can cause generalized loss of motor control and/or speech (e.g. brainstem stroke, amyotrophic lateral sclerosis, traumatic brain injuries or spinal cord lesions) (Glennen & DeCoste, D., 1996; Beukelman, Yorkston & Reichle, 2000). As modern medical care extends survival of people with marked motor disability (Laureys, Pellas & Eeckhout, 2005; Katz, Haig, Clark, & DiPaola, 1992), the impossibility to communicate has a large impact on their quality of life (Blain-Moraes S, Schaff R, Gruis KL, Huggins JE, Wren PA, 2012; Beukelman, Fager, & Nordness, 2011). Assistive technologies (AT) play an important role for enhancing or providing new communication channels to express their needs and desires, as well as to allow a more intense social contact largely beyond the very simple yes/no response. Even the most severely impaired patients can benefit from the modern Augmentative and Alterative Communication (AAC) facilities, in order to access text-to-speech and Internet tools, thus extending their communication possibilities to receive information and to participate in social networks (Light & Gulens, 2000; Nijboer, Birbaumer & Kübler, 2010; Smith & Delargy, 2005).

User interfaces to access to AAC devices are of utmost importance. Considering severe motor impairments, finding the proper sensors that fit to the user's specific physical conditions and that enable the user to efficiently generate control signals, is sometimes a difficult task. Biosignals have been explored in various ways as solutions for persons with severe motor impairments to access AAC devices and different applications. Pinheiro et al (2011) present a review on how electromyographic (EMG), electrooculographic (EOG) and electroencephalografic (EEG) signals have been extensively studied for access to AAC systems. In our study, we considered some characteristics of user interfaces based on biosignals that make these difficult to be used by users who have severe motor impairments:

1) Setup is complex and learning takes a long training process. Considering patients with severe neurological involvement, complexity may limit user's motivation. Moreover, the use of AAC devices is closely dependent on caregivers support, which is hard to achieve if the AT is difficult to setup and learn (Ball, Beukelman & Bardach, 2007).

2) No flexibility to use different sources of biosignals using the same AT system. If there is more than one choice, a proper clinical assessment can be made to study user interfaces that maximize the flow of information with the minimal physical and cognitive workload for the user (Abascal, 2008). Especially for progressive conditions, the AT should dynamically adapt to physical, physiological and psychological stages of the patients, along the course of the disease (Londral, Azevedo & Encarnação, 2009; Beukelman et al., 2000).

3) Many of the experimental results are obtained from non-disabled participants. Experimental studies with the involvement of the appropriate population are important. They can reveal usability factors that may be determinant for optimal design and effectiveness when applying these technologies (Clarke, Langley, Judge, Hawley, Hosking & Heron, 2011).

Aiming at avoiding the difficulties considered above, we present a new wireless control interface based on body sensors and underlying biosignals. We target patients with very few residual movements as a consequence of severe neurological conditions, either progressive or residual.

Firstly we briefly describe the proposed user computer interface, both in terms of hardware and software characteristics. We then describe the preliminary results from four patients with severe motor impairment: two women with ALS (late stage); and two men with long-standing incomplete locked-in syndrome (iLiS) (Smith & Delargy, 2005), due to brainstem stroke. We finish with a short discussion of our results.

Methodology

Design Requirements

The aim of our design was to develop a computer interface that can support the use of various sources of biosignals for accessing AAC devices.

Kintsch and DePaula (2001) have enumerated four important aspects to be considered in AT development: 1) must be customizable; 2) should be simple enough to set-up, customize and use; 3) be durable and robust 4) accommodate user's preferences, namely adapt to the users' environment and social context. Considering these aspects, we developed a new AT with the following design requirements:

a) Support different sources of biosignals, to accommodate users' characteristics;

b) Simple to setup and use in the daily environment of the user;

c) Wireless and adaptable to different body movements, reducing positioning constraints and follow progressive conditions;

d) Easy to learn (considering users and caregivers) and with minimal setup overhead.

System Description

General overview

The proposed platform was developed to enhance communication, through simple body-triggered activations. The activation (voluntarily controlled by the user) is detected using sensors placed on the body, which collect the underlying biosignals, and transmit them via Bluetooth® to the computer where they are processed in real time to detect the control activation signals (i.e user voluntary body-triggered activation). When any activation is detected, the software emulates a keystroke (e.g., to control a virtual keyboard using a scanning method) or a switch input command to an AAC software (e.g. ©TheGrid2, from Sensory Software Int). Figure 1 illustrates the block diagram of the proposed work.

Figure 1. Block diagram of the proposed system: (1) Data acquisition - user activates the sensor; (2) Signal processing - captured signal is sent via Bluetooth to the computer and processed to generate control signals; (3) Switch-based control - when an event is detected, the system sends a command to an assistive communication software (e.g a virtual keyboard)

Data Acquisition

For body signal acquisition we used a commercially available system (bioPLUX™) with 4 analog channels. This system can collect biosignals from different types of sensors and sends these signals via Bluetooth® wireless transmission to a computer (base station). Its wireless transmission range of up to 100m is appropriate for the purpose of a user computer interface.This system was setup to use a sampling rate of 1000Hz and a 12-bit resolution per channel.

Figure 2. bioPLUX system used for body signal acquisition. Sensors are connected to the system and biosignals are sent to a computer via Bluetooth.
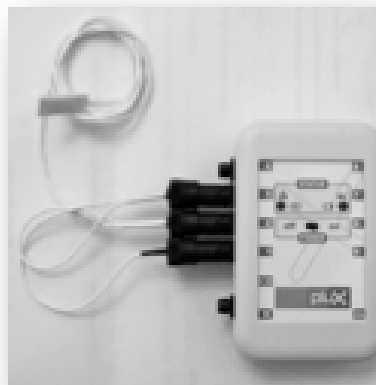
Different sources of biosignals

In addition to wireless communication, we applied miniaturized sensors to provide comfort and flexibility. Our platform allows the application of different sensors, working as a customized solution for each user. We focused on three different body sensors, namely: a surface electromyography (sEMG) sensor (gain 1000, CMRR 110dB, 25-500Hz passing band filter, and input impedance >100MOhm), an accelerometer (ACC) (3-axial MEMS device with ±3G measurement range), and a force sensor (FSR) sensor (force sensitive resistor with 0-10Kg range and response time <5µS). Figure 3 depicts the set of sensors evaluated in the proposed system.

Figure 3. Set of sensors evaluated in our work; from left to right: electromyography sensor, accelerometry sensor, and pressure sensor.

Signal Processing

The main result for signal processing, in the proposed system, is the detection in real time, of events within the control signal. We define control signal as the processed signal that the user of the interface will voluntarily control to generate command events. After collecting the biosignal (raw data), our system processes the control signal through an algorithm to detect commands that result from the user's intention to set an activation.

Calibration. Before the user starts to control the system, there is a simple calibration process where the user is asked to stay for 5 seconds at rest position. The power of the noise signal (1) is extracted from this "signal at rest", with a calculation of the mean value for the 5 seconds (5000 samplesin our case, due to the sampling rate of the system).

$$\bar{x} = \frac{1}{5000} \times \sum_{i=1}^{5000} x_i \qquad (1)$$

Variance Algorithm. When a user makes a voluntary activation, the detected signal (movement, muscle contraction, or force) shows a variation in amplitude that is associated with that activation. Considering the case of sEMG signal, increased activation is correlated to greater signal amplitude. As such, the variance of an EMG signal contains important information about the voluntary activation. In our algorithm, the maximum-likelihood estimate of the local variance is computed for the windowed signal parts, in real time (Bonato, D'Alessio & Knaflitz, 1998). The maximum-likelihood estimate of the variance, which is a biased estimate, is defined as:

$$\hat{\sigma}_x^2 = \frac{1}{n}\left[\sum_{i=1}^{n} x_i^2 - \frac{1}{n}\left(\sum_{i=1}^{n} x_i\right)^2\right] \quad (2)$$

where x_i is the magnitude of the signal in sample i and n is the number of samples defined for a data window. This function (2) is analogous to a moving average window, except for a square term, which increases the difference between voluntary activation and *no activations* (Choi & Kim, 2007). The onset of a voluntary activation is detected as the first point, which, in the variance signal, surpasses a pre-defined threshold (th) for at least an interval of 100ms (we used 100ms to ignore sporadic activations).

$$th = power(rest\_signal) + N \times std(rest\_signal) \quad (3)$$

th (3) was defined as the power of the noise signal (1) plus the standard deviation error of the noise signal multiplied by a scale factor N which depends on the type of signal used.

Although a variance analysis is particularly effective to detect voluntary activations for sEMG signals, the variance analysis can also be generalized to other signals that include an activation zone. This algorithm was then used in our system for all types of studied signals (ACC, sEMG and FSR), as illustrated in Figure 4 for detection of slight movements using an accelerometer.

Biofeedback Software

The developed software platform collects data streamed in real time by the bioPLUX system through the Bluetooth® port, and shows it in the computer screen. Users can then visualize both the body signal and the processed control signal in real time, and learn how to control them using biofeedback strategies (Figures 4 and 5).

Figure 4. Visual biofeedback window developed for the presented study. Both body signal and control signal (from the variance algorithm) are visually presented to the user. User learns to control the body signal by watching it on the screen. Horizontal green line.


Customizable features

The software platform includes a customization panel. Customization is an important factor to accommodate variability among users, particularly different tasks to perform. In this panel, the user can choose which type of body action (corresponding to a specific sensor) will be performed for control, and which third-party application should receive the events generated from control signal. This is particularly important for Switch based Control (SBC) of AAC (e.g. scanning method). As an example, Figure 5 shows our system controlling an onscreen keyboard for a writing task. The variable

th (3) is also customizable, by manually changing the height of a threshold line on the screen (shown in Figure 4).

Figure 5. Example of the developed platform, controlling an onscreen keyboard to perform a writing task in a ©WordPad (from Microsoft) document. In this example, when detecting a control signal from EMG generated by the user, the key "Enter" is sent by the platform to the application of onscreen keyboard. This command performs a selection using the scanning method.
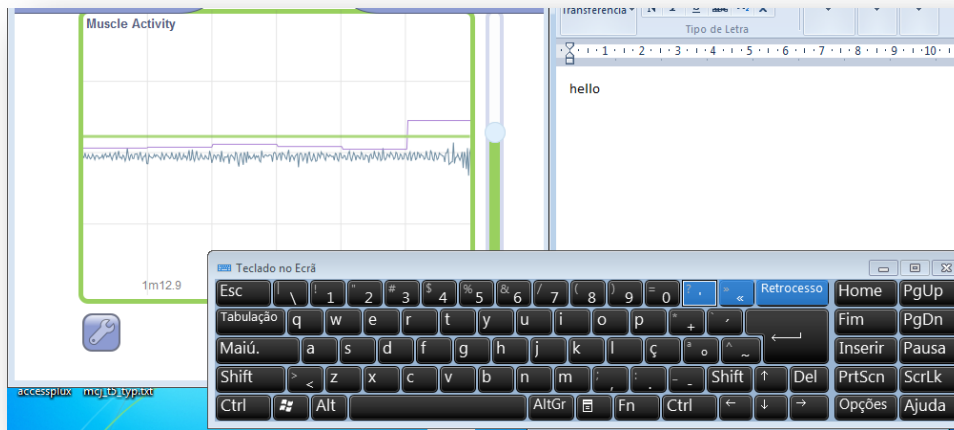
▪ Exploratory Study

With the objective of qualitatively evaluating: (1) the signal processing algorithm to detect control signals; and (2) usability issues related to sensors placement and environment adequacy, we performed an exploratory study including four participants.

Participants

The proposed system was tested in four patients with severe motor and speech impairments. All patients were between 40 and 65 years old. The selection criteria was the presence of just a few residual volitional movements, and marked difficulty to find a user interface that could fit both physical limitation and social context (considering acceptance and technical support abilities of the caregivers). Table 1 summarizes the clinical and social context of each participant.

Table 1. Selected patients included. For each participant, this table describes the place where they live, clinical condition, residual movements that were used for this study, speech preservation and the sensors tested

Participant	Residence	Clinical Condition	Residual Movement	Speech	Sensors
P1	Elderly residence	ALS	Left hand fingers and muscle contractions in the arm	Yes	ACC, FSR, sEMG
P2	Home	ALS	Right hand (closed) and head	No	ACC, FSR
P3	Long term Care clinic	Partial locked-in syndrome	Forehead muscle contractions	No	sEMG
P4	Palliative hospital service	Partial locked-in syndrome	Chin movements	No	ACC, sEMG

Procedure

Experiments were performed in a single session per participant, in their usual daily environment (see column *Residence* in Table 1), to evaluate the adequacy of the proposed system to the different environments. The purpose and procedures of the study were explained, to obtain informed consent. Furthermore, participants and caregivers were asked to give their opinion during the assessment period. They were asked to show their residual movements, and sensors were chosen according to the physical characteristics of those movements (see column *Residual Movement* in Table 1). Figure 6 illustrates a setup of an accelerometer sensor to detect slight movements of the index finger, in one of the participants of this exploratory study.

Figure 6. Detection of slight movements of index finger from the left hand in a patient with ALS.

The sensors used to assess residual movements were: accelerometer (ACC), electromyography (sEMG) and force (FSR), as described in the previous section. A computer screen was used to provide visual biofeedback of the biosignal (both raw signal from the sensor and processed control signal) to the participant, in real-time. For each setup, participants tried to execute and observe its corresponding response by visualizing biosignals on the biofeedback window (computer screen). After approximately 2 minutes watching the sensor signal and practicing simple cause-effect activities, participants were asked to fulfill two tasks, namely: T1) generate 5 to 10 onsets of the signal; and T2) generate an onset and hold it for 5 seconds. For the accelerometer sensor, task T2 was not considered. Just participant P3 had previous training sessions with a therapist, to learn how to control sEMG. Biosignals detected by the sensors during the experiments were saved for further analysis.

Outcomes were qualitative variables defined as: sources (body signals) with which users could fulfill the proposed tasks, types of sensors that the user could use to perform onsets and generate control signals and main difficulties observed in fulfilling the proposed tasks.

Results

All participants, except P4, were able to fulfill the first task in one body signal, at least. Table 1 shows which sensors were used to provide the control signals for each participant. P1 and P2 were able to use more than one sensor to generate control signals. Table 2 describes all the performed tasks and the characteristics for each measured control signal, showing the number of impulses detected by our system during the execution of task T1 and impulse lengths in both tasks. Figure 7 shows the biosignals corresponding to task T1.

Table 2. Characteristics of measured control signals for tasks T1 and T2

Participant	Sensor	Body placement	Task 1 Number of performed onsets / Detected activations	Task 1 Duration of control signal activations (ms)	Task 2 Onset duration (ms)
P1	FSR	Right thumb pressure	5 / 5	μ=839 ∂=152.97 Max=1049 Min=599	No control
P1	ACC	Right thumb movement	5 / 5	μ=1060.8 ∂=118.1 Max=1203 Min=856	n.a.
P2	ACC	Left index finger	10 / 10	μ=419 ∂=112.25 Max=599 Min=299	n.a.
P2	FSR	Left index finger	10 / 10	μ=404 ∂=96.05 Max=599 Min=299	5025.21
P2	sEMG	Left arm Biceps	10 / 10	μ=464 ∂=80.78 Max=599 Min=299	No control
P3	sEMG	Forehead Frontalis	6 / 6	μ=2734.6 ∂=3186.62 Max=10499 Min=998	No control
P4	ACC	Chin (inferior jaw)	5 / 0	not detected	n.a.

Figure 7. Biosignals for task T1 performed by the four participants. Titles of each plot indicate the participant, type of sensor and part of the body that actuates the sensor, as described in Table 2.
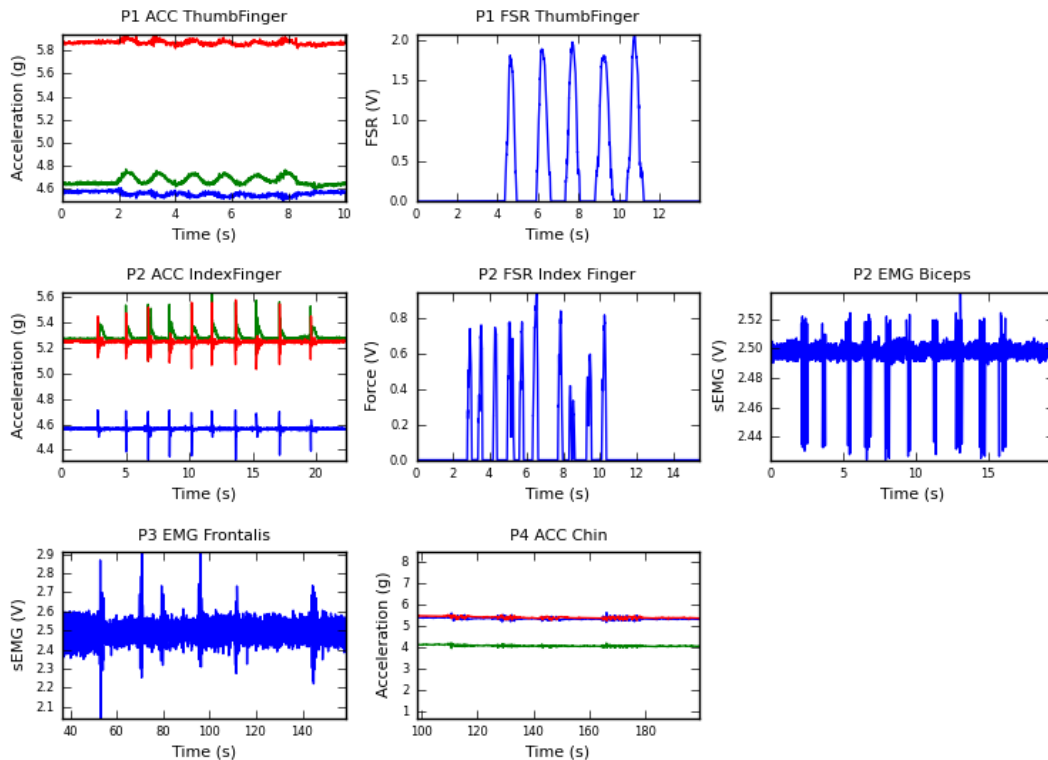
Discussion

In the described exploratory study, participants were able to perform voluntary onsets of one or more body signals, which were tested as sources of control signals. Users learned to control the movement to generate voluntary activations, using the biofeedback window. All patients were able to rapidly understand how to generate activation signals. The main difficulties in exploring different biosignals to perform the proposed tasks were the large reaction times and short awareness periods of some of the users. For participant P4, it was specially difficult to find a period of approximately 20 minutes, in which the test could be setup. Onsets were performed voluntarily by this participant (using accelerometry from chin movements), during experimental tests, though our variance algorithm could not detect them as activation signals due to their low amplitude.

One of the main positive aspects of our proposed system was the flexibility to adapt to each user context and position. Due to the wireless characteristics and use of different sensors, none of participants of the presented exploratory study had to change their environment or position to perform the proposed tasks. Moreover, we could observe that visual biofeedback is a very important tool for training control over residual movements. In our tests, this tool was used, both by the users, to learn to control the biosignal, and by the caregivers or therapists who gave feedback to the users in the learning and motivation process, in our tests.

Results from our exploratory study with four participants contribute to the implementation of design requirements defined for the development of the proposed computer interface, based on biosignals detection. In spite of the small number and difficult physical conditions of our target population, results from experimental tests with these users are important to support further developments.

Further experiments using the proposed system to perform communication tasks by access to an AAC software must be implemented. Tests will be performed on a broader range of users, exploring new algorithms for automatic activation signals detection from biosignals.

Conclusion

Patients with severe motor and speech impairments need AT to support communication. Due to patients' difficult physical conditions and strong dependence on caregivers support, ATs should be simple to setup, learn and use. We presented the development of a wireless user interface, based on the detection of biosignals and scanning access. Our system was developed to allow the use of different sensors and to detect various residual movements. Wireless connectivity and the use of sensors that are placed on the body were considered to reduce positioning constraints and open novel communication channels for those who are severely impaired. We presented an exploratory study that included four patients with severe motor impairment, in their daily care context. We evaluated biosignals from three

different sensors (ACC, FSR and sEMG) located in different body parts. From a qualitative analysis, we could observe that our interface is easy to setup and learn, and is flexible to robustly transduce residual movements from multiple sources into control signals. Biofeedback was observed as an important feature of this designed platform: participants could explore residual movements, visualize them in real time on the computer screen and learn how to control them. Particularly for progressive neuromuscular degenerative conditions, our system can be useful in the clinical assessment, to follow disease progression and search for alternative communication channels.

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INSTRUMENTATION AND BIOMECHANICAL MODEL FOR KINEMATIC AND KINETIC ANALYSIS OF UPPER LIMBS DURING GAIT WITH CRUTCHES

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Abstract: The goal of this study was to develop a three-dimensional kinematic and kinetic model of the right upper extremity and a Lofstrand crutch in order to analyze joint displacements and loads during crutch-assisted gait. A Lofstrand crutch was instrumented with a six-component load cell to measure forces and moments at the crutch tip. The crutch and the right upper extremity of a subject were instrumented with markers to obtain kinematic data. A biomechanical model based on rigid bodies was implemented in biomechanical analysis software. To demonstrate the functionality of the model, a pilot test was conducted on one healthy individual during Lofstrand crutch-assisted gait. The shoulder extended during the support phase and flexed in the swing phase, the elbow flexed during the swing, and the wrist remained in extension throughout the cycle. In the shoulder and elbow joints, the predominant reaction forces were upward, whereas the internal force moments were flexion and extension, respectively. This tool will be useful when it comes to identifying risk factors for joint pathology associated with pattern gait, aid design or crutch overuse.

Keywords: biomechanics; biomechanical model; upper extremity; crutch-assisted gait.

Introduction

Restoring the capacity to walk in patients with incomplete spinal cord injury (SCI) is one of the main goals in rehabilitation programs. In the United States, it is estimated that there are currently 236,000 to 327,000 persons living with the consequences of SCI.[1] Furthermore, the estimated annual incidence of 40 per million population means that every year there are an additional 8,000-11,000 persons with SCI:[1] indeed, the percentage of the population with incomplete SCI increased from 44% in 1973 to 54% in 1990.[2,3] Nevertheless, with the use of proper bracing and walking aids, many such individuals become functional walkers.[4]

Shoulder pain is one of the most usual complaints among persons with SCI, due to the increased mechanical demands placed on the upper extremities for the purpose of achieving different types of displacements. Musculoskeletal conditions caused by excessive use of the upper extremities -rotator cuff overload in particular- are the most common cause of shoulder pain among subjects with spinal cord injury. In tasks involving upper-extremity weight bearing, such as wheelchair propulsion or transfers, the application of repetitive forces to the shoulder joint can lead to rotator cuff impingements or injuries[5]. During these weight-bearing actions, a force is transmitted via the forearm to the glenohumeral joint, elevating the head of the humerus. Failure to control this movement can lead to impingement of the tendons of the shoulder rotator cuff or other structures found in the subacromial space. In the case of manual wheelchair propulsion, these demands have been widely studied, in terms of both incidence and biomechanical characteristics. For instance, Sie et al.[6] and Dalyan et al.[7] estimated that: 59% of patients with spinal cord injury complained of some type of pain in the upper extremities; 30% presented with important pain which required medication, or experienced limitation or pain in the performance of two or more activities of daily living; 66% had symptoms related with carpal tunnel syndrome; 36% reported shoulder pain; 16% reported elbow pain; 13% reported wrist pain; and 11% reported hand pain.

In the case of persons with incomplete SCI who can walk, a high incidence of shoulder pain has also been reported, with prevalence figures of 39% at ages

31-45 years and 61% at ages over 45 years.[8] As in the case of manual wheelchair propulsion or transfers, weight bearing on the upper extremities during walking with aids has been implicated as a causative factor in developing shoulder pain.[6] The use of canes, crutches or wheelchairs entails related risks, such as degenerative arthritis of the hand and wrist, carpal tunnel syndrome, or ulnar neuropathy at the wrist. These repetitive actions exerting a load on the palm of the hand cause radial deviation and extension of the wrist, pressure on the carpal tunnel, and prolonged or repetitive contractions of the muscles of the hand and wrist. A number of studies have shown that actions associated with the use of crutches can cause an increase in carpal-tunnel pressure which could, in turn, lead to ischemia of the median nerve.[9] It is important to identify the factors that predispose persons to such injuries. Shoulder joint loads during assisted walking are not as well documented as those experienced in wheelchair propulsion. Biomechanical gait analysis with crutches yields pertinent information. Hence, there is a case for conducting biomechanical gait analysis of the upper extremities of persons who require crutches, canes or walkers to move about, and thereby ascertain their kinematic and dynamic gait patterns, i.e., the movements, forces and moments undergone by their upper limb joints. A comprehensive biomechanical model is required to identify potential mechanical sources of shoulder injury.[4] An inverse relationship has been reported between assistive device axial load and lower extremity strength.[10] Efforts to relate assistive device loads to demands placed on the upper limbs during ambulation have been limited. Previous endeavors have gone some way to examine upper extremity dynamics during Lofstrand crutch-assisted gait.[4,11,12] Requejo et al.[4] presented a system involving sensors positioned around the crutch handle. A more recent experiment was performed with a four-sensor crutch system for the purpose of directly measuring crutch-cuff kinetics and fully quantifying the dynamics of the wrist, in addition to those of the elbow and shoulder.[11]

The goal of this study was to develop a 3D kinematic and kinetic model of the right upper extremity when walking with a Lofstrand crutch, using a six-component strain-gage load cell located at the crutch tip to measure forces and moments of force, and a system of active markers to capture movement.

Unlike previous experiments, which envisage one rigid body below the hand-handle interface, this model divides each part of the crutch into rigid bodies.[4,11] To verify the application of this biomechanical model, the kinematic and kinetic patterns registered by a single subject during Lofstrand crutch-assisted gait are described and discussed below.

Materials and methods

Instrumentation

For study purposes, a Guardian-model crutch (Sunrise Medical, Fresno, CA, USA) was instrumented with a six-degree-of-freedom MCW-6-1000 load cell (AMTI, Watertown, MA, USA) at the distal end. This sensor incorporates strain gages capable of measuring the components of the forces and moments received by the sensor (Fx, Fy, Fz, Mx, My, Mz), with respect to its effective origin and coordinate system. The manufacturer calibrated the sensor in-house, indicating the calibration constants of each variable, along with the location of the effective origin and orientation of the axes of the coordinate system. The signals generated by the dynamometer sensor were transmitted via a 10m-long cable to the pertinent amplifier, and after being amplified, to the motion-capture device, where they were recorded in synchronization with the positions of the markers.

Load-cell accuracy and precision were determined by simultaneously recording force data from the forceplate (Kistler, Winterthur, Switzerland) and crutch during crutch-assisted gait, with the crutch tip contacting the forceplate. Root-mean-square (RMS) error (i.e., the difference between the forceplate and the resultant three-component crutch force) was used to calculate sensor accuracy, and standard deviation was computed to assess precision.[4,13]

Tracking marker trajectories were recorded by two scanners, placed at either side of the walkway and fitted to an active-marker motion-capture system (Charnwood Dynamics Limited, Leicestershire, England).

Kinematic model

To implement the model of the right upper extremity, segments of the trunk, arm, forearm and wrist were defined.[4] Active markers were placed on different bony landmarks in order to compute segment movements and joint-center positions during the readings (Figure 1).

The center of the shoulder joint was deemed to be at the glenohumeral joint center (GHJC). This joint's position was calculated by rotating the plane formed by the markers of the greater tubercule of the right humerus (RTB), medial epicondyle (LE) and medial epicondyle of the humerus (ME), through an angle of 30º in an anticlockwise direction about the line formed by the RTB and LE markers. A straight line was created belonging to the rotated plane and parallel to the line formed by the LE and ME markers, which passed through the RTB marker. The GHJC was located on this latter line at a distance from the RTB equal to the mean distance between the LE and ME.

The elbow joint center (EJC) was located at the midpoint between the LE and ME markers; the wrist joint center (WRJC) was located at the midpoint between the markers of the ulnar styloid process (USP) and radial styloid process (RSP); and the third metacarpal joint center (3JC) was located at the point at which the third carpometacarpal (M3) marker projected onto the plane formed by the RSP, USP and lateral fifth metacarpal (LM5) markers.

In line with International Society of Biomechanics (ISB) guidelines:[14] the X axis was defined as anteroposterior, with the anterior direction being deemed positive; the Y axis was defined as longitudinal, with the upward direction being deemed positive; and the Z axis was defined as mediolateral, with the lateral direction being deemed positive.

The Yt axis of the trunk segment was parallel to the line formed by the midpoint between the right iliac crest (RIC) and left iliac crest (LIC) markers, and the midpoint between the RTB marker and the marker of the greater tubercule of the left humerus (LTB); the Zt axis was perpendicular to the Yt axis, pointing toward the RTB marker; and the Xt axis was perpendicular to the Yt and Zt axes described above.

The Ya axis of the arm segment passed through the EJC and GHJC, the Za axis was perpendicular to the Ya axis, pointing toward the LE marker, and the Xa axis was perpendicular to the Ya and Za axes. The Yf axis of the forearm passed through the WRJC and EJC, the Zf axis was perpendicular to the Yf axis, pointing toward the RSP marker, and the Xf axis was perpendicular to the Yh and Zh axes of the forearm. The Yh axis of the hand passed through the 3JC and WRJC, the Zh axis was perpendicular to the Yh axis, pointing toward the LM5 marker, and the Xh axis was perpendicular to the Yh and Zh axes of the hand.

As a general rule, rotations about the Z, Y and X axes were assumed to be flexoextension, abduction/adduction, and internal/external rotation movements, respectively. To describe the angular movements between coordinate systems, Cardan displacements were calculated following the ZXY sequence; and to describe joint movement, the angular displacement of the distal segment coordinate system was shown relative to that of the proximal segment coordinate system. In view of the shoulder's anatomical complexity, in which a number of joints are involved, its joint movement was regarded as the displacement of the arm coordinate system with respect to the trunk coordinate system.

Three active markers were used to obtain crutch kinematics: two were placed on the anterior surface of the crutch shaft, distally (DC) and proximally just below the handle (PC), and a third was extended on a wand, 25 mm from the anterior end of the handle (HC). Four segments were considered, i.e., handle, shaft, load cell, and crutch tip (Figure 2). To define each of these segments, some landmarks, calculated on the basis of the position of the markers and the dimensions of the crutch, had to be identified. These landmarks corresponded to the ends of each of the above segments. The Y axis of each segment of the crutch was assumed to coincide with its longitudinal axis; the X axis was perpendicular to the Y axis, pointing toward the DC marker; and the Z axis of each segment was perpendicular to the Y and X axes described above.

Figure 1. Right upper extremity marker set and segment coordinate systems:[4] RTB, right greater tubercule; LTB, left greater tubercule; RIC, right iliac crest; LIC, left iliac crest; LE, lateral epicondyle; ME, medial epicondyle; RSP, radial styloid process; USP, ulnar styloid process; M3, third metacarpal; LM5, fifth metacarpal; HC, crutch handle marker; PC, proximal crutch marker; DC, distal crutch marker.
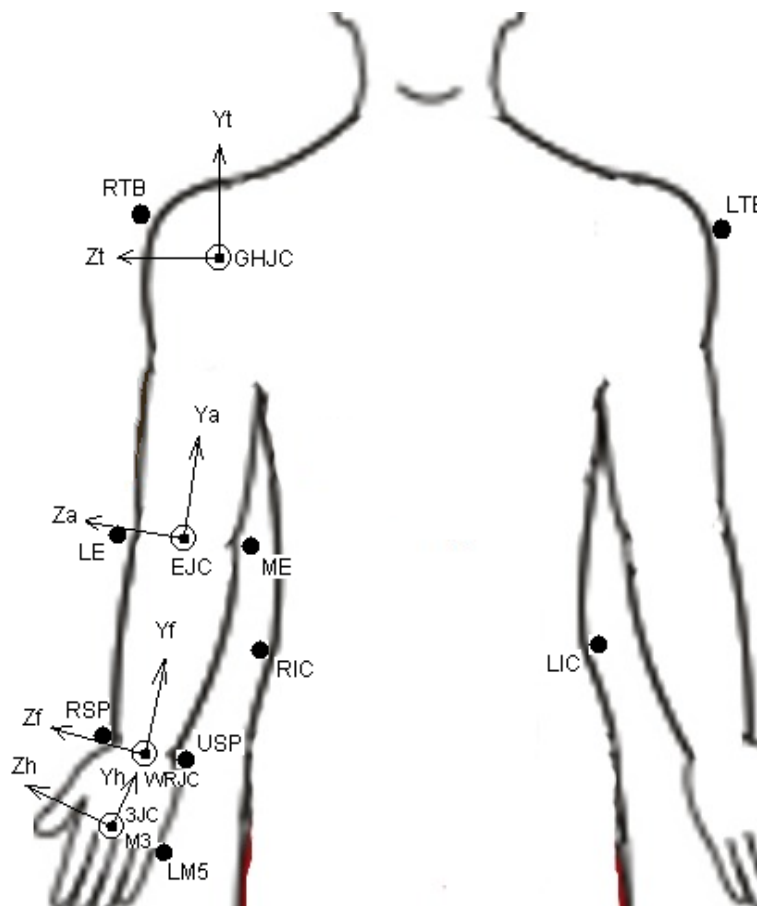
Kinetic model

Each segment was assumed to be a rigid body with its mass uniformly distributed,[4] with the trunk segment being deemed to be an elliptical cylinder, the arm and forearm segments being deemed to be cylinders, the hand segment being deemed to be a sphere, and each segment of the crutch being deemed to be a cylinder. Inverse dynamic Newton-Euler methodology (Figure 3) was used to calculate the joint-kinetic values between two consecutive rigid bodies, for both the upper extremity and crutch.[15] The joint-reaction forces, i.e., the forces exerted by the lower segment on the upper segment of any given joint, were calculated and referenced to the upper segment coordinate system. The internal moments of each joint were

calculated with respect to the proximal segment coordinate system, following the right-hand rule (Figure 1). The kinetic data of the wrist were not considered valid, due to a lack of data on load values between forearm and crutch cuff.

The point of application of the forces of the hand on the crutch handle was assumed to be at the projection of the 3JC on the line joining the HC to the proximal end of the shaft. The forces and the moments measured by the load cell were deemed to be applied on the load-cell segment, at the sensor's effective origin, and with respect to its coordinate system.

The biomechanical model was implemented in biomechanical analysis software (C-Motion, Inc., Germantown, MB, USA) in order to calculate kinematic and kinetic joint data on the basis of the tracking marker trajectories, and data on the forces and moments measured by the load cell.

Figure 2. Crutch model. The crutch model was divided into handle, shaft, load cell, and tip segments, defined on the basis of the handle marker (HC), proximal marker (PC), and distal marker (DC)
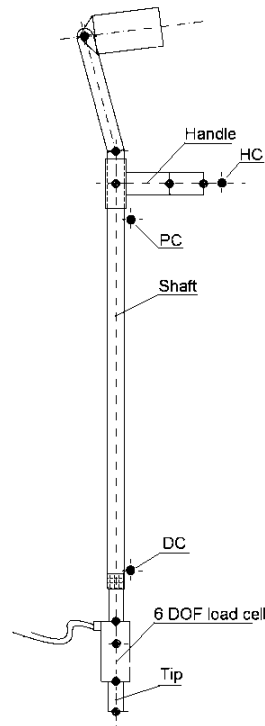
Figure 3. Proximal joint force and moment of any segment in the Global Coordinate System. mi = mass of segment i, ai = acceleration of segment i, n = number of distal segments connected in chain, q = number of external forces, Fj = applied external forces, p = number of external couples, τk applied external couples, Pj = vector from the application of the external force to the proximal joint, Ri = distance from the center of mass of each distal segment i to the proximal joint, Mi = inertial moment due to segment i.

$$\vec{F}_{proximal} = \sum_{i=1}^{n} m_i(\vec{a}_i + \vec{g}) + \sum_{i=1}^{q} \vec{F}_j$$

$$\vec{M}_{proximal} = \sum_{i=1}^{n} \left\{ \vec{M}_i + \vec{R}_i \times \left[m_i(\vec{a}_i + \vec{g}) \right] \right\} + \sum_{j=1}^{q} \left(\vec{P}_j \times \vec{F}_j \right) + \sum_{k=1}^{p} \vec{\tau}_k$$

Pilot test

To verify the application of this biomechanical model, a pilot study was conducted with the aid of a healthy subject. A healthy female, age 24 years, weight 68kg and height 1.66m was instrumented. The height of the instrumented crutch was adjusted so that the handle was at the level of the ulnar styloid apophysis, with the subject standing upright, and her arms hanging by her side and flexed 15º.[16,17] In order to perform the pilot test, the subject was required to walk at free speed along an 8 meters-long walkway, holding the crutch with the right hand and loading it simultaneously with left lower limb support. The subject was required to lean the trunk slightly to the right side when loading the crutch. The subject performed 10 trials to become familiar with the instrumentation. Then, she was asked to walk along the walkway 8 more trials. Five gait cycles were selected to be analyzed based on a good signal acquisition and a correct movement performance by the subject[12,13,18] .Informed consent was obtained from the subject as stipulated following project approval from the Clinical Research Ethics Committee and in accordance with the Declaration of the World Medical Association.

Data-collection, -processing, and -analysis

Kinematic data were recorded at a frequency of 200Hz. and load cell signals at a frequency of 1000 Hz. The kinematic signals were interpolated using a polynomial least squares fitting procedure, and a second-order bidirectional Butterworth filter and cut-off frequency of 6 Hz was then applied.[12,19] In the case of the force and moment signals, a second-order, bidirectional, low-pass Butterworth filter and cut-off frequency of 7 Hz was applied.[20]

To obtain the results, the subject completed the walkway 5 times. From each test run, a cycle corresponding to that performed in the center of the walkway was selected. The cycle so selected was referred to as the "crutch cycle", which was in turn defined as all the data recorded between two consecutive initial contacts between crutch tip and ground. The data reported here thus refer to 5 crutch cycles. In each cycle, the instants at which the crutch left the ground were also defined, in order to differentiate the crutch's support phase from its swing phase. The initial contacts and lifting of the shaft tip were manually detected by visualization of the vertical force recorded by the sensor.

Once the cycles targeted by this study had been selected: the kinematic and kinetic joint data were normalized with respect to 100% of the crutch-cycle duration; the mean and standard deviation of the normalized variables of the 5 cycles were calculated for each moment of the cycle; and the spatial-temporal, kinematic and kinetic variables were obtained.

The spatial-temporal variables were calculated on the basis of the data on the position and time point of the landmark located at the distal end of the crutch-tip segment. The following parameter definitions were used: Cycle Length, the anteroposterior distance between two consecutive initial contacts between crutch tip and ground; Cycle Time, the time elapsed between the two initial contacts defined above; Stance Time, the time during which there is contact between crutch and ground; Swing Time, the time during which the crutch is not in contact with the ground; Speed, Cycle Length divided by Cycle Time; Cycles/minute, the number of cycles performed per minute calculated on the basis of Cycle Time; and % Stance

Phase, the percentage duration of the cycle during which there is contact between crutch and ground. In the case of the kinematic variables, the joint displacements of the shoulder, elbow and wrist were analyzed; and in the case of the kinetic variables, the forces and moments of force of the shoulder and elbow were obtained.

Results

Accuracy

The accuracy test yielded an RMS error of 2.14N, which represented 0.67% with respect to the maximum value of the resultant crutch force obtained (317.60N). In the precision test, the standard deviation of the error was 1.40N (Figure 4).

Figure 4. Resultant crutch (solid line) and forceplate (dashed line) forces simultaneously recorded.

Spatial-temporal parameters

The mean velocity reached for the 5 cycles recorded was 0.35 (+/-0.03) m/s. The mean length of the right crutch cycle was 0.95 (+/-0.05) m. The remaining spatial-temporal parameters are shown in Table 1 below.

Table 1. Spatial-temporal crutch cycle data.

Parameter	Mean	Std. Dev.
Cycle Length (m)	0.95	0.05
Cycle Time (s)	2.7	0.2
Stance Time (s)	1.8	0.2
Swing Time (s)	0.9	0.0
Speed (m/s)	0.35	0.03
Cycles/minute	22.0	1.3
Stance Phase (%)	67.0	2.8

Kinematics of the upper extremity

Insofar as the shoulder was concerned, hardly any mobility was observed in the joint during the first half of the support phase (Figure 5). Once the loading response phase had passed, the shoulder extended. Then it flexed during the swing phase. In the case of the trunk, the humerus remained in abduction and internal rotation throughout the cycle.

The elbow remained flexed throughout the cycle, attaining a peak value in the swing phase (Figure 5). There was little variation in pronation across the cycle.

The wrist remained in extension throughout the cycle (Figure 4). During the support phase, it experienced slight consecutive flexion and extension movements, with flexion becoming more rapid and displaying a greater range of movement at the end of the phase. The wrist was observed to remain in ulnar deviation across the cycle.
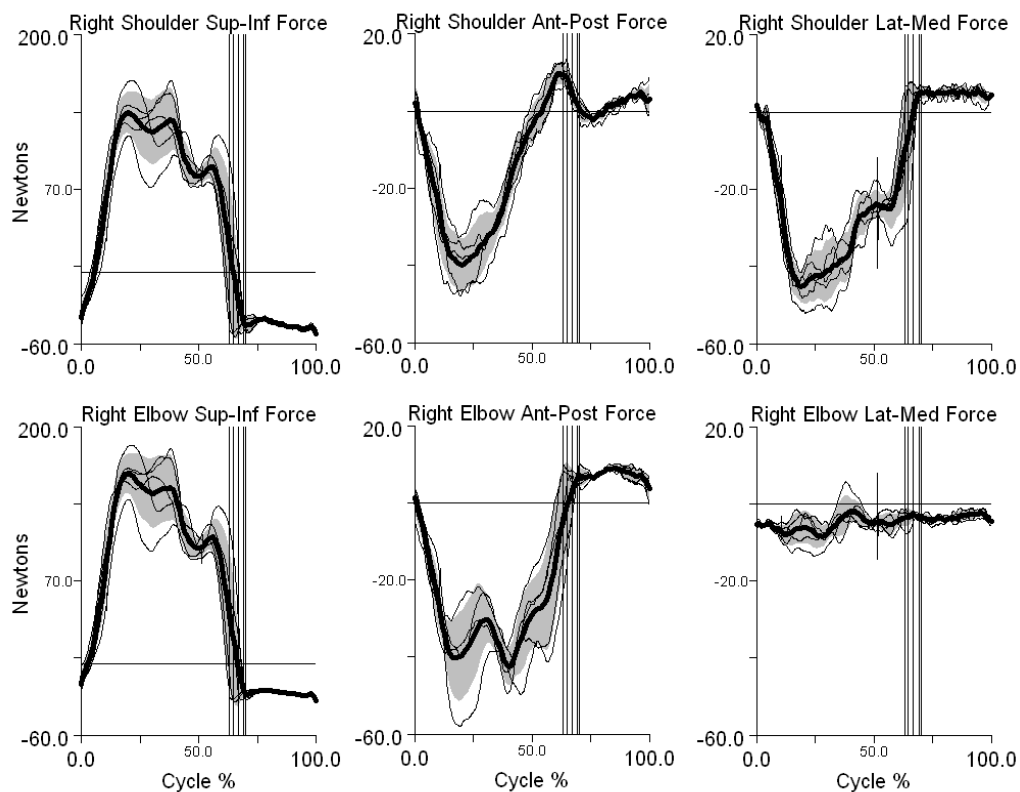
Figure 5. Graphs depicting joint displacement: thin lines represent the 5 recorded graphs, solid lines the mean, and grey bands the mean +/- standard deviation of the 5 readings
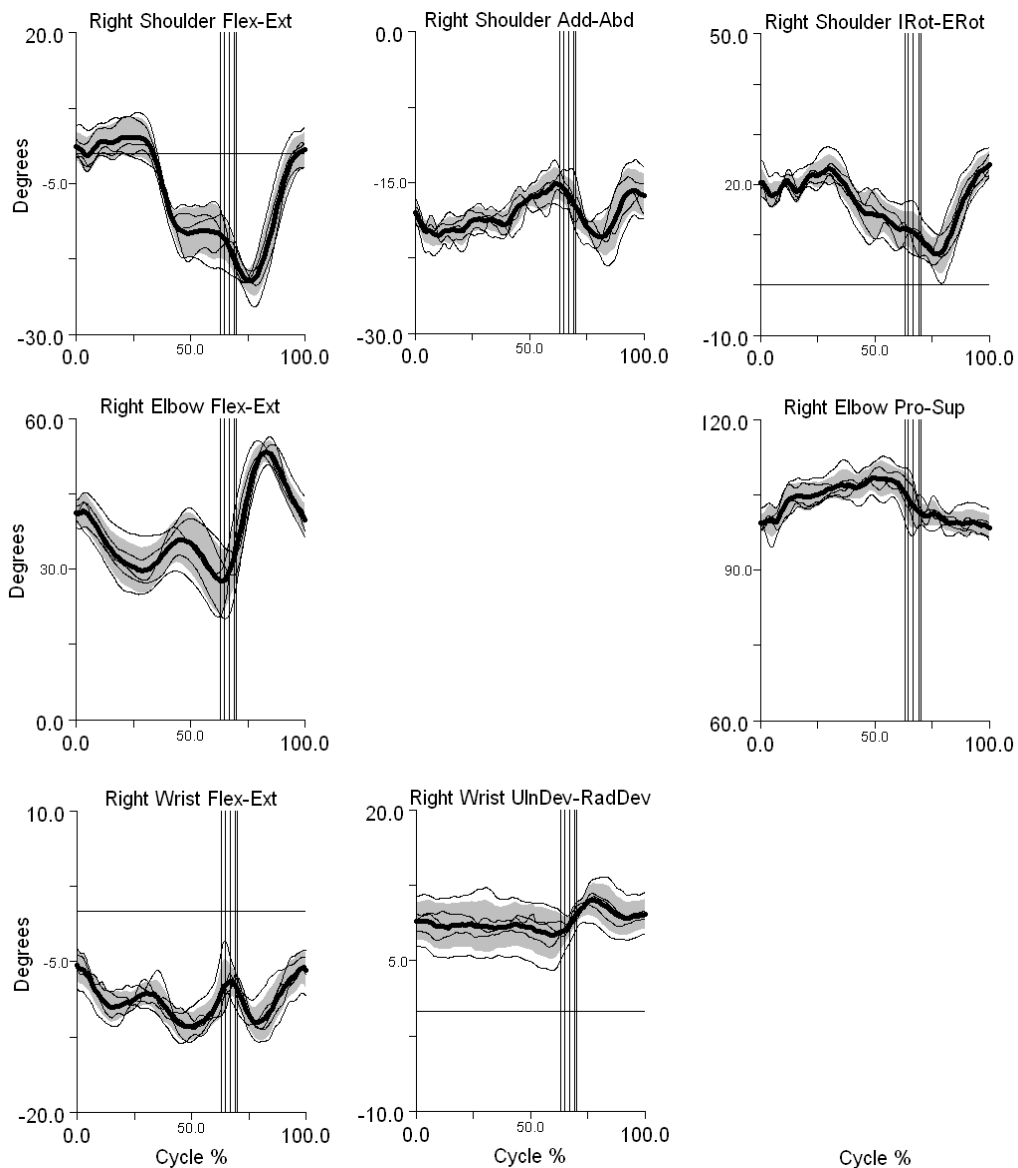
Joint kinetics in the upper extremity

Within the shoulder joint complex, the reaction force in the glenohumeral joint was observed to be mainly vertically upward, posterior and medial in the support phase (Figure 6). During this same phase, there was an internal flexor moment in this joint (Figure 7), which continued to be flexor in the swing phase, albeit with the minimum value required to advance the crutch in order for its tip to make the next point of contact with the ground.

As with the shoulder, the reaction force in the elbow joint was essentially vertically upward and posterior to the joint (Figure 6). As was to be expected, whereas the internal moment in the elbow was extensor during the support phase, it was mainly flexor in the swing phase (Figure 7).

Figure 6. Graphs depicting joint forces: thin lines represent the 5 recorded graphs, thick lines the mean, and grey bands the mean +/- standard deviation of the 5 readings.
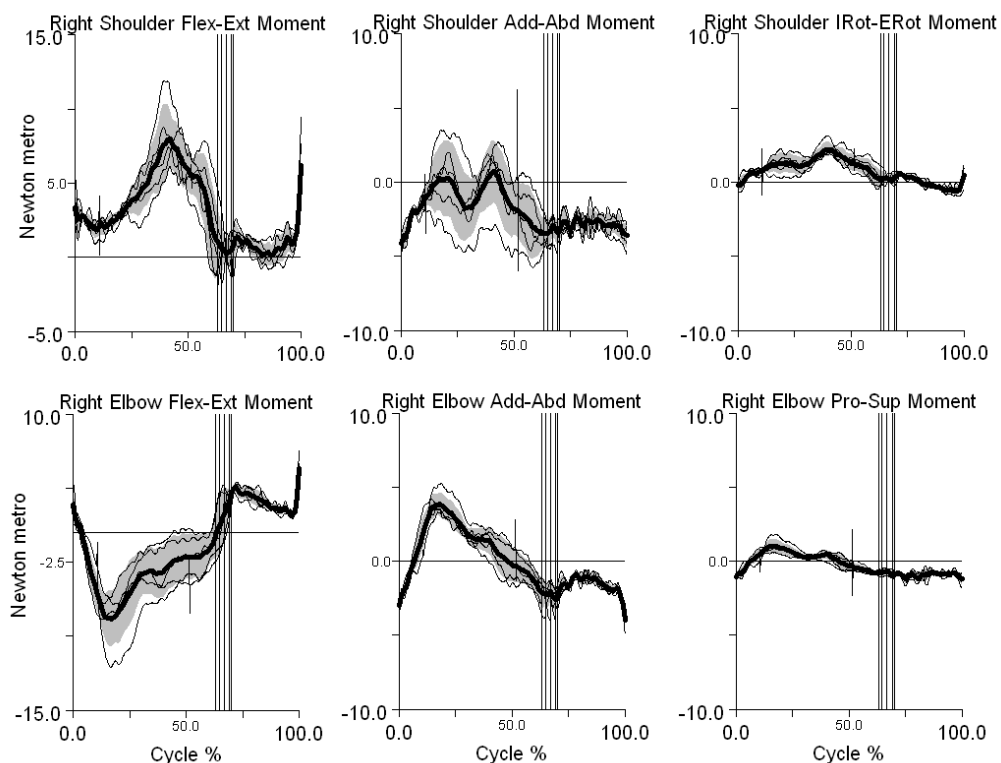
Figure 7. Graphs depicting joint moments: thin lines represent the 5 recorded graphs, solid lines the mean, and grey bands the mean +/- standard deviation of the 5 readings.

Discussion

The paper describes a biomechanical model, as well as the instrumentation and methodology, for the study of upper extremity joint kinematics and kinetics during crutch-assisted gait. Compared to previous works,[4,11,12,22,25] here we have presented a three-dimensional kinematic and kinetic model that incorporates a full description of forces application points and the rigid body geometry, and meets ISB joint coordinate system guidelines[15]. Studies following the kinetic model presented in this work, and that adhere the ISB standards will be reproducible and comparable. The functioning of the model was demonstrated by conducting a pilot test with a healthy subject.

The inclusion of a force sensor placed at the tip of the crutch enabled accurate estimation of the major net joint forces and moments of the upper extremities. The data obtained by comparing the resultant crutch and

forceplate forces confirmed the accuracy of the crutch instrumentation, as previously reported.[4,13]

For the purposes of this study an active-marker motion capture system was used, whereas other authors have used systems based on passive markers.[11,12,21,22] While the placement of the crutch markers, namely, one at the anterior end of the handle and two on the shaft, coincided with that used in previous experiments,[4,12] other earlier studies used two markers placed on the anterior and posterior positions of the sensor, which was itself located at the distal end of the crutch, and a further two markers on the shaft, one in the middle and another at the height of the handle.[18,23] In this study, the force sensor was situated on the distal part of the crutch, in line with previous models.[18,23,24] Other authors have included a load cell just below the handle to reduce inertial loading effects.[4,11,12] The kinetic model proposed in this study assumes the crutch to be an element formed by 3 rigid bodies, i.e., handle, shaft and force sensor, whereas others regard it as a single rigid body lying between the handle and the center of the sensor.[4,11,18]

This study reports the results registered in respect of a healthy subject who performed a three-point gait pattern, using the crutch with the upper right extremity. Other authors have studied subjects with different medical conditions, using reciprocal four-point or swing-through gait patterns, with two crutches, a walker or even a cane.[4,11,12,18,20-22,25]

With regard to joint kinematics, after the moment corresponding to the loading response phase, the shoulder extends to enable the trunk to advance vis-à-vis the crutch. Thereafter, the shoulder flexes in the swing to move the crutch tip forward, so as to ensure that the following initial contact is made at approximately the height of the foot of the contralateral extremity.

During approximately 40% of the cycle, the elbow has to flex to avoid raising the body's center of gravity when the tip of the shaft passes at the height of the body. The elbow then extends to ensure that the crutch maintains contact with the ground and continues transferring the weight of the upper part of the body to the ground until the crutch is lifted. In the swing phase, the elbow must flex to raise the shaft tip and prevent it from coming into contact with the ground, and then extends once again to prepare for the

following initial contact. As regards the adduction-abduction movement of the elbow, this is a passive rotation and is not reported.[15] The elbow's peak flexion in the swing phase proved of greater intensity in this than in a previous study.[4] Its value depends on the anthropometric characteristics of the user, cycle duration and crutch length.

Overall, the kinetic data obtained in this study were of the same order of magnitude as those reported by other similar studies.[4,11,18,22,23] During the support phase, the vertical reaction forces in the shoulder and elbow are similar in form but the magnitude of the forces observed in the elbow is superior to that of the shoulder, a finding in line with that of other authors.[4,11] During the swing phase, in contrast, this situation becomes inverted, with the vertically downward force being greater in the shoulder than in the elbow.

This effect could be due to the fact that, in the support phase, the weight of the lower extremity is transmitted to the ground, and the shoulder has to bear part of the trunk's weight which is loaded onto the crutch, whilst the elbow during the support phase has to bear the loading of the trunk plus the weight of the arm. In the swing phase, in contrast, the shoulder has to control the full weight of the upper extremity plus the crutch, whereas the elbow has to control the same weight, less that of the arm. As observed in previous studies, during the support phase there is a predominance of forces exerted in a superior, posterior and lateral direction in the shoulder.[4,12]

It is noteworthy that, for around 40% of the cycle, in the second half of the support phase there is an increase in the value of the vertically upward force and flexor moment in the shoulder while this joint rotates toward extension. This situation might entail an increase in the load borne by the shoulder and a moment of risk of joint overuse.

The predominance of the flexor moment observed in the shoulder during the support phase coincides with that found by other authors who have examined reciprocal gait.[11,18,25] Whereas studies of swing-through gait showed a predominance of internal adductor moments in the shoulder during

the support phase,[25] in reciprocal gait internal abductor moment values were detected in the shoulder during the support phase.[4] This difference between adductor and abductor moments in the shoulder in swing-through and reciprocal gait was subsequently confirmed.[18] In this study, which examined reciprocal gait with a crutch, the internal moments in the anteroposterior axis of the shoulder joint were in line with those described above, tending to values of small magnitude but with a trend toward abduction.

As reported by previous studies,[4,12] the elbow shows a predominance of superior and posterior forces during the support phase. In the loading response phase, there is a peak force directed vertically upward in the elbow, together with a posterior peak force and an increase in the extensor moment at this level which, as in the case of the shoulder, coincides with a displacement toward extension. This circumstance may also amount to the joint being compromised. This study observed an extensor moment of the elbow during the support phase, a finding in line with other studies that have likewise addressed reciprocal gait.[4,11,18]

Some limitations must be mentioned, however. First, there was no information about the load existing between the crutch cuff and the forearm, with the result that no kinetic wrist data were reported. Similarly, the study was performed on a single upper extremity. Accordingly, the model should be extended to include both of the upper limbs and two crutches, and the instrumentation should be modified to ascertain the kinetics of the carpus. Once these conditions have been met, the respective gait patterns of subjects with different medical conditions could then be analyzed

Conclusions

The procedure described in this work provides the basis for the kinematic and kinetic analysis of the upper extremity joints during crutch-assisted gait. The present model is described in detail, including all force application points and rigid body geometries, and meets ISB guidelines, warranting precise comparison with future analyses. The results of a pilot test demonstrates the functionality of the trial configuration and its potential

application for the clinical practice. The results of the present wil be used to initiate a study of different gait patterns using crutches (reciprocal versus swing trough) in order to evaluate which offer a higher risk to develop overuse upper limb joint pathology.

Acknowledgments

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