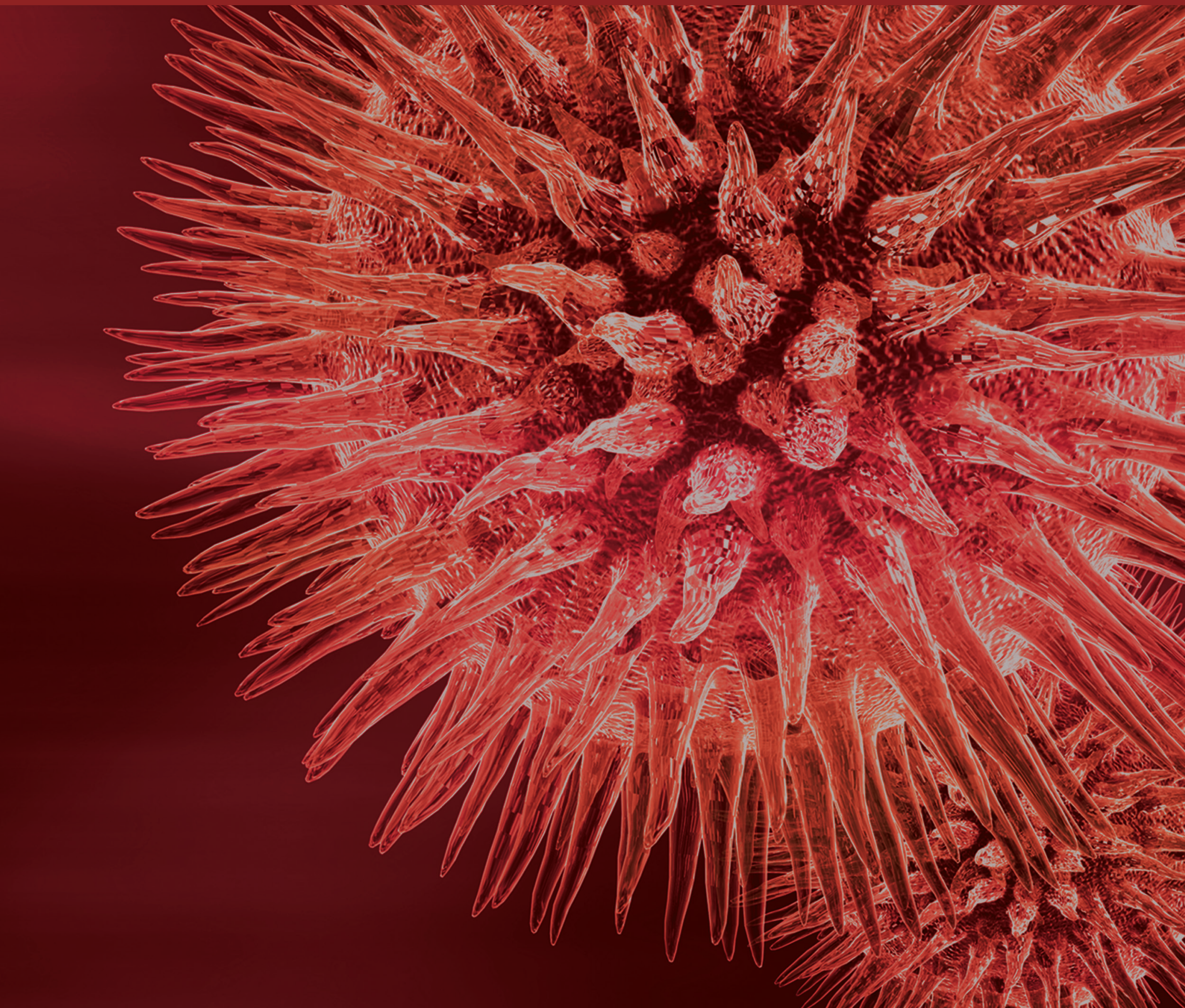
BioMed Research International

Rehabilitation and Improvement of the Postural Function

Guest Editors: Thierry Paillard, Massimiliano Pau, Frédéric Noé, and Luis-Millán González



Rehabilitation and Improvement of the Postural Function

BioMed Research International

Rehabilitation and Improvement of the Postural Function

Guest Editors: Thierry Paillard, Massimiliano Pau, Frédéric Noé, and Luis-Millán González

Contents

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 703679, 2 pages
http://dx.doi.org/10.1155/2015/703679

Editorial

Rehabilitation and Improvement of the Postural Function

Thierry Paillard,[1] Massimiliano Pau,[2] Frédéric Noé,[1] and Luis-Millán González[3]

[1]*Physical Activity, Performance and Health Laboratory, University of Pau and Pays de l'Adour, Department STAPS, ZA Bastillac Sud, 65 000 Tarbes, France*
[2]*Biomechanics and Industrial Ergonomics Laboratory, Department of Mechanical, Chemical and Materials Engineering (DIMCM), University of Cagliari, Piazza d'Armi, 09123 Cagliari, Italy*
[3]*Department of Physical Education and Sports, University of Valencia, C/Gascó Oliag 3, 46010 Valencia, Spain*

Correspondence should be addressed to Thierry Paillard; thierry.paillard@univ-pau.fr

Received 25 October 2015; Accepted 25 October 2015

Posture refers to the position of different body segments at a given time which can be modified through joint mobilization and the action of the neuromuscular system. Maintaining balance during bipedal quiet stance requires complex mechanisms from the postural control system in order to keep the vertical projection of the centre of mass (COM) within the base of support [1]. To achieve this aim, the centre of pressure (COP) plays a crucial role to compensate for any deviations of the COM, which can generate imbalance if they move beyond the limits of the base of support. The ability to control the COM depends on internal body representation in space. Internal representation is acquired by means of a learning process but also depends upon genetic factors [2]. This representation is elaborated by sensory inputs and is based on kinematic (segmental organization, whole body acceleration, and body orientation relative to earth gravity) and kinetic (joint torques and forces efforts between the plantar cutaneous surface and the ground) parameters [3]. Moreover, a postural attitude is never acquired definitively even in quiet stance. The body constantly undergoes changes caused by liquid movements and cardiac and respiratory muscular contractions. This phenomenon modifies its at-rest state and prevents it from maintaining a strict balance [4]. It is characterized by continuous body sway and results from an internal perturbation. In addition, muscle tone constantly varies which both accentuates body sway and complicates the possibility of cushioning it [5]. Postural control is thus a permanent process of balance regulation whose implementation is based on subtle mechanisms.

Postural regulation is organized in hierarchical and stereotypic patterns and requires the central integration of afferent inputs from the sensory systems as well as the motor command of antigravity muscles. The proprioceptive (myotendinous and joint sensors), exteroceptive (mainly visual and cutaneous plantar sensors, but also auditory sensors), and vestibular (vestibular sensors) inputs are integrated by the vestibular nuclei located in the brain stem and are controlled by the cerebral cortex and cerebellum [6–10]. The activation of postural muscles is organized in synergies (activation/inhibition of agonists/antagonists muscles) and is based on postural neural networks [3].

Each sensory, central, and motor component of the postural function is either healthy or pathological and will display normal or abnormal functions. In pathological subjects, the dysfunction of certain organs involved in postural control is likely to amplify body sway and/or to affect the ability to cushion it and it may also alter the segmental organization of postural control. Different evaluation methods enable the exploration of each component through protocols of motor perturbation (mechanical disturbance), sensory stimulation (sensory manipulation), and/or cognitive disturbance (e.g., virtual simulation, dual task). Postural behavior of healthy subjects can be characterized in terms of postural performance (i.e., the ability to minimize body sway) and segmental (i.e., the multijoint coordination) and neural strategies (i.e., the preferential involvement of short or long neuronal loops, i.e., myotatic or visuovestibular). A particular postural behavior can be easily considered as normal or abnormal through measures of magnitude, velocity, and acceleration of linear or angular displacements of the COM, COP, and body segments and also through measures of electromyographic activities and evaluations of the contribution of different sensory information. All these measures contribute to describe precisely the compensatory and anticipatory postural adjustments characterizing postural behavior. Compensatory postural adjustments act in a feedback manner to preserve balance in response to the actual balance disturbances whereas anticipatory postural adjustments precede the onset of a postural disturbance while minimizing its feedforward effects.

Neurological and muscular pathologies, sensitivity deficits (e.g., vestibular, visual), cerebellar syndrome, and many other diseases severely degrade postural control. The postural performance and strategy of healthy subjects notably differ from those of pathological subjects (e.g., [3, 11]). For a given pathology, the postural behaviour evolves in a specific way [12]. However, many scientific considerations in discovering testing and rehabilitation for each pathology of postural function still remain. When subjects present such pathologies, as mentioned above, they are liable to fall, which can have dramatic consequences for their physical integrity. The development of stimulation techniques of the sensory and motor functions in a rehabilitation context is likely to improve and help restore postural function while the refinement of testing techniques improves descriptions of dysfunction of the postural function. For these reasons, this special issue provides supplemental knowledge related to the evaluation and rehabilitation of the postural function in pathological subjects (from children to aged patients) that could advance their therapeutic management.

Moreover, among healthy subjects, the postural function can positively and negatively evolve according to age (e.g., development in children, involution in aged subjects) and the status of subjects in terms of physical activity (e.g., active or inactive). For all these populations, postural control can also be positively influenced by repeated regularly exercise or training. Exercise optimises the sensory, central, and motor outputs of the postural function and can induce motor program acquisitions which include specific postural adaptations [13, 14]. Indeed, in a working, leisure, or sporting context, highly skilled subjects are subjected to having a performant postural control since there is a close relationship between postural and motor skill (or postural and motor performance), specific training developing specific postural skills [15]. Postural strategy can also be modified by the effects of training [16]. The progress in the advancement of scientific knowledge in healthy subjects can help in the understanding of pathological postural mechanisms. Thus, this special issue also integrates work dealing with the effects of domestic and leisure physical activities and sport on the postural function in healthy subjects.

Thierry Paillard
Massimiliano Pau
Frédéric Noé
Luis-Millán González

References

[1] D. A. Winter, "Human balance and posture control during standing and walking," *Gait and Posture*, vol. 3, no. 4, pp. 193–214, 1995.

[2] C. Assaiante, F. Barlaam, F. Cignetti, and M. Vaugoyeau, "Body schema building during childhood and adolescence: a neurosensory approach," *Neurophysiologie Clinique*, vol. 44, no. 1, pp. 3–12, 2014.

[3] J. Massion, "Postural control system," *Current Opinion in Neurobiology*, vol. 4, no. 6, pp. 877–887, 1994.

[4] T. Paillard, "Effects of general and local fatigue on postural control: a review," *Neuroscience & Biobehavioral Reviews*, vol. 36, no. 1, pp. 162–176, 2012.

[5] C. J. De Luca, R. S. LeFever, M. P. McCue, and A. P. Xenakis, "Control scheme governing concurrently active human motor units during voluntary contractions," *Journal of Physiology*, vol. 329, pp. 129–142, 1982.

[6] J. Carriot, M. Jamali, and K. E. Cullen, "Rapid adaptation of multisensory integration in vestibular pathways," *Frontiers in Systems Neuroscience*, vol. 9, article 59, 2015.

[7] M. Gandolfi, C. Geroin, A. Picelli et al., "Robot-assisted vs. sensory integration training in treating gait and balance dysfunctions in patients with multiple sclerosis: a randomized controlled trial," *Frontiers in Human Neuroscience*, vol. 8, article 318, 2014.

[8] J.-L. Honeine and M. Schieppati, "Time-interval for integration of stabilizing haptic and visual information in subjects balancing under static and dynamic conditions," *Frontiers in Systems Neuroscience*, vol. 8, article 190, 2014.

[9] R. L. Vassar and J. Rose, "Motor systems and postural instability," in *Handbook of Clinical Neurology*, vol. 125, chapter 15, pp. 237–251, Elsevier, 2014.

[10] M. F. Gago, V. Fernandes, J. Ferreira et al., "Role of the visual and auditory systems in postural stability in Alzheimer's disease," *Journal of Alzheimer's Disease*, vol. 46, no. 2, pp. 441–449, 2015.

[11] R. Cabeza-Ruiz, X. García-Massó, R. A. Centeno-Prada, J. D. Beas-Jiménez, J. C. Colado, and L.-M. González, "Time and frequency analysis of the static balance in young adults with Down syndrome," *Gait and Posture*, vol. 33, no. 1, pp. 23–28, 2011.

[12] A. M. El-Kahky, H. Kingma, M. Dolmans, and I. de Jong, "Balance control near the limit of stability in various sensory conditions in healthy subjects and patients suffering from vertigo or balance disorders: impact of sensory input on balance control," *Acta Oto-Laryngologica*, vol. 120, no. 4, pp. 508–516, 2000.

[13] T. Paillard, R. Montoya, and P. Dupui, "Postural adaptations specific to preferred throwing techniques practiced by competition-level judoists," *Journal of Electromyography and Kinesiology*, vol. 17, no. 2, pp. 241–244, 2007.

[14] G. C. Gauchard, P. Gangloff, C. Jeandel, and P. P. Perrin, "Physical activity improves gaze and posture control in the elderly," *Neuroscience Research*, vol. 45, no. 4, pp. 409–417, 2003.

[15] M. Pau, F. Arippa, B. Leban et al., "Relationship between static and dynamic balance abilities in Italian professional and youth league soccer players," *Physical Therapy in Sport*, vol. 16, no. 3, pp. 236–241, 2015.

[16] T. Paillard, F. Noé, T. Rivière, V. Marion, R. Montoya, and P. Dupui, "Postural performance and strategy in the unipedal stance of soccer players at different levels of competition," *Journal of Athletic Training*, vol. 41, no. 2, pp. 172–176, 2006.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 891390, 15 pages
http://dx.doi.org/10.1155/2015/891390

Review Article

Techniques and Methods for Testing the Postural Function in Healthy and Pathological Subjects

Thierry Paillard and Frédéric Noé

Physical Activity, Performance and Health Laboratory, Department of STAPS, University of Pau and Pays de l'Adour, ZA Bastillac Sud, 65000 Tarbes, France

Correspondence should be addressed to Thierry Paillard; thierry.paillard@univ-pau.fr

Received 14 July 2015; Revised 5 October 2015; Accepted 21 October 2015

Academic Editor: Jacob J. Sosnoff

The different techniques and methods employed as well as the different quantitative and qualitative variables measured in order to objectify postural control are often chosen without taking into account the population studied, the objective of the postural test, and the environmental conditions. For these reasons, the aim of this review was to present and justify the different testing techniques and methods with their different quantitative and qualitative variables to make it possible to precisely evaluate each sensory, central, and motor component of the postural function according to the experiment protocol under consideration. The main practical and technological methods and techniques used in evaluating postural control were explained and justified according to the experimental protocol defined. The main postural conditions (postural stance, visual condition, balance condition, and test duration) were also analyzed. Moreover, the mechanistic exploration of the postural function often requires implementing disturbing postural conditions by using motor disturbance (mechanical disturbance), sensory stimulation (sensory manipulation), and/or cognitive disturbance (cognitive task associated with maintaining postural balance) protocols. Each type of disturbance was tackled in order to facilitate understanding of subtle postural control mechanisms and the means to explore them.

1. Introduction

The ability to maintain body balance depends on complex organization which is developed with sensory inputs and is based on body geometry (segmental organization), kinetics (ground force reaction), and body orientation and vertical perception (subjective verticality) cues [1]. Pathologies disturbing sensory output, force/movement control, and spatial orientation logically affect postural control [1]. Overall, all pathologies which alter organs specifically involved in the control of posture and movement degrade postural control. For instance, Alzheimer's and Parkinson's diseases, cerebellar and vestibular syndromes, low-vision and ankle sprains, which, respectively, affect the cerebral cortex (parietal lobe involved in spatial orientation and frontal lobe involved in cognition), basal ganglia (especially substantia nigra, whose neurons secrete dopamine involved in the control of movement and posture), cerebellum (involved in movement and balance control), vestibular (involved in head movements' detection), visual (involved in orientation in space), and the ankle capsulo-ligamental (involved in ankle sensitivity and stabilization) systems and disturb postural control [2–9]. Pathological postural attitudes such as idiopathic scoliosis also affect postural control [10]. Given the multitude of structures involved in postural control and because of its complexity, many other pathologies are likely to disturb postural control.

Although pathologies can alter postural control in a nonspecific way, for a given pathology known for affecting particularly the postural function, the postural behavior evolves specifically (e.g., [11, 12]). For caregivers, postural control tests can thus help to determine the pathology in question (or the diagnosis) in patients (e.g., [13–15]). However, it is essential to use adequate evaluation methods and techniques which give reliable quantitative and qualitative variables in order to pinpoint the functional state of the sensory, central, and motor components of the postural function.

Nevertheless, as regards the literature, one can observe that the different techniques and methods employed, as well as the different quantitative and qualitative variables measured, in order to objectify postural control, are often chosen either arbitrarily or on the basis of materials classically used by the authors without taking into account the population under consideration, the objective of the postural task, and the environmental conditions. For these reasons, the aim of this review was to present and justify the different testing techniques and methods with their different quantitative and qualitative variables to make it possible to precisely evaluate each sensory, central, and motor component of the postural function in pathological subjects but also in healthy subjects since they constitute the benchmark in terms of postural behaviour.

2. Principles of Analysis of Postural Control

2.1. Quantitative and Qualitative Analyses. Postural control can be quantitatively considered by measuring the movement of the centre of mass (COM), the centre of foot pressure (COP), and body segments but also by measuring electromyographic activities and evaluations of the contribution of different sensory information. The qualitative analysis consists of describing how postural control is organized in relation to the mechanical and neurophysiological aspects.

2.2. Postural Performance. Postural control can be characterized in terms of performance according to the postural condition under consideration. Postural performance refers to the ability to maintain body balance in challenging postural conditions (e.g., a stance classed as a handstand, monopedal dynamic stance) and thus avoiding postural imbalance and falls. Postural performance can also characterize the ability to minimize body sway in more conventional postural conditions (e.g., bipedal quiet stance).

2.3. Postural Strategy. It can be defined on the basis of the spatial and temporal organization of different body segments as well as the extent and order of recruitment of different muscles activated. The different sensory sensors involved in postural regulation as well as the weight of different sensory information and/or the preferential involvement of different neuronal loops can also contribute to describe postural strategy.

3. Testing for Postural Performance and Strategy

3.1. Testing for Postural Performance. The ability to ensure postural stability in challenging postural conditions can be evaluated with practical or experimental tests with different postural stances (e.g., bipedal stance, monopedal stance) on small bases of support and moving platforms leading to expected and unexpected postural disturbances. Subjects retain their body balance or not and then pass the test on offer or not which corresponds to a certain performance level. If the test consists of discriminating between the ability to minimize body sway in easy and unspecific postural conditions, different instrumented evaluation methods can be employed.

3.2. Testing for Postural Strategy. The use of instrumented evaluation methods is sometimes insufficient to precisely characterize the postural strategy employed by subjects. Evaluation of the contribution of each component of the postural function often involves motor disturbance (mechanical disturbance), sensory stimulation (sensory manipulation), or cognitive disturbance (e.g., virtual simulation, dual-task) protocols. Methods combining these different techniques also provide relevant information in analysis of the postural function.

4. Basic Noninstrumented Postural Tests

Most of the time, the evaluation of postural function requires technological materials but simple tests can also be used to identify postural dysfunctions in aged and frail subjects and subjects with pathologies (acute and chronic pathologies). However, basic postural tests were mainly designed to evaluate older subjects' postural abilities as well as their risk of falling whereas there are only a few tests for subjects with pathologies. A number of tests exist, such as the Berg Balance Scale [16, 17], Timed Up-and-Go [18, 19], Tinetti test [20], Short Physical Performance Battery [21], Mini Balance Evaluation Systems Test [22], Unified Balance Scale [23], Functional Ambulation Classification [24, 25], and the Postural Assessment Scale for Stroke patients [26] for example. Currently, the Berg Balance Scale or Mini Balance Evaluation Systems Test would be the tests particularly recommended by certain experts [27]. Moreover, it is known that failure to maintain the monopodal stance for 5 seconds constitutes a strong risk of falling for older people even if this very discriminating test on its own does not predict all falls that might occur in their life [28]. This monopedal stance can be suggested for 30 seconds (3 trials), and either the subject passes the test or he/she fails. In the event of the latter, one can record the holding time from the best trial (if this is less than 14 seconds, postural abilities are considered very weak). Moreover, walking speed tests over a 4-metre distance with a chronometer also make it possible to evaluate the postural abilities of older subjects. For example, a walking speed corresponding to $0.8\,\mathrm{m{\cdot}s^{-1}}$ is predictive of weak functional abilities while a speed corresponding to $0.6\,\mathrm{m{\cdot}s^{-1}}$ constitutes a threshold below which the risk of falling is critical [29]. However, these practical tests are of interest to subjects whose postural abilities are very weak but they do not make it possible to carry out qualitative analyses of postural control, especially for (young) subjects with pathologies. Only technology and instrumented tests offer this possibility.

5. Material and Technology for Instrumented Tests

Even though noninstrument tests can be useful to the clinician in diagnosing sensory-motor disorders, they only provide a gross indicator of postural control efficiency. Detailed analysis of postural control performance and associated strategies require the use of instrumented tests with various materials to make it possible to carry out kinetic, kinematic, and electrophysiological analysis.

5.1. Kinetic Devices. Force platforms are the most widely used devices in assessing postural function. Force platforms are made of a dimensionally stable board under which load sensors are positioned. They can be incorporated in specific motorized or nonmotorized devices in order to generate instability.

5.1.1. Main Technologies. The most widely used nonmotorized devices are wobble boards, usually made of wood or plastic materials, with hemispherical or hemicylindrical bases (seesaws) that create instability in all spatial directions or a given plane [30]. Instability can be modulated according to the radius and height of the base. While reducing ground surface contact and raising feet surface contact, wobble boards challenge both sensory and motor components of the postural control system [31]. Indeed, standing on a wobble board requires the centre of mass (COM) to be projected onto the board's point of contact with the floor [32], thereby increasing postural sway and challenging the postural control system when compared to standing on stable ground [30]. Wobble boards sometimes include autonomous measurement devices—mainly potentiometers recording the discrepancies of the seesaw from the horizontal plane—and do not require the use of a force platform [33]. Although such devices are affordable and can be used for sports training and balance rehabilitation, they only provide a macroscopic postural sway index without directional characteristics that are required for a suitable assessment of postural function. Since the pioneering work by Nashner [34], many studies have been conducted with servo-controlled motorized force platforms. Most current advanced devices can provoke cyclic or sudden translational movements in the medial lateral (ML) and/or anterior/posterior (AP) direction and rotational movements in all directions or a given plane.

When focusing on the technology of force platforms, two "families" of platforms can be considered: (1) those equipped with monoaxial load cells that only measure the vertical component of the ground reaction force (FZ), usually with at least three strain gauges (uniaxial plates) and (2) those equipped with load cells (usually four strain gauges or piezoelectric sensors) that measure the three components of the ground reaction force (FX, FY, and FZ) and the moment of force acting on the plate (MX, MY, and MZ) (multiaxial pates) [35]. Both uni- and multiaxial plates can be used to calculate the ML and AP time series of the centre of pressure (COP, the point of application of the vertical ground reaction force) over time during a postural test. The COP is the most measured parameter to assess postural function. Postural sway is commonly applied to variations in the COP position, whereas displacements of the COM are applied to body sway [36]. With multiaxial plates, the relative horizontal COM displacements can be calculated thanks to a double integration of the ML and AP components of the ground reaction force (divided by the mass). Nevertheless, it is not possible to calculate the initial velocity and position of the COM, even though some methods have been suggested to estimate these initial constants [35]. COM horizontal positions can also be evaluated from COP displacements measured with both uni- and multiaxial platforms by using an inverted pendulum model and a filtering method based on the COM/COP relationship in the frequency domain [37]. Nevertheless, only kinematic analyses make calculation of COM motions in three spatial directions possible.

Force platforms initially designed to be used as video game controllers have also been recently suggested as very affordable tools to assess postural function. Many studies have been conducted in order to compare multiaxial platforms with these particular unidirectional platforms, characterized by inconsistent and low sample rates with a large amount of irrelevant results. Such devices tend to overestimate COP parameters such as velocity [38]. The overestimation of COP parameters appears to be a typical feature of uniaxial force plates and depends on the postural task's complexity—the easier the postural task, the larger the overestimate [38, 39]. Despite this limitation, monoaxial force plates provide appropriate accuracy for most standing balance assessments. Nevertheless, measurements from unidirectional and multiaxial platforms should not be used interchangeably [39]. Whatever the type of platform used, they must meet further requirements whose standards have been recently updated [40]: accuracy should be better than 0.1 mm, precision should be better than 0.05 mm, resolution should be higher than 0.05 mm, frequency bandwidth should be 0.01–10 Hz, and linearity should be better than 90% across the whole range of measurement parameters.

Since the COP comes from the muscle actions of both feet, the use of two platforms placed side by side can be required in order to analyze in detail the balance mechanisms in the frontal plane by measuring the ground reaction forces under each foot [41, 42], especially if body weight distribution asymmetry is suspected, as with hemiparetic or amputee patients [43]. It is also possible to distinguish the hip loading/unloading mechanism from the ankle inversion/eversion mechanism acting on the frontal plane with two platforms [42]. Some force plates also make it possible to separately analyze the COP movements at the heel and those of the metatarsus under each foot. Some authors have developed specific measurement devices to analyze more complex postural conditions. Examples include the concomitant use of the force platform and force transducers positioned on handles to analyze postural tasks performed while using hand supports [44] and specific ergometers equipped with 3D load sensors positioned on feet and hand supports to analyze horizontal [45] and vertical [46] quadrupedal postures. Devices using pressure sensors as flexible instrumented insoles [47] or pressure plates [48] can be used to measure plantar pressure distribution, especially when modulating conditions related to footwear. Plantar pressure measurements provide information regarding potential impairments of the foot and its disorders [49]. Among all these kinetic devices, force platforms are considered to be the gold standard [38], with COP being the most widely measured parameter from which various variables can be calculated to assess postural function [35].

5.1.2. Main COP Variables. Raw COP recordings are mainly used by clinicians and researchers as gross visual representations of the output of the postural control system. Two representations can be obtained, the statokinesigram (construction of the COP map in the horizontal plane) and the stabilogram (time series showing variation of the COP in the AP and ML directions). Nevertheless, the calculation of other COP variables from raw COP data is necessary in order to analyze the mechanisms involved in postural regulation. COP variables can be categorized as global and structural variables [35, 50–52]. Global variables characterize the magnitude of the resultant and/or the ML and AP components of the COP traces in both time and frequency domains. Authors usually consider that the greater the magnitude or deviations of a global variable, the poorer the postural stability. Nevertheless, global variables are not sensitive to the structure of variation which can potentially provide essential insights into the postural control process in a variety of contexts [50, 51]. Hence structural variables can be considered. These variables decompose the COP sway patterns into subunities and correlate them with the motor control process [35, 50–53].

(1) Global COP Variables. Many different global variables have been put forward [50]. Making an exhaustive list of all these variables is not the concern of this study and only the most common and relevant ones are given here and commented on.

(i) Mean coordinates reflect the topographical features of plantar pressure distribution [54] and depend on the position of the subjects on the force plate. They can be influenced by wearing specific shoes (e.g., ski-boots [55]) and anthropometric characteristics [56]. They can also be used as a clinical index to detect specific pathologies resulting from bilateral unbalance.

(ii) Ellipse area/surface quantifies 90 or 95% of the total area covered in the ML and AP direction using an ellipse to fit the data. It is considered to be an index of overall postural performance—the smaller the surface, the better the performance [57]. Caution must be taken when calculating this variable and the use of prediction ellipses should be preferred to confidence ellipses [58, 59].

(iii) Path length quantifies the magnitude of the two-dimensional displacement based on the total distance travelled. It is considered to be a valid outcome measurement in numerous populations and balance conditions—the smaller the path length, the better the postural stability [39].

(iv) Amplitude of displacement is the distance between the maximum and minimum COP displacement for each direction—the greater the values, the worse the postural stability. COP amplitude is a reliable parameter which has been widely used in order to analyze postural deficits with patients suffering from neuromotor disorders such as cerebral palsy, especially when analysis was conducted on the ML direction [60].

(v) Velocity is calculated by dividing the COP excursion by the trial time. One can consider the ML and AP components or the resultant velocity. It reflects the efficiency of the postural control system (the smaller the velocity, the better the postural control) while characterizing the net neuromuscular activity required to maintain balance [61] and has been considered as the measurement with the greatest reliability among trials [35]. Additional authors considered COP velocity as the most sensitive parameter in comparing individuals from different age groups and with different neurological diseases [62–64]. Vsetecková and Drey [65] also underlined the major role of COP velocity in the feedforward mechanisms of the postural control system during quiet stance.

(vi) Standard deviation (SD), root mean square (RMS): RMS is defined as the square root of the mean of the squares of a sample. If the COP signal has zero mean, RMS and SD provide the same result. RMS and SD are variability indexes of COP movements which offer good reliability in discriminating between young and older subjects and subjects who are healthy and those with pathologies [35, 66–68].

(vii) Total power frequency is considered an energy-expenditure index.

(viii) Mean, median, centroid, and 80–95% power frequency: these parameters provide a general view of the frequency content of the COP signal. Higher frequencies of postural sway are indicative of postural control with faster and smaller postural adjustments [69]. Mean and median frequency can also be viewed as indexes of ankle stiffness—the higher the frequency of postural sway, the higher the stiffness around the ankle joint [70]. 80–95% power frequency characterizes the frequency band with 80–95% of the spectral power. Baratto et al. [50] suggest 80% of spectral power is the best value to characterize modifications in the postural control system.

(ix) Frequency bands distribution: the frequency content of the COP signal is studied by incorporating amplitudes within frequency bands in order to characterize the preferential involvement of specific neuronal loops in postural regulation [71–74]. Three frequency bands are usually considered: low frequencies (0–0.02/0.5 Hz) which mostly account for visuovestibular regulation, medium frequencies (0.2/0.5–2 Hz) for cerebellar participation, and high frequencies for proprioceptive participation (>2 Hz), the limits of these bands being different according to the authors [66, 71].

Spectral analyses of COP sway are usually performed with algorithms based on Fourier transforms [35, 69]. These methods should be used with caution since the COP can demonstrate nonstationary characteristics [69, 75]. Computational approaches such as discrete wavelet analysis [75] or empirical mode decomposition [69] are more suitable for nonstationary signals.

(2) Structural COP Variables. Because of the nonstationary characteristic of the COP signal, standard time and frequency analysis methods cannot adequately describe the dynamic changes of postural sway. Because the postural control system must be considered as a nonlinear system (where reactions are not proportional to the applied stimuli), various methods of nonlinear dynamics and quantitative descriptors have been put forward for the analysis of the COP signal [76].

Collins and De Luca [53] introduced a method for analyzing time evolutionary properties of the COP known as stabilogram diffusion analysis. They assumed that maintenance of erect posture could be considered as a stochastic process governed by the laws of probability. Stochastic analysis is focused on the evolution of complex structures resulting from interactions between numerous elements. From a stochastic perspective, the COP time series is considered as the performance of a theoretical process consisting of random variables relating to points in time, with this random theoretical process being analyzed by performing a statistical inference on its properties [77]. Collins and De Luca [53] decomposed the COP signal into two stochastic processes modelled as Brownian fractional movements: a long-term process with a large exponent characterizing a persistent structure and a short-term one with a small exponent characterizing an antipersistent structure. These two structural sub unities were considered to, respectively, characterize the closed and open-loop mechanisms of human postural control [53]. Nevertheless, the existence of a critical point in time that distinguishes open- and closed-loop processes in postural control has been challenged and the functional significance of this model is no longer widely accepted [78, 79].

With the rambling-trembling hypothesis, Zatsiorsky and Duarte [80] put forward an alternative method which also differentiates between two timescale components in the COP signal. In the context of the equilibrium-point hypothesis [81], they suggest that equilibrium is adopted according to a migrating reference point, characterized by the conservative rambling subsystem, whose movements reflect an exploratory behaviour which does not induce substantial restoring forces. The oscillations around this reference point characterize the operative trembling subsystem which aims at maintaining equilibrium around the reference point thanks to restoring forces [82]. Rambling and trembling subsystems describe two different processes in the control of an upright stance: rambling reflects the supraspinal processes involved in the control of the movements of the reference point, whereas trembling reflects spinal reflexes and changes in the intrinsic mechanical properties of the muscles and joints [82].

Another approach of COP structural analysis is based on the assumption that the postural control system is a chaotic system with a deterministic nature [76]. Fractal dimension methods have been put forward in order to detect chaos in posturographic signals. Decreased postural stability due to lack of visual cues [76] or neurological pathologies [83] is characterized by an increase in the signal's fractal dimension. Fractal analysis of COP signals represents a reliable and sensitive tool to assess subtle changes in postural control caused by a pathology and/or age [76, 83]. Sample entropy (SampEn, a measure of regularity), approximate entropy (ApEn, a measure of unpredictability), and Lyapunov exponent (LyE, a measure of divergence) are nonlinear dynamic parameters that can be extracted from COP plot points in order to perform structural analyses [52, 84, 85]. Significant regularity in postural control resulting in low values for SampEN, ApEn and LyE characterizes constraint systems with reduced adaptation and response aptitudes to potential disturbances and increased risk of falling. Patients suffering from neurological disorders typically demonstrate lower SampEN, ApEn, and LyE values compared to healthy subjects and this reflects impairment in postural function [85, 86]. Unconstrained and irregular postural oscillations reflect the efficiency of postural control related to the complex mechanisms with structured variability but not exact repetition [87].

Additional authors have put forward other specific methods for COP structural analysis. One can mention the sway-density curve concept from Baratto et al. [50], based on the idea that COP movements are incompatible with Brownian movement, the structural analysis proposed by Duarte and Zatsiorsky [88], which requires carrying out prolonged postural tasks in order to identify timescale components in the COP signal, the empirical mode decomposition put forward by Pachori et al. [69], which decomposes the COP signal into intrinsic mode functions (i.e., local oscillations that compose the raw COP signal), the entropic half-life approach from Baltich et al. [89], which makes it possible to quantify the SampEn of the COP with different time scales without affecting the signal length, or the rotary spectra approach proposed by Agostini et al. [90], which separates rotational iso-frequential components of the COP signals from nonrotational ones. These approaches are developments of preexisting methods detailed more precisely above or which require more research in order to test their reliability and validate their relevance for clinical applications.

5.2. Kinematic Devices

5.2.1. Main Technologies. Even though basic video recordings can provide both qualitative and quantitative information about segmental postural strategies, especially when using specific advanced software [91], only 3D motion capture systems offer the high level of accuracy and reliability necessary to record the small motions which characterize the unperturbed upright stance [92]. Two different technologies can be identified. Passive marker systems use reflective markers with a set of further high-resolution, high-speed cameras with incorporated infrared/near infrared strobes. The cameras record the reflection from the markers placed on specific anatomic landmarks whose identification is performed thanks to the software. Active marker systems use powered markers sending an infrared signal which is captured by sensor units. Each active marker has its proper frequency. Active marker systems avoid the postprocessing identification procedures required with passive marker systems but require small powered boxes to be attached to the subjects' skin.

The use of 3D body-worn accelerometers has recently been suggested as an alternative to force platforms for measuring postural sway [93]. Accelerometers can be positioned on the posterior trunk to give an estimate of COM movements [93] or on specific joints to assess joint movements and/or COM movements thanks to subsequent modelling and calculation [94]. Accelerometer-based devices provide a sensitive means of measuring subtle balance deficits in clinical settings [93, 94].

Electrogoniometers make it possible to measure joint angular displacements and have been mainly used to analyze changes in segmental postural coordination while using the dynamic approach to postural control (e.g., [95]). Electrogoniometers provide a first level of accuracy, which is acceptable for dynamic postural tasks [95], but it might be inadequate for measuring joint movements in static postural tasks with healthy subjects.

Laser-displacement sensors can also offer interesting possibilities for kinematic measurements in order to compute joint angle measurements [96, 97] or to analyze the movements of a specific body landmark like a lumbar vertebra, whose movements can be incorporated into a procedure to estimate COM displacements [64]. Laser displacement sensors can provide a very high level of accuracy, making it possible to get reliable measurements of angular motion for subsequent derivative calculations [96].

5.2.2. Main Kinematic Variables. Because of the complexity of the musculoskeletal system, kinematic analyses are always associated with using a biomechanical model with a different degree of complexity. Biomechanical models usually consider the body as a system made up of rigid articulated segments—the more segments and the more freedom of the joints, the more complex the model. Whatever the complexity of the model, the calculation of joint angles can be viewed as a first step that makes it possible to characterize skeletal alignment and assess the overall segmental postural organization [98]. Velocity, acceleration, and jerk calculations provide additional information about joint movement characteristics. Joint moments can be calculated by inverse dynamics when performing more complex analysis combining force plate and kinematic measurements [92]. While using accelerometric devices on the belt to quantify postural sway, Mancini et al. [93] have shown that jerk was the most discriminating parameter to differentiate sway in subjects with Parkinson's disease compared to healthy control subjects. It must be noted that classic movement descriptors can be calculated independently of the type of kinematic device employed while using integration/derivation procedures with accurate filtering and data smoothing procedures [41].

As COM is the only variable that characterizes body sway, its calculation has been of major importance particularly in understanding the relationships between the COM and COP [36]. Despite the widespread use of COM, its calculation is a complex and time-consuming operation which requires a multisegmental model of the body. Winter et al. [99] recommends a 14-segment model with 21 markers. Segmental inertial characteristics must also be estimated thanks to anthropometric tables [99] or optimization procedures [100]. Once the COM is being calculated, similar parameters to those described in Section 5.1.2 with the COP can be calculated.

The analysis of interjoint coordination is a major concern when studying multisegment movement strategies of postural control. Many methods have been put forward in order to identify joint synergies and/or quantify the respective contributions made by joint motions in the control of COM or COP, such as principal component analyses [101], multivariate canonical correlation analyses [102], coherence and cophase analyses [103], coherence spectrum analyses [96], relative phase estimates [104], cross-correlation analyses [92], or wavelet coherence analyses [105]. Similar analyses can also be conducted in order to analyze organization and coordination of physiological tremors during postural tasks [106].

5.3. Electromyography. Electromyographic (EMG) recordings have been widely used in the assessment of postural function. Amplitude, temporal, and frequency parameters can be differentiated [107]. Temporal EMG analyses have been extensively used in order to characterize postural responses following platform-movement disturbances or anticipatory postural adjustments with voluntary movements [108] when identifying bursts of muscle activity [109]. EMG amplitude analyses, like RMS or area calculations [55], are used to reflect the magnitude of muscular activity in maintaining specific postural tasks. Frequency domain analyses have been used with moving oscillating platforms and have shown that increased frequency of platform oscillations increases the amplitude spectrum of muscle activity [110].

EMG recordings can also be used in order to study postural segmental strategies and interjoint coordination. Cross-correlation analysis can be applied to investigate the relationships between COP/COM time series and time domain EMG data [111]. The EMG activity of postural muscles during stable standing tasks has also been analyzed in the frequency domain with coherence analysis in order to determine coordination patterns [112].

6. Main Postural Conditions

6.1. Postural Stances. Generally speaking, postural tests are done either in a bipedal stance or in a monopedal stance. In a bipedal stance, the feet are positioned according to a multitude of possible conditions, the main ones being: feet apart, feet together, semitandem, and full tandem (Figure 1). The legs are generally straight (they can be also flexed if required for the protocol) and the feet either form an angle of 15–30° or are positioned parallel (e.g., [113]). The intermalleolar distance is usually 5 cm for an angle of 30° and 10–15 cm for an angle of 15° [61, 113] and varies between 0 and 15–20 cm for quasi-parallel feet [114]. In a monopedal stance, the supporting leg can be the dominant leg or the nondominant leg according to the aim of the postural test (the dominant leg is the one used for kicking a ball). The supporting leg can be extended or flexed according to the requirements of the experiment. The nonsupporting leg can be either raised and flexed 90° at knee-level or lifted so that the subject's big toe touches the medial malleolus of the supporting leg lightly (no support). The two hips are placed in a neutral position (0° of flexion or slight flexion if the toes

Figure 1: Illustration of the different foot positions.

touch the medial malleolus) if the supporting leg is extended, which is different if the supporting leg is flexed. Whatever the postural stance chosen from among the different possibilities mentioned above, subjects stand in a relaxed manner with arms extended out to the sides or crossed in front of their chest. When arms are moving freely, postural performance is modified [115].

In certain circumstances with pathological subjects or healthy highly skilled subjects, other postural stances can be adopted. Hence, the other possible main supports are ischium (seated with or without feet support), knee (kneeling), and hands (a stance classed as a handstand for skilled sportsmen or quadrupedal postures) [45, 46, 57].

For all the postural stances mentioned above, other body segments can be also used as additional supports: hand (one or two), trunk, head, thigh, shank, arm, and forearm. Analyses of the contribution of haptic information (proprioception and cutaneous sensitivity combined) in postural regulation as well as the application of different biomechanical conditions justify the use of these additional supports.

6.2. Visual Condition. Most of time the postural tests are completed without and/or with visual information. The suppression of visual cues may occur through closing the eyes or blindfolding but also by putting subjects in total darkness. Moreover, the contribution of visual cues (calculated with quantitative and qualitative variables obtained in the closed eyes condition compared to the same variables measured in the open eyes condition) constitutes relevant data in the analysis of postural control in subjects who are healthy and who have pathologies.

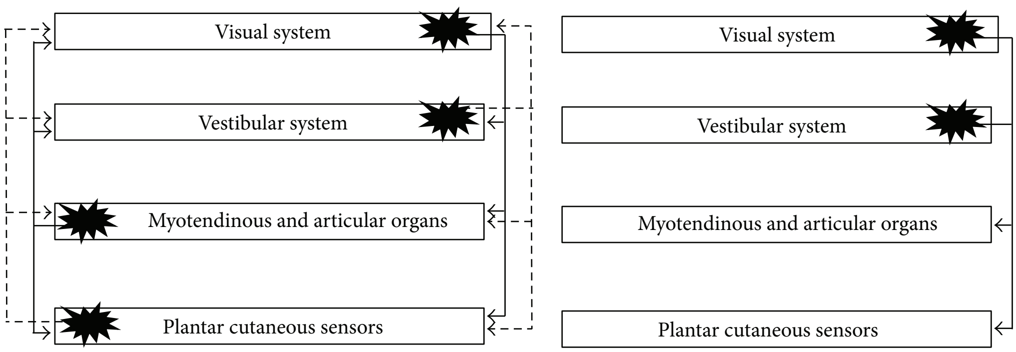
6.3. Balance Condition. Both static and dynamic conditions are used when testing postural control. For subjects with pathologies, it is prudent to start with postural tests in static conditions. Dynamic conditions are more discriminating than static conditions in terms of postural control [5]. The contribution of visual cues is essential in static conditions while the contribution of proprioception inputs is fundamental in dynamic conditions [56]. However, when the difficulty of postural task increases in dynamic conditions, the contribution of visual information increases [5].

6.4. Duration of Tests. In the literature, one can find different durations of test for evaluating postural control. Generally speaking, the duration of tests in static conditions is longer than that observed in dynamic conditions. One can estimate that appropriate durations mainly vary between 20 and 60 s for static conditions and between 10 and 30 s for the dynamic conditions depending on the difficulty of the postural task and the population under consideration (e.g., subjects with pathologies, older subjects, highly skilled subjects). In static conditions, a 20-s duration would be the minimum under which the postural test may lose consistency since the stationary process (stationariness of the posturographic signals) of postural control requires some seconds of adjustment time [40, 116]. The last meeting of the International Society for Posture and Gait Research suggested that from a recording time of 25–40 s the posturographic parameters are steady and reliable and a reasonable comprise could be 30 s with 5 s of adjustment time before starting the recording [40]. In turn, the complexity of evaluation protocols sometimes involves longer durations in specific physiological and/or psychological (or cognitive) conditions. Nevertheless, the experimenter should ensure that the test duration does not cause fatigue especially in subjects with pathologies. In dynamic conditions, a 30-s duration seems to be the maximum in order to avoid fatigue in healthy subjects, while this duration should be shorter for subjects with pathologies. A 15/25-s duration for healthy subjects and a 10/20-s duration for subjects with pathologies seem to be appropriate.

7. Disturbing Postural Conditions

Different evaluation methods make it possible to explore each component of the postural function with motor disturbance (mechanical disturbance), sensory stimulation (sensory manipulation), and/or cognitive disturbance (e.g., cognitive task associated to postural balance maintenance) protocols.

7.1. External Mechanical Disturbance. The first principle making it possible to destabilize body balance consists of mechanically creating COM displacements thanks to external disturbances. To this end, unexpected disturbances produced



(a) Disturbance of one sensory sensor (b) Disturbance of two sensory sensors

Figure 2: The disturbance of one (a) or two (b) sensors leads to an increase in the sensory contribution of other sensors (not disturbed) in postural regulation. The disturbance is indicated by a star-shaped sign while the increase in sensory contribution is indicated by an arrow.

by percussion or pushing a large body segment such as the trunk can produce mechanical disturbances which require effective postural reactions in order to maintain body balance. The second principle consists of modifying the state of the base of support with moving platforms (e.g., translation, pitching, rolling or yaw movements) and surface reductions to this base. Finally, the third principle consists of applying articular constraints by limiting or blocking joint movements (cervical and lumbar spine, hip, knee, and ankle) by means of orthotic devices, specific equipment (specific shoes or clothing), collars, and so forth. This principle involves mechanical compensation of joint constraints by changes in postural strategy by reorganising muscle coordination which is made possible by inherent redundancies in the human body [55]. These particular constraints result in not only mechanical constraints but also sensory disturbances since a mere cervical collar effectively joins the head and trunk which limits the information from cervical articulations [117].

7.2. Sensory Disturbance. In order to study the contribution of different sensory sensors to postural regulation, the experimenter often uses sensory manipulations of one (simple manipulation) or two/three (combined manipulation) sensory sensors. The disturbance in one or several sensory sensors impacts the contribution of other sensory sensors (Figure 2). The sensory manipulation technique makes it possible to evaluate the efficiency of different sensory sensors (i.e., the ones that are not manipulated and make it possible to regulate postural control), to identify the predominance of a particular sensor among all the sensors or the preferential use of certain sensory information (i.e., the sensor that when manipulated induces greater postural disturbances than when the other sensors are individually manipulated), and to define the capacities to compensate and/or switch the different sensory inputs (i.e., the abilities to limit the effects of postural disturbance through the increased contribution of sensory sensors which have not been not manipulated) [118–120].

7.2.1. Visual Disturbance. The alteration of visual cues can be generated through the reduction or suppression of brightness and/or field visual. The experimenter can reduce the visual flow with stroboscopic light, light filters and other processes intended to limit the availability of visual information. He/she can also move the visual target away from the subject in order to attenuate the visual effects on postural control [121]. Visual disturbances can also be created by giving erroneous visual cues through the application of the optokinetic technique [121]. This makes it possible to project a moving visual scene on a subject/patient who is standing. It triggers nystagmus in the direction selected by the experimenter and causes postural deviation [122]. An optokinetic stimulus induced by the rotation of a disc from the left side to the right side causes an inclination of the body to the right side to compensate for the body motion illusion to the left. The purpose is not to destabilize the subject/patient, but to provoke neurosensory conflicts since proprioceptive, vestibular, and plantar cutaneous inputs indicate no movement.

7.2.2. Vestibular Disturbance. In order to study the contribution of vestibular inputs in postural regulation, the vestibular afferences can be disturbed with particular electrical stimulations. These disturbance stimulations are done with the galvanic vestibular stimulation technique. It consists of provoking neurosensory conflicts by applying an electrical current via surface electrodes to the mastoid processes [118, 119, 123]. This electrical current disturbs the transduction of ciliated cells in ampullary crests (in semicircular canals) and macula (in otoliths) which induces body motion illusions [123, 124] and modifies postural attitude [125–127] but does not change the internal representation of the subjective vertical [127]. Galvanic vestibular stimulation can be applied unilaterally or bilaterally through monopolar or bipolar stimulus [128, 129]. A bilateral and bipolar stimulation provokes tilting on the medio-lateral axis towards the anode electrode [130]. Bilateral and monopolar stimulation creates tilting on the anteroposterior axis, backward for anode electrodes, forward for cathode electrodes [129]. The head should be vertically placed (not inclined) because its position can influence the postural response [131]. The intensity of stimulation influences the postural response—the higher the intensity, the greater the postural reaction [129, 130]. The disturbance intensities raised in the literature go from 1 mA to 5 mA

[124, 132]. Higher intensities are feasible but would not be harmless in terms of the risk of burning the subject's skin [132]. The delay in postural response to stimulation is about 1-2 s [124]. The experience of more natural stimulation of the vestibular system, that is, through accelerations of the head movement through specific physical activities (e.g., control subjects versus pilots), can limit the magnitude of body deviation [125]. This study showed that pilots have a stronger ability to suppress vestibular illusions than control subjects. Moreover, the risk of body destabilization (falling) of subjects/patients is real when using galvanic vestibular stimulation, so the experimenter must ensure he/she applies progressive intensities especially with subjects who are impaired or have pathologies. For example, individuals with Down's syndrome showed greater sensitivity to galvanic vestibular stimulation than control subjects and were not able to select the appropriate motor strategy to efficiently maintain balance and compensate for the effects of galvanic vestibular stimulation [133].

7.2.3. Proprioception Disturbance. The proprioceptive disturbance is mainly studied by manipulating myotatic and tendon sensors since the manipulation of articular sensors is done with articular constraints or blocking. Tendon vibration and neuromuscular electrical stimulation are the two techniques mainly used to disturb the myotendinous complex.

Tendon vibration applied onto muscle belly or tendon modulates the afferences of fibres of type Ia [134, 135]. Muscle spindle secondary endings (fibres II) and Golgi tendon organs (fibres Ib) would be either insensitive or only slightly sensitive to tendon vibration in relaxed muscles [135]. Tendon vibration induces perceived muscle stretching as well as body motion illusion which results in modification of body orientation [136]. Vibratory stimulation provokes body inclination backwards when it is applied to the triceps surae [137] and provokes body inclination forwards when it is applied to the tibilis anterior [138]. The vibratory frequency and amplitude usually used are, respectively, between 30 and 100 Hz and between 0.2 and 3 mm [135, 139–141]. The stimulation frequency influences the muscle response—the higher the values, the greater the postural reaction. Vibrations below 20 Hz induce mechanical resonance [142]. Finally, the risk of body destabilization in subjects/patients is real when using tendon vibration, so the experimenter must ensure he/she applies progressive frequencies especially in subjects who are impaired or who have pathologies.

Neuromuscular electrical stimulation can also be employed to disturb the contribution of myotatic loop in postural regulation [117, 143]. It is applied either onto muscle belly or on nerve. The frequency and intensity values of stimulation probably influence the disturbance effects on postural control but they are currently unknown.

7.2.4. Plantar Cutaneous Disturbance. Overall, there could be three main techniques to reduce or suppress plantar cutaneous sensitiveness. The first technique consists of anesthetizing the sensitivity of cutaneous receptors through hypothermia by placing the plantar sole in iced water (0–2°C or 0–5°C) for some minutes (e.g., 10 or 20 min) in order to disturb postural control [117, 139, 144]. The second technique consists of using a foam-supporting surface which appears to be an appropriate tool to challenge postural control and produces substantial and multidirectional balance disturbance [145, 146]. Static standing on a foam surface would change the multiple biomechanical variables in the foot, resulting in an alteration to the distribution of plantar pressures [147]. The third technique consists of provoking ischemia by partially blocking blood circulation in the ankle or thigh [148–150]. Ischemia produces local metabolic changes that would alter the sensory pathways and would consequently affect the activity of the muscles involved in postural control [151]. This study suggested that these changes would cause a decrease in the monosynaptic facilitation of homonymous motoneurons linked to afferents Ia and a polysynaptic disfacilitation in motoneurons linked to cutaneous afferents. In a clinical context, the foam-supporting surface seems easier to safely use than the cooling technique (hypothermia) and especially the ischemia technique in order to study the contribution of plantar cutaneous inputs in postural regulation.

7.2.5. Combined Materials. This type of device comprises a force platform and a cabin which can be mobilized (tilted) either together or separately. Tilting the platform and/or the cabin combined with the elimination of visual information consists of creating sensory conflicts. Tests are performed in different sensory conditions in order to study how subjects cope with modifications to the environment. This type of device makes it possible to conduct postural evaluations in different sensory conditions:

(i) all the sensory information is available,

(ii) the visual information is eliminated: blindfolded,

(iii) the visual information is disturbed: the cabin is tilted (eyes open),

(iv) the proprioceptive information is modified: the force platform is tilted,

(v) the visual information is removed and proprioceptive information is changed, blindfolded, and the cabin is tilted,

(vi) the visual and proprioceptive information is inadequate: the platform and the cabin are tilted.

7.3. Cognitive Disturbance. Postural control system is not totally autonomous and requires attentional resources [152]. Many studies have produced evidence that the attentional demands of postural control increased with ageing, the difficulty of the postural task, the absence of information from a sensory system, and pathology or injury [153–157]. The investigation of the attentional demands of postural control broadly involves the use of dual-task paradigms [152, 157]. Dual-task paradigms are based on the assumption that the central nervous system has limited processing resources and when two tasks are performed at the same time, they can interfere if they imply the use of shared resource requirements from similar specialized structures [152, 158]. Hence, when postural control is associated with a secondary cognitive task, interference implies a shared requirement for attentional processes. Dual-task paradigms can be used to focus on just the attentional demands required for postural control during a cognitive task [159, 160]. Cognitive tasks such as a calculation task, memory task, visual search task, or verbal fluency task, as well as tasks based on biofeedback techniques (e.g., games-based balance exercise), are generally undertaken simultaneously during postural tasks [161]. Other tasks such as the ones which imply responding as quickly as possible to auditory, visual, and sensory signals make it possible to evaluate reaction time alone or combined with postural tasks. According to the population under consideration (e.g., subjects with pathologies and who are older and subjects who are healthy and sportsmen), cognitive tasks can be developed to suit their abilities.

8. Conclusion

Currently, whatever the population under consideration (healthy or subjects with pathologies), the objective of the postural task and the environmental conditions, postural control can be appropriately evaluated in terms of postural performance and strategy by using reliable technological tools and tests. However, all the theoretical considerations related to the postural function are not yet experimentally verifiable through postural analyses. Refinement in the analysis of the contribution of sensory, central, and motor components to postural behaviour is subject to future technological progress as well as advances in knowledge about postural function.

Conflict of Interests

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

References

[1] J. Massion, "Postural control system," *Current Opinion in Neurobiology*, vol. 4, no. 6, pp. 877–887, 1994.

[2] C. Bruttini, R. Esposti, F. Bolzoni, A. Vanotti, C. Mariotti, and P. Cavallari, "Temporal disruption of upper-limb anticipatory postural adjustments in cerebellar ataxic patients," *Experimental Brain Research*, vol. 233, no. 1, pp. 197–203, 2014.

[3] G. G. Tangen, K. Engedal, A. Bergland, T. A. Moger, and A. M. Mengshoel, "Relationships between balance and cognition in patients with subjective cognitive impairment, mild cognitive impairment, and Alzheimer disease," *Physical Therapy*, vol. 94, no. 8, pp. 1123–1134, 2014.

[4] G. Dirnberger and M. Jahanshahi, "Executive dysfunction in Parkinson's disease: a review," *Journal of Neuropsychology*, vol. 7, no. 2, pp. 193–224, 2013.

[5] M. S. V. Tomomitsu, A. C. Alonso, E. Morimoto, T. G. Bobbio, and J. M. D. Greve, "Static and dynamic postural control in low-vision and normal-vision adults," *Clinics*, vol. 68, no. 4, pp. 517–521, 2013.

[6] C. E. Hiller, E. J. Nightingale, C.-W. C. Lin, G. F. Coughlan, B. Caulfield, and E. Delahunt, "Characteristics of people with recurrent ankle sprains: a systematic review with meta-analysis," *British Journal of Sports Medicine*, vol. 45, no. 8, pp. 660–672, 2011.

[7] B. Schoch, A. Hogan, E. R. Gizewski, D. Timmann, and J. Konczak, "Balance control in sitting and standing in children and young adults with benign cerebellar tumors," *Cerebellum*, vol. 9, no. 3, pp. 324–335, 2010.

[8] N. Chastan, M. C. Do, F. Bonneville et al., "Gait and balance disorders in Parkinson's disease: impaired active braking of the fall of centre of gravity," *Movement Disorders*, vol. 24, no. 2, pp. 188–195, 2009.

[9] L. Borel, C. Lopez, P. Péruch, and M. Lacour, "Vestibular syndrome: a change in internal spatial representation," *Neurophysiologie Clinique*, vol. 38, no. 6, pp. 375–389, 2008.

[10] J.-Y. Park, G. D. Park, S.-G. Lee, and J.-C. Lee, "The effect of scoliosis angle on center of gravity sway," *Journal of Physical Therapy Science*, vol. 25, no. 12, pp. 1629–1631, 2013.

[11] A. M. El-Kahky, H. Kingma, M. Dolmans, and I. de Jong, "Balance control near the limit of stability in various sensory conditions in healthy subjects and patients suffering from vertigo or balance disorders: impact of sensory input on balance control," *Acta Oto-Laryngologica*, vol. 120, no. 4, pp. 508–516, 2000.

[12] A. Patat, P. Foulhoux, M. Huchet, Y. Menin, and R. Venet, "Etude du comportement postural orthostatique d'une population de sujets agés athéroscléreux a partir de l'analyse spectrale de leurs stabilogrammes," *Agressologie*, vol. 26, no. 1, pp. 73–77, 1985.

[13] M. G. Montecchi, A. Muratori, F. Lombardi, E. Morrone, and K. Briant, "Trunk recovery scale: a new tool to measure posture control in patients with severe acquired brain injury. A study of the psychometric properties," *European Journal of Physical and Rehabilitation Medicine*, vol. 49, no. 3, pp. 341–351, 2013.

[14] M. Vikram, K. Sundaraganesh, M. Justine, M. Kurup, and J. H. Leonard, "Evaluation of postural control impairment using Balance Error Scoring System among athletes with ankle injury: an effective tool in daily clinical practice," *Clinica Terapeutica*, vol. 163, no. 5, pp. 383–386, 2012.

[15] B. Missaoui, P. Portero, S. Bendaya, O. Hanktie, and P. Thoumie, "Posture and equilibrium in orthopedic and rheumatologic diseases," *Neurophysiologie Clinique*, vol. 38, no. 6, pp. 447–457, 2008.

[16] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: preliminary development of an instrument," *Physiotherapy Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[17] K. O. Berg, S. L. Wood-Dauphinee, J. I. Williams, and B. Maki, "Measuring balance in the elderly: validation of an instrument," *Canadian Journal of Public Health*, vol. 83, supplement 2, no. 2, pp. 7–11, 1992.

[18] S. Mathias, U. S. L. Nayak, and B. Isaacs, "Balance in elderly patients: the 'get-up and go' test," *Archives of Physical Medicine and Rehabilitation*, vol. 67, no. 6, pp. 387–389, 1986.

[19] D. Podsiadlo and S. Richardson, "The timed "Up & Go": a test of basic functional mobility for frail elderly persons," *Journal of the American Geriatrics Society*, vol. 39, no. 2, pp. 142–148, 1991.

[20] M. E. Tinetti, D. I. Baker, G. McAvay et al., "A multifactorial intervention to reduce the risk of falling among elderly people living in the community," *The New England Journal of Medicine*, vol. 331, no. 13, pp. 821–827, 1994.

[21] J. M. Guralnik, E. M. Simonsick, L. Ferrucci et al., "A short physical performance battery assessing lower extremity function: association with self-reported disability and prediction of mortality and nursing home admission," *Journals of Gerontology*, vol. 49, no. 2, pp. 85–94, 1994.

[22] F. Franchignoni, F. Horak, M. Godi, A. Nardone, and A. Giordano, "Using psychometric techniques to improve the balance evaluation systems test: the mini-BESTest," *Journal of Rehabilitation Medicine*, vol. 42, no. 4, pp. 323–331, 2010.

[23] F. La Porta, M. Franceschini, S. Caselli, S. Susassi, P. Cavallini, and A. Tennant, "Unified Balance Scale: classic psychometric and clinical properties," *Journal of Rehabilitation Medicine*, vol. 43, no. 5, pp. 445–453, 2011.

[24] M. K. Holden, K. M. Gill, M. R. Magliozzi, J. Nathan, and L. Piehl-Baker, "Clinical gait assessment in the neurologically impaired: reliability and meaningfulness," *Physical Therapy*, vol. 64, no. 1, pp. 35–40, 1984.

[25] F. M. Collen, D. T. Wade, and C. M. Bradshaw, "Mobility after stroke: reliability of measures of impairment and disability," *International Disability Studies*, vol. 12, no. 1, pp. 6–9, 1990.

[26] C. Benaim, D. A. Pérennou, J. Villy, M. Rousseaux, and J. Y. Pelissier, "Validation of a standardized assessment of postural control in stroke patients: the Postural Assessment Scale for Stroke patients (PASS)," *Stroke*, vol. 30, no. 9, pp. 1862–1868, 1999.

[27] K. M. Sibley, T. Howe, S. E. Lamb et al., "Recommendations for a core outcome set for measuring standing balance in adult populations: a consensus-based approach," *PLoS ONE*, vol. 10, no. 3, Article ID e0120568, 2015.

[28] B. J. Vellas, S. J. Wayne, L. Romero, R. N. Baumgartner, L. Z. Rubenstein, and P. J. Garry, "One-leg balance is an important predictor of injurious falls in older persons," *Journal of the American Geriatrics Society*, vol. 45, no. 6, pp. 735–738, 1997.

[29] G. Abellan van Kan, Y. Rolland, S. Andrieu et al., "Gait speed at usual pace as a predictor of adverse outcomes in community-dwelling older people an International Academy on Nutrition and Aging (IANA) task force," *Journal of Nutrition, Health and Aging*, vol. 13, no. 10, pp. 881–889, 2009.

[30] G. Cimadoro, C. Paizis, G. Alberti, and N. Babault, "Effects of different unstable supports on EMG activity and balance," *Neuroscience Letters*, vol. 548, pp. 228–232, 2013.

[31] P. R. Rougier, M. Mathias, and A. Tanzi, "Short-term effects on postural control can be evidenced using a seesaw," *Neuroscience Letters*, vol. 488, no. 2, pp. 133–137, 2011.

[32] Y. P. Ivanenko, Y. S. Levik, V. L. Talis, and V. S. Gurfinkel, "Human equilibrium on unstable support: the importance of feet-support interaction," *Neuroscience Letters*, vol. 235, no. 3, pp. 109–112, 1997.

[33] S. K. Patel, M. L. Shende, and S. M. Khatri, "MFT a new diagnostic tool to check the balance in a normal healthy individuals," *IOSR Journal of Dental and Medical Sciences*, vol. 5, no. 6, pp. 14–18, 2013.

[34] L. M. Nashner, "Adapting reflexes controlling the human posture," *Experimental Brain Research*, vol. 26, no. 1, pp. 59–72, 1976.

[35] M. Duarte and S. M. De Freitas, "Revision of posturography based on force plate for balance evaluation," *Revista Brasileira de Fisioterapia*, vol. 14, no. 3, pp. 183–192, 2010.

[36] D. A. Winter and P. Eng, "Kinetics: our window into the goals and strategies of the central nervous system," *Behavioural Brain Research*, vol. 67, no. 2, pp. 111–120, 1995.

[37] O. Caron, T. Gelat, P. Rougier, and J.-P. Blanchi, "A comparative analysis of the center of gravity and center of pressure trajectory path lengths in standing posture: an estimation of active stiffness," *Journal of Applied Biomechanics*, vol. 16, no. 3, pp. 234–247, 2000.

[38] A. Huurnink, D. P. Fransz, I. Kingma, and J. H. van Dieën, "Comparison of a laboratory grade force platform with a Nintendo Wii Balance Board on measurement of postural control in single-leg stance balance tasks," *Journal of Biomechanics*, vol. 46, no. 7, pp. 1392–1395, 2013.

[39] L. Donath, R. Roth, L. Zahner, and O. Faude, "Testing single and double limb standing balance performance: comparison of COP path length evaluation between two devices," *Gait and Posture*, vol. 36, no. 3, pp. 439–443, 2012.

[40] F. Scoppa, R. Capra, M. Gallamini, and R. Shiffer, "Clinical stabilometry standardization: basic definitions—acquisition interval—sampling frequency," *Gait and Posture*, vol. 37, no. 2, pp. 290–292, 2013.

[41] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, John Wiley & Sons, Hoboken, NJ, USA, 2009.

[42] C. T. Bonnet, S. Cherraf, S. Szaffarczyk, and P. R. Rougier, "The contribution of body weight distribution and center of pressure location in the control of mediolateral stance," *Journal of Biomechanics*, vol. 47, no. 7, pp. 1603–1608, 2014.

[43] P. R. Rougier, "Relative contribution of the pressure variations under the feet and body weight distribution over both legs in the control of upright stance," *Journal of Biomechanics*, vol. 40, no. 11, pp. 2477–2482, 2007.

[44] F. Noé and F. Quaine, "Insertion of the force applied to handles into centre of pressure calculation modifies the amplitude of centre of pressure shifts," *Gait and Posture*, vol. 24, no. 3, pp. 382–385, 2006.

[45] T. Gélat, O. Caron, T. Metzger, and P. Rougier, "Human posturo-kinetic strategies from horizontal quadrupedal stance. Temporal and biomechanical aspects," *Gait and Posture*, vol. 2, no. 4, pp. 221–226, 1994.

[46] F. Quaine and L. Martin, "A biomechanical study of equilibrium in sport rock climbing," *Gait and Posture*, vol. 10, no. 3, pp. 233–239, 1999.

[47] S. A. Bergstra, B. Kluitenberg, R. Dekker et al., "Running with a minimalist shoe increases plantar pressure in the forefoot region of healthy female runners," *Journal of Science and Medicine in Sport*, vol. 18, pp. 463–468, 2015.

[48] H. Rice, M. Nunns, C. House, J. Fallowfield, A. Allsopp, and S. Dixon, "High medial plantar pressures during barefoot running are associated with increased risk of ankle inversion injury in Royal Marine recruits," *Gait and Posture*, vol. 38, no. 4, pp. 614–618, 2013.

[49] S. Angin, N. Ilçin, S. S. Yeşilyaprak, and I. E. Şimşek, "Prediction of postural sway velocity by foot posture index, foot size and plantar pressure values in unilateral stance," *Eklem Hastaliklari ve Cerrahisi*, vol. 24, no. 3, pp. 144–148, 2013.

[50] L. Baratto, P. G. Morasso, C. Re, and G. Spada, "A new look at posturographic analysis in the clinical context: sway-density versus other parameterization techniques," *Motor control*, vol. 6, no. 3, pp. 246–270, 2002.

[51] E. A. F. Ihlen, N. Skjæret, and B. Vereijken, "The influence of center-of-mass movements on the variation in the structure of human postural sway," *Journal of Biomechanics*, vol. 46, no. 3, pp. 484–490, 2013.

[52] C. K. Rhea, A. W. Kiefer, W. G. Wright, L. D. Raisbeck, and F. J. Haran, "Interpretation of postural control may change due to

data processing techniques," *Gait & Posture*, vol. 41, no. 2, pp. 731–735, 2015.

[53] J. J. Collins and C. J. De Luca, "Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories," *Experimental Brain Research*, vol. 95, no. 2, pp. 308–318, 1993.

[54] T. R. Han, N. J. Paik, and M. S. Im, "Quantification of the path of center of pressure (COP) using an F-scan in-shoe transducer," *Gait and Posture*, vol. 10, no. 3, pp. 248–254, 1999.

[55] F. Noé, D. Amarantini, and T. Paillard, "How experienced alpine-skiers cope with restrictions of ankle degrees-of-freedom when wearing ski-boots in postural exercises," *Journal of Electromyography and Kinesiology*, vol. 19, no. 2, pp. 341–346, 2009.

[56] T. Paillard, *Vieillissement et Condition Physique*, Editions Ellipses, Paris, France, 2009.

[57] F. Asseman, O. Caron, and J. Crémieux, "Is there a transfer of postural ability from specific to unspecific postures in elite gymnasts?" *Neuroscience Letters*, vol. 358, no. 2, pp. 83–86, 2004.

[58] P. Schubert and M. Kirchner, "Ellipse area calculations and their applicability in posturography," *Gait and Posture*, vol. 39, no. 1, pp. 518–522, 2014.

[59] M. Duarte, "Comments on "Ellipse area calculations and their applicability in posturography" (Schubert and Kirchner, vol. 39, pages 518–522, 2014)," *Gait and Posture*, vol. 41, no. 1, pp. 44–45, 2014.

[60] S. L. Pavão, G. S. Nunes, A. N. Santos, and N. A. C. F. Rocha, "Relationship between static postural control and the level of functional abilities in children with cerebral palsy," *Brazilian Journal of Physical Therapy*, vol. 18, no. 4, pp. 300–307, 2014.

[61] T. Paillard and F. Noé, "Effect of expertise and visual contribution on postural control in soccer," *Scandinavian Journal of Medicine and Science in Sports*, vol. 16, no. 5, pp. 345–348, 2006.

[62] T. E. Prieto, J. B. Myklebust, R. G. Hoffmann, E. G. Lovett, and B. M. Myklebust, "Measures of postural steadiness: differences between healthy young and elderly adults," *IEEE Transactions on Biomedical Engineering*, vol. 43, no. 9, pp. 956–966, 1996.

[63] J. A. Raymakers, M. M. Samson, and H. J. J. Verhaar, "The assessment of body sway and the choice of the stability parameter(s)," *Gait and Posture*, vol. 21, no. 1, pp. 48–58, 2005.

[64] K. Masani, A. H. Vette, M. O. Abe, and K. Nakazawa, "Center of pressure velocity reflects body acceleration rather than body velocity during quiet standing," *Gait and Posture*, vol. 39, no. 3, pp. 946–952, 2014.

[65] J. J. Vsetecková and N. Drey, "What is the role body sway deviation and body sway velocity play in postural stability in older adults?" *Acta Medica*, vol. 56, pp. 117–123, 2013.

[66] T. S. Kapteyn, W. Bles, C. J. Njiokiktjien, L. Kodde, C. H. Massen, and J. M. Mol, "Standardization in platform stabilometry being a part of posturography," *Agressologie*, vol. 24, no. 7, pp. 321–326, 1983.

[67] S. D. Lucy and K. C. Hayes, "Postural sway profiles. Normal subjects and subjects with cerebellar ataxia," *Physiotherapy Canada*, vol. 37, no. 3, pp. 140–148, 1985.

[68] D. Lafond, H. Corriveau, and F. Prince, "Postural control mechanisms during quiet standing in patients with diabetic sensory neuropathy," *Diabetes Care*, vol. 27, no. 1, pp. 173–178, 2004.

[69] R. B. Pachori, D. J. Hewson, H. Snoussi, and J. Duchêne, "Analysis of center of pressure signals using empirical mode decomposition and fourier-bessel expansion," in *Proceedings of the IEEE Region 10 Conference (TENCON '08)*, pp. 1–6, Hyderabad, India, November 2008.

[70] M. J. Warnica, T. B. Weaver, S. D. Prentice, and A. C. Laing, "The influence of ankle muscle activation on postural sway during quiet stance," *Gait and Posture*, vol. 39, no. 4, pp. 1115–1121, 2014.

[71] E. Golomer, P. Dupui, and P. Bessou, "Spectral frequency analysis of dynamic balance in healthy and injured athletes," *Archives of Physiology and Biochemistry*, vol. 102, no. 3, pp. 225–229, 1994.

[72] T. Paillard, C. Costes-Salon, C. Lafont, and P. Dupui, "Are there differences in postural regulation according to the level of competition in judoists?" *British Journal of Sports Medicine*, vol. 36, no. 4, pp. 304–305, 2002.

[73] T. Paillard, F. Noé, T. Rivière, V. Marion, R. Montoya, and P. Dupui, "Postural performance and strategy in the unipedal stance of soccer players at different levels of competition," *Journal of Athletic Training*, vol. 41, no. 2, pp. 172–176, 2006.

[74] T. Paillard, C. Lafont, M. C. Costes-Salon, D. Rivière, and P. Dupui, "Effects of brisk walking on static and dynamic balance, locomotion, body composition, and aerobic capacity in ageing healthy active men," *International Journal of Sports Medicine*, vol. 25, no. 7, pp. 539–546, 2004.

[75] J. R. Chagdes, S. Rietdyk, J. M. Haddad et al., "Multiple timescales in postural dynamics associated with vision and a secondary task are revealed by wavelet analysis," *Experimental Brain Research*, vol. 197, no. 3, pp. 297–310, 2009.

[76] J. W. Blaszczyk and W. Klonowski, "Postural stability and fractal dynamics," *Acta Neurobiologiae Experimentalis*, vol. 61, no. 2, pp. 105–112, 2001.

[77] M. Oliva Domínguez, J. Bartual Magro, J. Roquette Gaona, and J. Bartual Pastor, "Spectrum analysis in postural strategy on static tests in a healthy population," *Acta Otorrinolaringologica Espanola*, vol. 64, no. 2, pp. 124–132, 2013.

[78] K. M. Newell, S. M. Slobounov, E. S. Slobounova, and P. C. M. Molenaar, "Stochastic processes in postural center-of-pressure profiles," *Experimental Brain Research*, vol. 113, no. 1, pp. 158–164, 1997.

[79] R. J. Peterka, "Postural control model interpretation of stabilogram diffusion analysis," *Biological Cybernetics*, vol. 82, no. 4, pp. 335–343, 2000.

[80] V. M. Zatsiorsky and M. Duarte, "Instant equilibrium point and its migration in standing tasks: rambling and trembling components of the stabilogram," *Motor Control*, vol. 3, no. 1, pp. 28–38, 1999.

[81] A. G. Feldman and M. F. Levin, "The equilibrium-point hypothesis-past, present and future," *Advances in Experimental Medicine and Biology*, vol. 629, pp. 699–726, 2009.

[82] V. M. Zatsiorsky and M. Duarte, "Rambling and trembling in quiet standing," *Motor Control*, vol. 4, no. 2, pp. 185–200, 2000.

[83] V. Cimolin, M. Galli, C. Rigoldi et al., "The fractal dimension approach in posture: a comparison between Down and Prader-Willi syndrome patients," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 17, no. 14, pp. 1535–1541, 2014.

[84] Z. Mei, G. Zhao, K. Ivanov et al., "Sample entropy characteristics of movement for four foot types based on plantar centre of pressure during stance phase," *BioMedical Engineering Online*, vol. 12, article 101, 2013.

[85] C. M. Hunt, G. Widener, and D. D. Allen, "Variability in postural control with and without balance-based torso-weighting in people with multiple sclerosis and healthy controls," *Physical Therapy*, vol. 94, no. 10, pp. 1489–1498, 2014.

[86] J. M. Huisinga, J. M. Yentes, M. L. Filipi, and N. Stergiou, "Postural control strategy during standing is altered in patients with multiple sclerosis," *Neuroscience Letters*, vol. 524, no. 2, pp. 124–128, 2012.

[87] J. T. Cavanaugh, K. M. Guskiewicz, and N. Stergiou, "A nonlinear dynamic approach for evaluating postural control: new directions for the management of sport-related cerebral concussion," *Sports Medicine*, vol. 35, no. 11, pp. 935–950, 2005.

[88] M. Duarte and V. M. Zatsiorsky, "Patterns of center of presure migration during prolonged unconstrained standing," *Motor Control*, vol. 3, no. 1, pp. 12–27, 1999.

[89] J. Baltich, V. von Tscharner, P. Zandiyeh, and B. M. Nigg, "Quantification and reliability of center of pressure movement during balance tasks of varying difficulty," *Gait and Posture*, vol. 40, no. 2, pp. 327–332, 2014.

[90] V. Agostini, E. Chiaramello, and M. Knaflitz, "Circular components in center of pressure signals," *Motor Control*, vol. 17, no. 4, pp. 355–369, 2013.

[91] M. Goffredo, M. Schmid, S. Conforto, and T. D'Alessio, "A markerless sub-pixel motion estimation technique to reconstruct kinematics and estimate the centre of mass in posturography," *Medical Engineering and Physics*, vol. 28, no. 7, pp. 719–726, 2006.

[92] M. Günther, S. Grimmer, T. Siebert, and R. Blickhan, "All leg joints contribute to quiet human stance: a mechanical analysis," *Journal of Biomechanics*, vol. 42, no. 16, pp. 2739–2746, 2009.

[93] M. Mancini, A. Salarian, P. Carlson-Kuhta et al., "ISway: a sensitive, valid and reliable measure of postural control," *Journal of NeuroEngineering and Rehabilitation*, vol. 9, article 59, 2012.

[94] B. Najafi, D. Horn, S. Marclay, R. T. Crews, S. Wu, and J. S. Wrobel, "Assessing postural control and postural control strategy in diabetes patients using innovative and wearable technology," *Journal of Diabetes Science and Technology*, vol. 4, no. 4, pp. 780–791, 2010.

[95] O. Oullier, B. G. Bardy, T. A. Stoffregen, and R. J. Bootsma, "Postural coordination in looking and tracking tasks," *Human Movement Science*, vol. 21, no. 2, pp. 147–167, 2002.

[96] Y. Aramaki, D. Nozaki, K. Masani, T. Sato, K. Nakazawa, and H. Yano, "Reciprocal angular acceleration of the ankle and hip joints during quiet standing in humans," *Experimental Brain Research*, vol. 136, no. 1, pp. 463–473, 2001.

[97] S. Sasagawa, J. Ushiyama, M. Kouzaki, and H. Kanehisa, "Effect of the hip motion on the body kinematics in the sagittal plane during human quiet standing," *Neuroscience Letters*, vol. 450, no. 1, pp. 27–31, 2009.

[98] K. O'Brien, E. Culham, and B. Pickles, "Balance and skeletal alignment in a group of elderly female fallers and nonfallers," *Journals of Gerontology A: Biological Sciences and Medical Sciences*, vol. 52, no. 4, pp. B221–B226, 1997.

[99] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-perczak, "Stiffness control of balance in quiet standing," *Journal of Neurophysiology*, vol. 80, no. 3, pp. 1211–1221, 1998.

[100] S.-C. Chen, H.-J. Hsieh, T.-W. Lu, and C.-H. Tseng, "A method for estimating subject-specific body segment inertial parameters in human movement analysis," *Gait and Posture*, vol. 33, no. 4, pp. 695–700, 2011.

[101] P. Federolf, L. Roos, and B. M. Nigg, "Analysis of the multi-segmental postural movement strategies utilized in bipedal, tandem and one-leg stance as quantified by a principal component decomposition of marker coordinates," *Journal of Biomechanics*, vol. 46, no. 15, pp. 2626–2633, 2013.

[102] M. C. Kilby, P. C. M. Molenaar, and K. M. Newell, "Models of postural control: shared variance in joint and COM motions," *PLoS ONE*, vol. 10, no. 5, Article ID e0126379, 2015.

[103] Z. Wang, J. H. Ko, J. H. Challis, and K. M. Newell, "The degrees of freedom problem in human standing posture: collective and component dynamics," *PLoS ONE*, vol. 9, no. 1, Article ID e85414, 2014.

[104] B. G. Bardy, L. Marin, T. A. Stoffregen, and R. J. Bootsma, "Postural coordination modes considered as emergent phenomena," *Journal of Experimental Psychology: Human Perception and Performance*, vol. 25, no. 5, pp. 1284–1301, 1999.

[105] H. Zhang, M. A. Nussbaum, and M. J. Agnew, "Use of wavelet coherence to assess two-joint coordination during quiet upright stance," *Journal of Electromyography and Kinesiology*, vol. 24, no. 5, pp. 607–613, 2014.

[106] M.-C. Guo, J.-F. Yang, C.-T. Huang, and I.-S. Hwang, "Organization of physiological tremors and coordination solutions to postural pointing on an uneven stance surface," *Journal of Electromyography and Kinesiology*, vol. 22, no. 4, pp. 589–597, 2012.

[107] R. Merletti and P. A. Parker, *Electromyography: Physiology, Engineering, and Non-Invasive Applications*, John Wiley, Hoboken, NJ, USA, 2004.

[108] H. Saito, M. Yamanaka, S. Kasahara, and J. Fukushima, "Relationship between improvements in motor performance and changes in anticipatory postural adjustments during whole-body reaching training," *Human Movement Science*, vol. 37, pp. 69–86, 2014.

[109] F. B. Horak and L. M. Nashner, "Central programming of postural movements: adaptation to altered support-surface configurations," *Journal of Neurophysiology*, vol. 55, no. 6, pp. 1369–1381, 1986.

[110] K. Fujiwara, H. Toyama, T. Kiyota, and K. Maeda, "Postural muscle activity patterns during standing at rest and on an oscillating floor," *Journal of Electromyography and Kinesiology*, vol. 16, no. 5, pp. 448–457, 2006.

[111] P. Gatev, S. Thomas, T. Kepple, and M. Hallett, "Feedforward ankle strategy of balance during quiet stance in adults," *Journal of Physiology*, vol. 514, no. 3, pp. 915–928, 1999.

[112] M. Saffer, T. Kiemel, and J. Jeka, "Coherence analysis of muscle activity during quiet stance," *Experimental Brain Research*, vol. 185, no. 2, pp. 215–226, 2008.

[113] M. Mehdikhani, N. Khalaj, T. Y. Chung, and M. Mazlan, "The effect of feet position on standing balance in patients with diabetes," *Proceedings of the Institution of Mechanical Engineers Part H: Journal of Engineering in Medicine*, vol. 228, no. 8, pp. 819–823, 2014.

[114] J.-W. Kim, Y. Kwon, H.-M. Jeon et al., "Feet distance and static postural balance: implication on the role of natural stance," *Bio-Medical Materials and Engineering*, vol. 24, no. 6, pp. 2681–2688, 2014.

[115] M. Milosevic, K. M. V. McConville, and K. Masani, "Arm movement improves performance in clinical balance and mobility tests," *Gait and Posture*, vol. 33, no. 3, pp. 507–509, 2011.

[116] M. G. Carpenter, J. S. Frank, D. A. Winter, and G. W. Peysar, "Sampling duration effects on centre of pressure summary measures," *Gait and Posture*, vol. 13, no. 1, pp. 35–40, 2001.

[117] T. Paillard, R. Bizid, and P. Dupui, "Do sensorial manipulations affect subjects differently depending on their postural abilities?" *British Journal of Sports Medicine*, vol. 41, no. 7, pp. 435–438, 2007.

[118] J. Maitre, I. Serres, L. Lhuisset, J. Bois, Y. Gasnier, and T. Paillard, "Regular physical activity reduces the effects of Achilles tendon vibration on postural control for older women," *Scandinavian Journal of Medicine and Science in Sports*, vol. 25, no. 1, pp. e82–e88, 2015.

[119] J. Maitre, Y. Gasnier, N. Bru, J.-L. Jully, and T. Paillard, "Discrepancy in the involution of the different neural loops with age," *European Journal of Applied Physiology*, vol. 113, no. 7, pp. 1821–1831, 2013.

[120] J. Maitre, J.-L. Jully, Y. Gasnier, and T. Paillard, "Chronic physical activity preserves efficiency of proprioception in postural control in older women," *Journal of Rehabilitation Research and Development*, vol. 50, no. 6, pp. 811–820, 2013.

[121] A. M. F. Barela, J. A. Barelá, N. M. Rlnaldi, and D. R. de Toledo, "Influence of imposed optic flow characteristics and intention on postural responses," *Motor Control*, vol. 13, no. 2, pp. 119–129, 2009.

[122] Y. Mizuno, M. Shindo, S. Kuno, T. Kawakita, and S. Watanabe, "Postural control responses sitting on unstable board during visual stimulation," *Acta Astronautica*, vol. 49, no. 3–10, pp. 131–136, 2001.

[123] I. S. Curthoys, "A critical review of the neurophysiological evidence underlying clinical vestibular testing using sound, vibration and galvanic stimuli," *Clinical Neurophysiology*, vol. 121, no. 2, pp. 132–144, 2010.

[124] R. C. Fitzpatrick and B. L. Day, "Probing the human vestibular system with galvanic stimulation," *Journal of Applied Physiology*, vol. 96, no. 6, pp. 2301–2316, 2004.

[125] Y. Yang, F. Pu, X. Lv et al., "Comparison of postural responses to galvanic vestibular stimulation between pilots and the general populace," *BioMed Research International*, vol. 2015, Article ID 567690, 6 pages, 2015.

[126] D. L. Wardman, B. L. Day, and R. C. Fitzpatrick, "Position and velocity responses to galvanic vestibular stimulation in human subjects during standing," *Journal of Physiology*, vol. 547, no. 1, pp. 293–299, 2003.

[127] D. L. Wardman, J. L. Taylor, and R. C. Fitzpatrick, "Effects of galvanic vestibular stimulation on human posture and perception while standing," *The Journal of Physiology*, vol. 551, no. 3, pp. 1033–1042, 2003.

[128] A. Séverac Cauquil, P. Bousquet, M.-C. Costes Salon, P. Dupui, and P. Bessou, "Monaural and binaural galvanic vestibular stimulation in human dynamic balance function," *Gait and Posture*, vol. 6, no. 3, pp. 210–217, 1997.

[129] A. S. Cauquil, M. F. T. Gervet, and M. Ouaknine, "Body response to binaural monopolar galvanic vestibular stimulation in humans," *Neuroscience Letters*, vol. 245, no. 1, pp. 37–40, 1998.

[130] B. L. Day, A. Séverac Cauquil, L. Bartolomei, M. A. Pastor, and I. N. Lyon, "Human body-segment tilts induced by galvanic stimulation: a vestibularly driven balance protection mechanism," *Journal of Physiology*, vol. 500, no. 3, pp. 661–672, 1997.

[131] D. L. Wardman and R. C. Fitzpatrick, "What does galvanic vestibular stimulation stimulate?" *Advances in Experimental Medicine and Biology*, vol. 508, pp. 119–128, 2002.

[132] I. S. Curthoys and H. G. MacDougall, "What galvanic vestibular stimulation actually activates," *Frontiers in Neurology*, vol. 3, article 117, 2012.

[133] R. L. Carvalho and G. L. Almeida, "The effect of galvanic vestibular stimulation on postural response of Down syndrome individuals on the seesaw," *Research in Developmental Disabilities*, vol. 32, no. 5, pp. 1542–1547, 2011.

[134] J. P. Roll and J. P. Vedel, "Kinaesthetic role of muscle afferents in man, studied by tendon vibration and microneurography," *Experimental Brain Research*, vol. 47, no. 2, pp. 177–190, 1982.

[135] J. P. Roll, J. P. Vedel, and E. Ribot, "Alteration of proprioceptice messages induced by tendon vibration in man: a microneurographic study," *Experimental Brain Research*, vol. 76, no. 1, pp. 213–222, 1989.

[136] H. Ceyte, C. Cian, R. Zory, P.-A. Barraud, A. Roux, and M. Guerraz, "Effect of Achilles tendon vibration on postural orientation," *Neuroscience Letters*, vol. 416, no. 1, pp. 71–75, 2007.

[137] C. Thompson, M. Bélanger, and J. Fung, "Effects of bilateral Achilles tendon vibration on postural orientation and balance during standing," *Clinical Neurophysiology*, vol. 118, no. 11, pp. 2456–2467, 2007.

[138] A. Polónyová and F. Hlavacka, "Human postural responses to different frequency vibrations of lower leg muscles," *Physiological Research*, vol. 50, no. 4, pp. 405–410, 2001.

[139] M. Billot, G. A. Handrigan, M. Simoneau, and N. Teasdale, "Reduced plantar sole sensitivity induces balance control modifications to compensate ankle tendon vibration and vision deprivation," *Journal of Electromyography and Kinesiology*, vol. 25, no. 1, pp. 155–160, 2015.

[140] M. Bove, A. Nardone, and M. Schieppati, "Effects of leg muscle tendon vibration on group Ia and group II reflex responses to stance perturbation in humans," *Journal of Physiology*, vol. 550, no. 2, pp. 617–630, 2003.

[141] N. Deshpande and A. E. Patla, "Postural responses and spatial orientation to neck proprioceptive and vestibular inputs during locomotion in young and older adults," *Experimental Brain Research*, vol. 167, no. 3, pp. 468–474, 2005.

[142] M. Piecha, G. Juras, P. Król, G. Sobota, A. Polak, and B. Bacik, "The effect of a short-term and long-term whole-body vibration in healthy men upon the postural stability," *PLoS ONE*, vol. 9, no. 2, Article ID e88295, 2014.

[143] M. A. Hoffman and D. M. Koceja, "Dynamic balance testing with electrically evoked perturbation: a test of reliability," *Archives of Physical Medicine and Rehabilitation*, vol. 78, no. 3, pp. 290–293, 1997.

[144] F. Stål, P. A. Fransson, M. Magnusson, and M. Karlberg, "Effects of hypothermic anesthesia of the feet on vibration-induced body sway and adaptation," *Journal of Vestibular Research: Equilibrium and Orientation*, vol. 13, no. 1, pp. 39–52, 2003.

[145] M. Patel, P. A. Fransson, R. Johansson, and M. Magnusson, "Foam posturography: standing on foam is not equivalent to standing with decreased rapidly adapting mechanoreceptive sensation," *Experimental Brain Research*, vol. 208, no. 4, pp. 519–527, 2011.

[146] C. Fujimoto, T. Murofushi, Y. Chihara et al., "Assessment of diagnostic accuracy of foam posturography for peripheral vestibular disorders: analysis of parameters related to visual and somatosensory dependence," *Clinical Neurophysiology*, vol. 120, no. 7, pp. 1408–1414, 2009.

[147] J.-H. Chiang and G. Wu, "The influence of foam surfaces on biomechanical variables contributing to postural control," *Gait & Posture*, vol. 5, no. 3, pp. 239–245, 1997.

[148] S. Demura, S. Yamaji, T. Kitabayashi, T. Yamada, and M. Uchiyama, "Attention of postural control on foot somatosensor disturbance caused by the compression of blood vessels," *Journal of Human Ergology*, vol. 37, no. 2, pp. 91–102, 2008.

[149] H. C. Diener, J. Dichgans, B. Guschlbauer, and H. Mau, "The significance of proprioception on postural stabilization as

assessed by ischemia," *Brain Research*, vol. 296, no. 1, pp. 103–109, 1984.

[150] K.-H. Mauritz and V. Dietz, "Characteristics of postural instability induced by ischemic blocking of leg afferents," *Experimental Brain Research*, vol. 38, no. 1, pp. 117–119, 1980.

[151] P. Thoumie and M. C. Do, "Changes in motor activity and biomechanics during balance recovery following cutaneous and muscular deafferentation," *Experimental Brain Research*, vol. 110, no. 2, pp. 289–297, 1996.

[152] E. V. Fraizer and S. Mitra, "Methodological and interpretive issues in posture-cognition dual-tasking in upright stance," *Gait and Posture*, vol. 27, no. 2, pp. 271–279, 2008.

[153] A. C. H. Geurts, T. W. Mulder, B. Nienhuis, and R. A. J. Rijken, "Dual-task assessment of reorganization of postural control in persons with lower limb amputation," *Archives of Physical Medicine and Rehabilitation*, vol. 72, no. 13, pp. 1059–1064, 1991.

[154] Y. Lajoie, N. Teasdale, C. Bard, and M. Fleury, "Attentional demands for static and dynamic equilibrium," *Experimental Brain Research*, vol. 97, no. 1, pp. 139–144, 1993.

[155] A. Shumway-Cook and M. Woollacott, "Attentional demands and postural control: the effect of sensory context," *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, vol. 55, no. 1, pp. M10–M16, 2000.

[156] L. Yardley, M. Gardner, A. Bronstein, R. Davies, D. Buckwell, and L. Luxon, "Interference between postural control and mental task performance in patients with vestibular disorder and healthy controls," *Journal of Neurology Neurosurgery and Psychiatry*, vol. 71, no. 1, pp. 48–52, 2001.

[157] M. Woollacott and A. Shumway-Cook, "Attention and the control of posture and gait: a review of an emerging area of research," *Gait and Posture*, vol. 16, no. 1, pp. 1–14, 2002.

[158] S. Schaefer, "The ecological approach to cognitive-motor dual-tasking: findings on the effects of expertise and age," *Frontiers in Psychology*, vol. 5, article 1167, 2014.

[159] G. Ebersbach, M. R. Dimitrijevic, and W. Poewe, "Influence of concurrent tasks on gait: a dual-task approach," *Perceptual and Motor Skills*, vol. 81, no. 1, pp. 107–113, 1995.

[160] N. Teasdale, C. Bard, J. LaRue, and M. Fleury, "On the cognitive penetrability of posture control," *Experimental Aging Research*, vol. 19, no. 1, pp. 1–13, 1993.

[161] S. Gobbo, M. Bergamin, J. C. Sieverdes, A. Ermolao, and M. Zaccaria, "Effects of exercise on dual-task ability and balance in older adults: a systematic review," *Archives of Gerontology and Geriatrics*, vol. 58, no. 2, pp. 177–187, 2014.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 309386, 7 pages
http://dx.doi.org/10.1155/2015/309386

Research Article

Effects of Fourteen-Day Bed Rest on Trunk Stabilizing Functions in Aging Adults

Nejc Sarabon[1,2,3] and Jernej Rosker[2,3]

[1]*Department for Health Study, Andrej Marušič Institute, University of Primorska, Muzejski Trg 2, SI-6000 Koper, Slovenia*
[2]*Faculty of Mathematics, Natural Sciences and Information Technologies, University of Primorska, Glagoljaska 8, SI-6000 Koper, Slovenia*
[3]*S2P Ltd., Laboratory for Motor Control and Motor Behaviour, Tehnoloski Park 19, SI-1000 Ljubljana, Slovenia*

Correspondence should be addressed to Nejc Sarabon; nejc.sarabon@s2p.si

Received 24 April 2015; Revised 15 June 2015; Accepted 1 July 2015

Academic Editor: Thierry Paillard

Bed rest has been shown to have detrimental effects on structural and functional characteristics of the trunk muscles, possibly affecting trunk and spinal stability. This is especially important in populations such as aging adults with often altered trunk stabilizing functions. This study examined the effects of a fourteen-day bed rest on anticipatory postural adjustments and postural reflex responses of the abdominal wall and back muscles in sixteen adult men. Postural activation of trunk muscles was measured using voluntary quick arm movement and sudden arm loading paradigm. Measurements were conducted prior to the bed rest, immediately after, and fourteen days after the bed rest. Immediately after the bed rest, latencies of anticipatory postural adjustments showed significant shortening, especially for the obliquus internus and externus muscles. After a fourteen-day recuperation period, anticipatory postural adjustments reached a near to complete recovery. On the contrary, reactive response latencies increased from pre-bed-rest to both post-bed-rest measurement sessions. Results indicate an important effect of bed rest on stabilizing functions of the trunk muscles in elderly adults. Moreover, there proved to be a significant deterioration of postural reactive responses that outlasted the 14-day post-bed-rest rehabilitation.

1. Introduction

Trunk and spinal stability have been proposed as an important factor for preventing spinal injuries and disease [1, 2]. Passive (i.e., skeletal and connective tissue system), active (i.e., muscular system), and neural (i.e., central and peripheral nervous system) systems have been proposed to be fundamental for preserving posture and stabilizing the spine [3]. Hoffman and Gabel [2] extended this model arguing that a stable spine is important for effective and injury-free movements of the limbs. For more than twenty years different training intervention strategies have emerged, focusing on improving spinal and trunk stability [4, 5]. On the other hand, new insights into changes in spinal and trunk stability following injury or disease are available. However, much less is known about the alterations in spinal and trunk stability following long term inactivity (i.e., bed ridden medical intervention) [6]. This is especially important for more vulnerable populations such as aging adults who are more prone to medical issues demanding hospitalisation [6, 7].

For studying the ability of the trunk to maintain stability during anticipated perturbations, a quick arm movement paradigm has been used. In this type of movement the neural controlling centres anticipate the extent to which posture will be perturbed and initiate compensatory actions prior to the perturbation onset (i.e., anticipatory postural adjustments or APAs). As in conditions of unanticipated perturbations, the sudden arm loading paradigm has been used. These responses to perturbation are primarily reflexive/reactive in their nature (i.e., reflex responses or RRs). The magnitude of stability loss has been quantified using biomechanical parameters (i.e., centre of body mass (CoM) and centre of pressure (CoP)) [8] and electromyographic parameters (i.e., latency and magnitude of the EMG) [9, 10]. The electromyographic responses have been shown to be correlated with the CoP amplitude as muscle activation tunes the body and the trunk stiffness consequently affecting the body sway measured via CoP and CoM [8, 10].

Changes in postural and trunk-spinal stability have been well documented in aging adults [10–12]. Their reactions to postural perturbations change from distal to proximal, suggesting an increased dependence on the hip strategy [13]. Moreover, the RRs of the lower limb and trunk muscles to unanticipated perturbations are delayed when compared to their younger peers [14]. The APAs have been shown to be altered but generally preserved, suggesting less robust stabilization strategies [12]. As suggested by the literature more profound changes have been observed for RRs than for APAs [14].

In addition to the functional changes, aging adults often suffer from medical issues demanding prolonged bed rest (BR). To our knowledge, no studies have been done that have addressed the effects of prolonged BR on the neuromuscular aspects of spinal (i.e., trunk) stability. The majority of the BR studies have been completed on young healthy adults indicating significant anatomical changes (i.e., decrease in trunk muscle cross-sectional area [15–17] and changes in spinal morphology [18]) and functional changes (shift from tonic to phasic activity of the lower back muscles [19, 20] and changes in body sway [21]). These changes have been shown to persist for at least 56 days after a prolonged BR [16, 17]. These findings suggest that the active and neural subsystems for spine stabilization could be affected by prolonged BR and cause changes in APAs and RRs.

The purpose of this study was to assess changes in APAs and RRs latencies of the trunk stabilizing muscles following a fourteen-day BR in aging adults. In addition we wanted to assess the recuperation of APAs and RRs after a fourteen-day-long post-BR phase (including a training intervention). Our first hypothesis states that the onset of APAs will be closer to the activation of the prime movement muscle and RRs would be significantly delayed following the fourteen days of BR. Our second hypothesis states that in the post-BR recuperation phase these changes would reverse to pre-bed-rest levels for both APAs and RRs.

2. Materials and Methods

2.1. Participants. Sixteen healthy men ([mean (standard error)] of 59.6 (3.4) years and with body height of 173.3 (4.9) cm, body mass 77.3 (11.8) kg, and BMI 25.8 (3.7) kg/m^2) that were recruited from the local community volunteered in the study. Prior to the enrolment, each participant underwent a medical evaluation. The following were used as exclusion criteria: diabetes, active malignancy, uncontrolled hypertension, history of cardiovascular disease, history of deep vein thrombosis/pulmonary embolism, significant hepatic or renal disease, chronic inflammatory disease, any significant impairment of the locomotor system, and vestibular or uncorrected visual disturbance. Prior to the enrolment each participant was informed of the protocol and potential risks of the study and was required to sign a written informed consent, confirmed by the Slovenian National Committee for Medical Ethics. All procedures were in accordance with the Declaration of Helsinki and Oviedo Convention.

2.2. Study Protocol. During the 14 days of horizontal BR without physical countermeasures, participants were required to restrict physical activity and reduce deviations from the horizontal lying position to a minimum, also during showering and toileting. Following BR, participants completed 2 weeks of a rehabilitation protocol (three times per week). A typical session consisted of a warm-up (6 min of low intensity Nordic walking and 6 min of active stretching), main part (4 balance exercises, 6 strength exercises, and Nordic walking endurance protocol), and cooldown (simple breathing exercises). For more information on the Valdoltra BR study and rehabilitation protocol the reader is directed to the work by Goswami and colleagues [22]. Follow-up measurements were performed 14 days after the end of the BR period. Participants were evaluated (i) before the BR (PRE), (ii) immediately after the BR (POST0), and, finally, (iii) 14 days after the BR (POST14). During the assessment protocol, participants were evaluated with the measurement protocol for the evaluation of trunk postural (pre/re)actions as described below.

2.3. Measurement Tasks and Procedures. Each of the evaluations consisted of two tests: measurements of APAs and measurements of RRs to sudden loading. At the beginning, a 5-minute standardized warm-up was performed (spot running with high knees 2.5 min, 10 squats, and 10 push-ups with hands supported on the wall). Before undertaking each test, participants performed five introductory trials. Participants were barefoot and asked to place their feet at the hip-width during all of the measurements. Participants were constantly reminded to maintain their normal posture and to stand relaxed. All trials were triggered in a random manner every 5 to 12 s with 24 repetitions in total for each test (3 sets × 8 repetitions, 1 min breaks).

Measurements of APAs were performed on a random visual cue (LED light, eyes high, and 1.5 m distance). The participant stood with his/her arms extended down by their sides and upon the visual signal the task was to raise a 1.2 kg bar as fast as possible with straight arms up to shoulder height, hold the position for ~1 s, and return the bar back down to the starting position slowly (Figure 1(a)) [23, 24]. In *measurements of RRs to sudden loading* participants stood relaxed, with their elbows flexed to 90° and palms slightly touching the weight handle (8% of the individual's body mass, for more detail see Figure 1(b)) [25]. A sudden release of the load was achieved by a custom built electromagnetic mechanism. After load release, the participants' task was to return to and settle at the initial position, as quickly as possible.

The setup was controlled by bespoke software (Labview 2012, National Instruments, Texas, USA), which triggered the visual clues and the quick release mechanism. Triggering was synchronized with an electrocardiogram (QRS-wave + 200 ms) so that the postural activation of the trunk muscles appeared during two consecutive QRS-peaks [26]. This prevented ECG interference [27], which could hamper the analysis of the electromyographic (EMG) signals.

2.4. Electromyography. The activity of five trunk muscles was acquired with surface EMG. Signals were 3,000x amplified

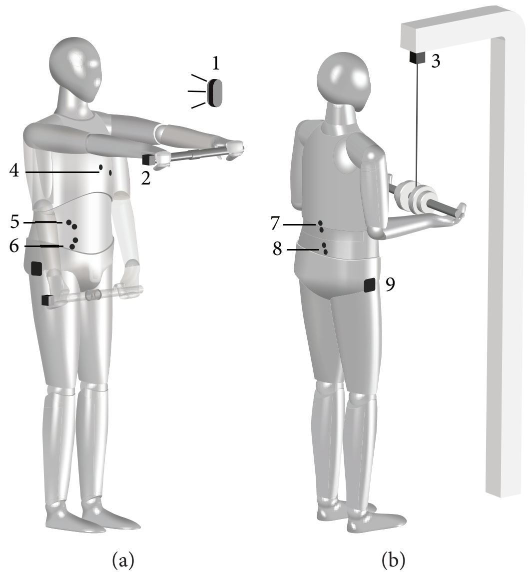


Figure 1: Presentation of fast bilateral shoulder flexion (a) and sudden arm loading task (b). A detailed positioning of the visual cue (1), accelerometer (2), load release system (3), electrocardiogram electrodes (4) and EMG electrodes for obliquus externus (5), muscle obliquus internus (6), muscle erector spine (7), muscle multifidus (8), and reference electrode (9) is presented.

(Biovision, Wehrheim, Germany), A/D converted, and sampled at 10,000 Hz (USB-6343, National Instruments, Texas, USA). Self-adhesive pairs of electrodes (Blue Sensor N, Ambu A/S, Ballerup, Denmark) were used, placed with 2 cm center-to-center distance. EMG of obliquus externus (EO), multifidus at level L5 (MU), erector spine at level L1 (LE), obliquus internus (OI), and deltoideus anterior (DE) were recorded on the right side of the body. Skin preparation and electrode placement were done according to SENIAM recommendations [28]. As this standard does not include recommendations for abdominal muscles, we followed the electrode placement from previous studies with regard to the external and internal oblique muscles [29–32]. An additional pair of electrodes was placed for ECG detection, with one electrode in the region of xiphoid process of sternum and second on the 1/3 of the left ribcage arc. A reference electrode was positioned on the area of the right greater trochanter.

2.5. Signal Processing. Signals were band-pass filtered (zero lag Butterworth filter 10 Hz/1 kHz, order 2), rectified using a root mean square smoothing filter (window of 20 ms), and low-pass filtered (zero lag Butterworth filter 10 Hz, order 2) to get a linear envelope. Approximated generalized likelihood-ratio step algorithm (AGLRstep) [33] for automatic detection was used to determine the beginning of muscle activity. In APA measurements, activation onset detection was limited with a time window from 200 ms before to 50 ms after the activation of the prime mover muscle (being the DE) [34, 35]. APAs onset times were calculated as the difference between the onset of trunk muscle activation and the activation of the DE muscle. When the activation of the postural muscle preceded the hand movement initiation, the value was negative. In the RRs measurements the activation onset detection was limited to the time window from the moment of the mechanism release (t_0) to 200 ms after (t_{end}) (Figure 2) [36]. RRs onset times were calculated as a delay from the mechanism release to the trunk muscle activation.

Each muscle was considered active when the processed EMG signal exceeded the average plus 2 standard deviations of the baseline EMG signal [37]. Baseline EMG signal was calculated from the 50 ms reference window directly before the activation onset detection limits, where there was no task related activation or ECG artefacts [37]. All signals were later manually inspected and corrected when activation was incorrectly detected. When there was no activation of the investigated muscle, or activation was not within the detection limits, the trial was not used in further analysis.

2.6. Statistical Analysis. Statistical analyses (SPSS 18.0 software, SPSS Inc., Chicago, USA) were performed as follows. Descriptive statistics were calculated for all variables and reported as *mean (standard errors)*. One-way repeated-measures ANOVA were run to test the differences between the testing sessions (PRE, POST0, and POST14). Two-tailed pairwise t-tests with Bonferroni corrections were used for pairwise comparisons. The level of statistical significance (p) was set at 0.05 and effect size (ES) values were calculated.

3. Results

From 24 trials in each task, the following number of trials were removed due to an unrecognizable response (responses not exceeding 2 standard deviations of the baseline): (1) for anticipation task EO: [mean (standard error)] 3.8 (2.4), IO: 3.7 (2.3), LE: 3.6 (2.5), and MU: 3.3 (2.7) and (2) for reaction task EO: 2.1 (2.0), IO: 2.6 (2.6), LE: 2.1 (1.8), and MU: 2.0 (2.0). Latencies in the anticipation task ranged from −22 to −5 ms for the back muscles (LE and MU) and from −3 to 22 ms for

Table 1: One-way repeated-measures ANOVA and corresponding post hoc t-tests for both tests of automatic (re)actions of the selected trunk muscles (obliquus externus (EO), obliquus internus (OI), lumbar erector spine (LE), and multifidus (MU)). Statistical significance of differences (p) and effect sizes (ES) are presented.

Muscle	RANOVA PRE-POST0-POST14			Post hoc t-tests PRE-POST0			PRE-POST14			POST0-POST14		
	F	p	ES	t	p	ES	t	p	ES	t	p	ES
Trunk anticipatory postural adjustments												
EO	1.874	.175	.135	1.139	.277	.312	1.773	.102	.456	0.782	.449	.220
IO	1.126	.341	.086	0.383	.708	.110	1.169	.265	.320	1.944	.076	.489
LE	3.761	.038	.239	0.931	.370	.259	2.445	.031	.577	2.223	.046	.540
MU	4.553	.021	.275	1.320	.211	.356	3.030	.010	.658	1.786	.099	.458
Trunk reflex reactions on mechanical perturbation												
EO	5.560	.015	.410	−1.994	.081	.576	−3.438	.009	.772	−1.172	.275	.383
IO	7.193	.005	.444	−3.610	.006	.769	−2.707	.024	.670	0.306	.766	.102
LE	10.439	.001	.487	−4.304	.001	.792	−2.963	.013	.666	1.496	.163	.411
MU	7.860	.003	.440	−3.501	.006	.742	−2.272	.046	.584	1.934	.082	.522

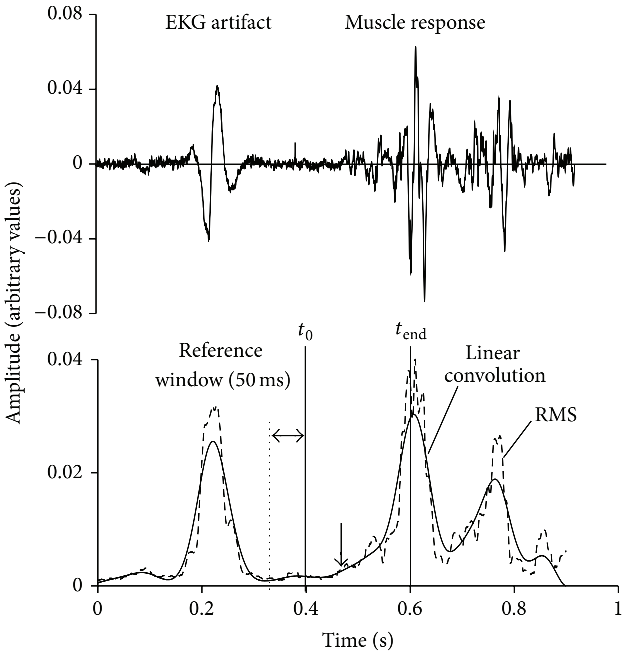


Figure 2: An example of a postural reflex response to sudden loading (muscle obliquus abdominis externus). Instance of load drop (t_0): 200 ms window for analysis (t_{end}) and 50 ms reference window for calculating baseline EMG activation are shown. The arrow depicts the muscle activation onset, solid line depicts the linear convolution, and the dotted line depicts the root mean square (RMS).

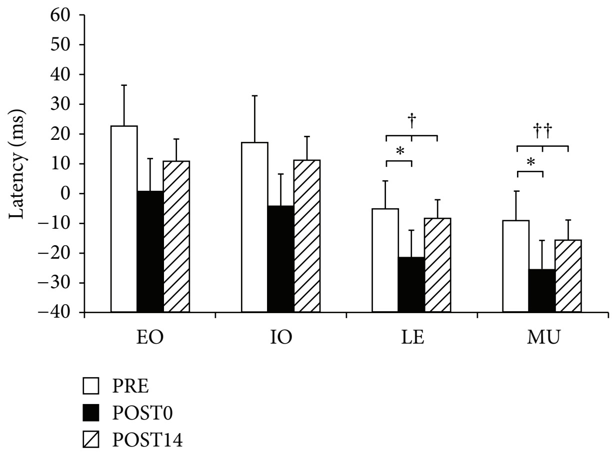


Figure 3: Results for APAs of individual muscles at all three sessions. The statistical significant change assessed by one-way repeated-measures ANOVA is presented by † ($p < 0.05$) and †† ($p < 0.01$). The results of t-test are represented by ∗ ($p < 0.05$). Muscles presented are obliquus externus (EO), obliquus internus (OI), erector spinae at level L1 (LE), and multifidus (MU).

the abdominal wall muscles (EO and IO). Similar behaviour was observed for the reaction task where the back muscle latencies ranged between 98 and 108 ms and the abdominal wall muscle latencies ranged between 109 and 122 ms.

APA latencies for all three sessions are presented in Figure 3 and additional details of statistical analysis are presented in Table 1. There were statistically significant differences among all sessions for LE and MU muscle but not for the EO and IO. The decrease in the latency from PRE to POST0 was observed for all muscles. However, only changes in LE muscle and MU muscle were statistically significant. There was a clear increase in the latency from POST0 to POST14 for all muscles, but these changes were not statistically significant ($p > 0.05$). None of the muscles' latencies at POST14 returned completely to the level of PRE. However, differences between these two sessions were not statistically significant ($p > 0.05$) for any of the muscles.

All muscles showed statistically significant differences ($p < 0.01$) in the reaction latencies among all sessions (Figure 4 and Table 1). An evident increase in the latency from PRE to POST0 was observed for all muscles ($p < 0.05$). A slight decrease in the latency from POST0 to POST14 was observed for all muscles, but these changes were not statistically significant. Consequently, the latency for any muscle

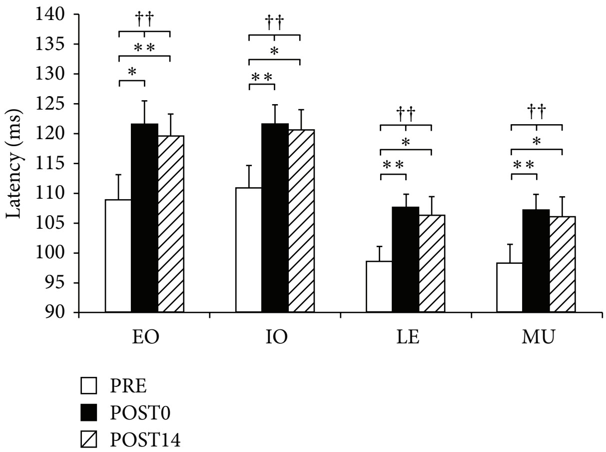


Figure 4: Results for RRs of individual muscles at all three sessions. The statistical significant change assessed by one-way repeated-measures ANOVA is presented by † ($p < 0.05$) and †† ($p < 0.01$). The results of t-test are represented by ∗ ($p < 0.05$) and ∗∗ ($p < 0.01$). Muscles presented are obliquus externus (EO), obliquus internus (OI), erector spinae at level L1 (LE), and multifidus (MU).

at POST14 did not return to the level of PRE. Moreover, statistically significant differences in latency were observed for all muscles ($p < 0.05$).

4. Discussion

This study was the first to analyse the effects of a short term BR on the EMG aspects of the APAs and RRs of the trunk muscles in aging adults. As predicted in our first hypothesis, APAs and RRs EMG latencies were changed following fourteen days of inactivity. Interestingly, the trend of the adaptations of the APAs and RRs EMG latencies to BR was not uniform regarding to the different muscle groups. The second hypothesis can only be partially confirmed. The recovery to pre-BR levels was observed only for APAs but not for RRs regardless of the muscle group observed.

The changes in APAs and RRs EMG latencies following BR were expected, as deconditioning of the passive, active, and neural mechanisms was shown to take place during BR. The passive system, especially the ligamentous system, could have probably been affected by changes in spinal curvature (posterior lumbar ligaments are put under additional tension) and changes in the intervertebral discs height and volume as shown in previous studies [18]. These structures are rich in proprioceptors and are possibly responsible for the changes in afferent sensory input to the nervous system [38]. Previous BR studies have also shown significant effect of BR on the active (i.e., muscular) system due to muscle wasting [16, 17]. And finally, changes in muscle activation properties dependent on the activity of the nervous system have also been proposed as indicators of central changes following BR [19, 20].

A trend of earlier muscle EMG onset during APAs was present for abdominal wall and back muscles by observing the group averages. However, only changes in EMG onset for the back muscles were statistically significant. These observations could be partially explained by the results of previous studies, reporting selective changes in the muscle cross-sectional area. Erector spinae, quadratus lumborum, and especially the multifidus muscles have been shown to suffer from the most prominent decreases in the cross-sectional area [15–17], but much less or no atrophy was observed in the abdominal muscles (OE, IO, and rectus abdominis) [16, 17]. Central inhibition usually accompanies muscle inactivity-immobilization [39] possibly explaining the central mechanisms of delayed EMG onset during APAs in more affected muscle groups. An additional cause could be the specific adaptation of the intramuscular coordination during APAs. In the elderly these responses have been shown to be less uniform and include more pronounced changes in proximal stabilization strategies [10, 12]. However further research is needed to assess these possible changes in aging adults following longer periods of inactivity. Based on the reports, showing APA sensitivity to aging and BR to a lesser extent than RRs [14], APAs can be considered as a robust trunk and spine stabilizing strategy. However, future research should include observations of EMG amplitude (i.e., EMG impulse) and centre of pressure amplitude during APAs to more thoroughly study the nature of adaptations in APAs and stability of the spine and trunk.

An important finding of this study was the difference in the recovery of EMG onset during APAs and RRs following the fourteen days of active recuperation. The EMG latencies during RRs have not shown any significant return to pre-BR levels. These differences might have been due to the specifics in the rehabilitation protocol. Unanticipated postural perturbations were applied primarily via lower limbs, but to a lesser extent using the arms. Future research should incorporate unanticipated postural perturbations applied via upper limbs or directly onto the torso. The APAs on the other hand showed a trend towards full recovery. This might additionally be due to the upper limb strengthening exercises.

The discrepancy between changes in EMG onset during APAs and RRs can also be attributed to their different physiological and neurological background. The APAs are centrally controlled, being primarily dependent on anticipation and prior experience. Their initiation is dependent on the anticipation of a forthcoming perturbation and its effect on posture and stability and not as a response to a perturbation in a feedback manner as is the case with RRs [12]. We can speculate that the timing of EMG onset during RRs is dependent on possible peripheral changes in sensory mechanisms such as changes in ligamentous apparatus of the spine due to changes in spinal curvature and length, atrophy of sensory enriched MU [17], and deprivation due to the lack of tonic stimulation [19] resulting in an altered sensory drive.

An alternative explanation of prolonged EMG latencies during RRs following BR can be based on the model proposed by Liebetrau et al. [40]. In this model, prolonged EMG latencies represent adjustments of the neuromuscular system to decreased trunk muscle activation amplitudes. These could be expected by generalizing findings from the observed decrease in lower limb power output following BR [41]. Based on these conclusions prolonged RRs might mirror changes in the neuromuscular strategies for preserving spinal stabilization. However, these adaptations suggest decreased capacity to preserve spinal stability under heavier load or fast occurring perturbations. To confirm this hypothesis, future research is needed. This should assess the muscle activation magnitude in order to assess the possible imbalances in force impulses of the trunk flexor and extensor muscles.

The above results represent an interesting insight into the effects of prolonged BR on muscle activation timing. However an important weakness of all BR studies, as well as of this one, is their small sample size due to the physical and social demands put on the participating subjects and high organizational demands. Future studies should continue measuring APAs and RRs in the context of BR enabling meta-analytical studies and consequently more valid information on the changes in APAs and RRs following prolonged physical inactivity.

5. Practical Relevance of the Results

BR has been shown to have a detrimental effect on specific aspects of the trunk stabilization functions (onset of EMG activity), especially during the unanticipated postural perturbations. As prolonged BR in the elderly is usually followed by a specialized rehabilitation, trunk stability must be a central concern to the therapists. These functions represent the base of any other rehabilitation activity and should be addressed accordingly. Future studies should assess the possible effect of vibration application [18, 42, 43], resistive and quick leg/arm movements [44], and unexpected perturbations to limb position on preserving trunk neuromuscular stabilization functions during and after BR. Special focus should be given to reactive trunk stabilizing actions, as these have been shown to be the most affected.

Conflict of Interests

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

Acknowledgment

The authors acknowledge financial support by (1) Slovenian Research Agency, Grant no. L5—4293 and (2) standard Project: Physical Activity and Nutrition for Healthy Aging (PANGeA), cofinanced in the framework of Slovenia-Italy cross-border collaboration programme 2007–2013 related to the funding of European fund for regional development and national funds (*Slo.: Standardni projekt: Telesna aktivnost in prehrana za kakovostno staranje (PANGeA), je sofinanciran v okviru Programa čezmejnega sodelovanja Slovenija-Italija 2007–2013 iz sredstev Evropskega sklada za regionalni razvoj in nacionalnih sredstev; Ita.: Progetto sandard: Attivity fisica e nutrizione per un invecchiamento di qualita (PANGeA), e finanziato nell'ambito del Programma per la Cooperazione Transfrontaliera Italia-Slovenia 2007–2013, da. Fondo europeo di sviluppo regionale e dai fondi nazionali*).

References

[1] D. Lintner, T. J. Noonan, and W. B. Kibler, "Injury patterns and biomechanics of the athlete's shoulder," *Clinics in Sports Medicine*, vol. 27, no. 4, pp. 527–551, 2008.

[2] J. Hoffman and P. Gabel, "Expanding Panjabi's stability model to express movement: a theoretical model," *Medical Hypotheses*, vol. 80, no. 6, pp. 692–697, 2013.

[3] M. M. Panjabi, "The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement," *Journal of Spinal Disorders*, vol. 5, no. 4, pp. 383–397, 1992.

[4] A. S. Aruin, N. Kanekar, Y. Lee, and M. Ganesan, "Enhancement of anticipatory postural adjustments in older adults as a result of a single session of ball throwing exercise," *Experimental Brain Research*, vol. 233, no. 2, pp. 649–655, 2015.

[5] B. Zazulak, J. Cholewicki, and N. P. Reeves, "Neuromuscular control of trunk stability: clinical implications for sports injury prevention," *The Journal of the American Academy of Orthopaedic Surgeons*, vol. 16, no. 9, pp. 497–505, 2008.

[6] T. Ikezoe, N. Mori, M. Nakamura, and N. Ichihashi, "Effects of age and inactivity due to prolonged bed rest on atrophy of trunk muscles," *European Journal of Applied Physiology*, vol. 112, no. 1, pp. 43–48, 2012.

[7] O. Levin, K. Cuypers, Y. Netz et al., "Age-related differences in human corticospinal excitability during simple reaction time," *Neuroscience Letters*, vol. 487, no. 1, pp. 53–57, 2011.

[8] M. J. Santos, N. Kanekar, and A. S. Aruin, "The role of anticipatory postural adjustments in compensatory control of posture: 2. Biomechanical analysis," *Journal of Electromyography and Kinesiology*, vol. 20, no. 3, pp. 398–405, 2010.

[9] M. J. Santos, N. Kanekar, and A. S. Aruin, "The role of anticipatory postural adjustments in compensatory control of posture: 1. Electromyographic analysis," *Journal of Electromyography and Kinesiology*, vol. 20, no. 3, pp. 388–397, 2010.

[10] N. Kanekar and A. S. Aruin, "Aging and balance control in response to external perturbations: role of anticipatory and compensatory postural mechanisms," *AGE*, vol. 36, no. 3, article 9621, 2014.

[11] S. Bleuse, F. Cassim, J. Blatt et al., "Effect of age on anticipatory postural adjustments in unilateral arm movement," *Gait & Posture*, vol. 24, no. 2, pp. 203–210, 2006.

[12] N. Kanekar and A. S. Aruin, "The effect of aging on anticipatory postural control," *Experimental Brain Research*, vol. 232, no. 4, pp. 1127–1136, 2014.

[13] S. Okada, K. Hirakawa, Y. Takada, and H. Kinoshita, "Age-related differences in postural control in humans in response to a sudden deceleration generated by postural disturbance," *European Journal of Applied Physiology*, vol. 85, no. 1-2, pp. 10–18, 2001.

[14] C. D. Tokuno, A. G. Cresswell, A. Thorstensson, and M. G. Carpenter, "Age-related changes in postural responses revealed by support-surface translations with a long acceleration–deceleration interval," *Clinical Neurophysiology*, vol. 121, no. 1, pp. 109–117, 2010.

[15] J. A. Hides, G. Lambrecht, C. A. Richardson et al., "The effects of rehabilitation on the muscles of the trunk following prolonged bed rest," *European Spine Journal*, vol. 20, no. 5, pp. 808–818, 2011.

[16] J. A. Hides, D. L. Belav??, W. Stanton et al., "Magnetic resonance imaging assessment of trunk muscles during prolonged bed rest," *Spine*, vol. 32, no. 15, pp. 1687–1692, 2007.

[17] D. L. Belavý, G. Armbrecht, C. A. Richardson, D. Felsenberg, and J. A. Hides, "Muscle atrophy and changes in spinal morphology: is the lumbar spine vulnerable after prolonged bed-rest?" *Spine*, vol. 36, no. 2, pp. 137–145, 2011.

[18] D. L. Belavy, G. Armbrecht, U. Gast, C. A. Richardson, J. A. Hides, and D. Felsenberg, "Countermeasures against lumbar spine deconditioning in prolonged bed rest: resistive exercise with and without whole body vibration," *Journal of Applied Physiology*, vol. 109, no. 6, pp. 1801–1811, 2010.

[19] D. L. Belavý, J. K. Ng, S. J. Wilson et al., "Influence of prolonged bed-rest on spectral and temporal electromyographic motor control characteristics of the superficial lumbo-pelvic musculature," *Journal of Electromyography and Kinesiology*, vol. 20, no. 1, pp. 170–179, 2010.

[20] D. L. Belavy, C. A. Richardson, S. J. Wilson, D. Felsenberg, and J. Rittweger, "Tonic-to-phasic shift of lumbo-pelvic muscle activity during 8 weeks of bed rest and 6-months follow up," *Journal of Applied Physiology*, vol. 103, no. 1, pp. 48–54, 2007.

[21] N. Sarabon and J. Rosker, "Effect of 14 days of bed rest in older adults on parameters of the body sway and on the local ankle function," *Journal of Electromyography and Kinesiology*, vol. 23, no. 6, pp. 1505–1511, 2013.

[22] N. Goswami, V. Kovacic, U. Marucic et al., "Effect of computerized cognitive training with virtual spatial navigation task during bed rest immobilization and recovery on vascular function: a pilot study," *Clinical Interventions in Aging*, vol. 10, pp. 453–459, 2015.

[23] J. V. Jacobs, S. M. Henry, and K. J. Nagle, "People with chronic low back pain exhibit decreased variability in the timing of their anticipatory postural adjustments," *Behavioral Neuroscience*, vol. 123, no. 2, pp. 455–458, 2009.

[24] M. Klous, P. Mikulic, and M. L. Latash, "Two aspects of feedforward postural control: anticipatory postural adjustments and anticipatory synergy adjustments," *Journal of Neurophysiology*, vol. 105, no. 5, pp. 2275–2288, 2011.

[25] M. Voglar and N. Sarabon, "Kinesio taping in young healthy subjects does not affect postural reflex reactions and anticipatory postural adjustments of the trunk: a pilot study," *Journal of Sports Science and Medicine*, vol. 13, no. 3, pp. 673–679, 2014.

[26] J. Skotte, N. Fallentin, M. Pedersen, M. Essendrop, J. Strøyer, and B. Schibye, "Adaptation to sudden unexpected loading of the low back—the effects of repeated trials," *Journal of Biomechanics*, vol. 37, no. 10, pp. 1483–1489, 2004.

[27] S. L. Morris and G. T. Allison, "Effects of abdominal muscle fatigue on anticipatory postural adjustments associated with arm raising," *Gait & Posture*, vol. 24, no. 3, pp. 342–348, 2006.

[28] H. J. Hermens, B. Freriks, C. Disselhorst-Klug, and G. Rau, "Development of recommendations for SEMG sensors and sensor placement procedures," *Journal of Electromyography and Kinesiology*, vol. 10, no. 5, pp. 361–374, 2000.

[29] A. E. Hibbs, K. G. Thompson, D. N. French, D. Hodgson, and I. Spears, "Peak and average rectified EMG measures: which method of data reduction should be used for assessing core training exercises?" *Journal of Electromyography and Kinesiology*, vol. 21, no. 1, pp. 102–111, 2011.

[30] K. Masani, V. W. Sin, A. H. Vette et al., "Postural reactions of the trunk muscles to multi-directional perturbations in sitting," *Clinical Biomechanics*, vol. 24, no. 2, pp. 176–182, 2009.

[31] A. Radebold, J. Cholewicki, M. M. Panjabi, and T. C. Patel, "Muscle response pattern to sudden trunk loading in healthy individuals and in patients with chronic low back pain," *Spine*, vol. 25, no. 8, pp. 947–954, 2000.

[32] I. A. Stokes, J. R. Fox, and S. M. Henry, "Trunk muscular activation patterns and responses to transient force perturbation in persons with self-reported low back pain," *European Spine Journal*, vol. 15, no. 5, pp. 658–667, 2006.

[33] G. Staude and W. Wolf, "Objective motor response onset detection in surface myoelectric signals," *Medical Engineering & Physics*, vol. 21, no. 6-7, pp. 449–467, 1999.

[34] S. De Wolf, H. Slijper, and M. L. Latash, "Anticipatory postural adjustments during self-paced and reaction-time movements," *Experimental Brain Research*, vol. 121, no. 1, pp. 7–19, 1998.

[35] T. Robert and M. L. Latash, "Time evolution of the organization of multi-muscle postural responses to sudden changes in the external force applied at the trunk level," *Neuroscience Letters*, vol. 438, no. 2, pp. 238–241, 2008.

[36] P. W. Hodges and B. H. Bui, "A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography," *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control*, vol. 101, no. 6, pp. 511–519, 1996.

[37] B. R. Santos, C. Larivière, A. Delisle, D. McFadden, A. Plamondon, and D. Imbeau, "Sudden loading perturbation to determine the reflex response of different back muscles: a reliability study," *Muscle & Nerve*, vol. 43, no. 3, pp. 348–359, 2011.

[38] M. M. Panjabi, "The stabilizing system of the spine. Part II. Neutral zone and instability hypothesis," *Journal of Spinal Disorders*, vol. 5, no. 4, pp. 390–397, 1992.

[39] B. C. Clark, J. L. Taylor, R. L. Hoffman, D. J. Dearth, and J. S. Thomas, "Cast immobilization increases long-interval intracortical inhibition," *Muscle & Nerve*, vol. 42, no. 3, pp. 363–372, 2010.

[40] A. Liebetrau, C. Puta, C. Anders, M. H. de Lussanet, and H. Wagner, "Influence of delayed muscle reflexes on spinal stability: model-based predictions allow alternative interpretations of experimental data," *Human Movement Science*, vol. 32, no. 5, pp. 954–970, 2013.

[41] G. Ferretti, H. E. Berg, A. E. Minetti, C. Moia, S. Rampichini, and M. V. Narici, "Maximal instantaneous muscular power after prolonged bed rest in humans," *Journal of Applied Physiology*, vol. 90, no. 2, pp. 431–435, 2001.

[42] D. L. Belavý, J. A. Hides, S. J. Wilson et al., "Resistive simulated weightbearing exercise with whole body vibration reduces lumbar spine deconditioning in bed-rest," *Spine*, vol. 33, no. 5, pp. E121–E131, 2008.

[43] D. L. Belavý, S. J. Wilson, G. Armbrecht, J. Rittweger, D. Felsenberg, and C. A. Richardson, "Resistive vibration exercise during bed-rest reduces motor control changes in the lumbo-pelvic musculature," *Journal of Electromyography and Kinesiology*, vol. 22, no. 1, pp. 21–30, 2012.

[44] I. B. van der Fits, A. W. Klip, L. A. van Eykern, and M. Hadders-Algra, "Postural adjustments accompanying fast pointing movements in standing, sitting and lying adults," *Experimental Brain Research*, vol. 120, no. 2, pp. 202–216, 1998.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 136969, 8 pages
http://dx.doi.org/10.1155/2015/136969

Clinical Study

The Impact of a Vestibular-Stimulating Exercise Regime on Postural Stability in People with Visual Impairment

Ida Wiszomirska, Katarzyna Kaczmarczyk, Michalina Błażkiewicz, and Andrzej Wit

Faculty of Rehabilitation, Józef Piłsudski University of Physical Education in Warsaw, Ulica Marymoncka 34, 00-968 Warsaw 45, Poland

Correspondence should be addressed to Ida Wiszomirska; idwi@wp.pl

Received 8 April 2015; Revised 12 June 2015; Accepted 14 June 2015

Academic Editor: Thierry Paillard

The aim of the study was to assess the impact of a vestibular-stimulating exercise regime on postural stability in individuals with visual impairment. The study group consisted of 70 people, including 28 persons (15 female and 13 male) with visual impairment and 42 (21 female and 21 male) without visual impairment. Each individual in the group with visual impairment was medically qualified for a 3-month training program. The research methodology included medical examination, anthropometric tests, and stabilometry tests on a Biodex Balance System SD (BBS). The tests were conducted twice: once before the start of training and again after 3 months of rehabilitation. The group with visual impairment showed significantly worse postural stability results than the control group for most of the stability parameters evaluated (OSI, APSI, and MLSI). Differences were noted between the groups with and without visual impairment for dynamic tests in women and for static tests in men. After training, the two groups showed roughly similar results for the stabilometry test with eyes closed. We conclude that exercises stimulating the vestibular system with head and body movements should be recommended for individuals with visual impairments to achieve better balance retention.

1. Introduction

Balance in the human body, the stability of its characteristic upright postural position, is maintained, controlled, and monitored by a complex system, consisting of the vestibular organs, the visual organs, and the organs of deep perception, touch, and pressure. Nerve signals then reach the effector organs, such as the muscles of the torso, limbs, and eyes, provoking reflex reactions coordinating body posture [1]. Impaired postural stability may be caused by dysfunction in the integration of each of the four sensory systems: visual [2, 3], proprioceptor [4, 5], exteroceptor [6], and vestibular [7, 8].

Sight, in particular, has been found to play a key role [9–11] in maintaining postural stability. Visual impairment reduces the ability to maintain balance [12] and may cause frequent injury [2, 13], including falls, occupational injuries, and traffic-related injuries. A thorough review of the literature [14] established that the risk of injury due to falls is higher for those with visual impairment than for the general population. Studies comparing individuals with/without visual impairment undergoing static and dynamic tests have confirmed that about 80% of our sensory perception is gathered in such situations by the visual system. Our movements are mainly controlled and coordinated by the eyes. Hence not only is the visual system responsible for cognition of objects, it is also used to give information to the brain about the position of our body [15], which processes and integrates information from other sensory systems to select the strategy for maintaining balance [16].

Nevertheless, the literature contains inconclusive and limited results on the possibility of compensating for balance in individuals with visual impairment. Some studies have shown that persons who are blind can maintain balance better than individuals without visual impairment [17], while others have reported results pointing in the opposite direction [18, 19]. One reason for this may be because blind individuals are able to compensate to mitigate their disability: other elements involved in ensuring postural stability—deep sensation and the vestibular system—may function more actively in individuals who are blind in vertical posture maintenance and thus compensate for the lack of visual stimuli [2, 8]. Vestibular system function is the main source of such compensation [15], as vestibular and labyrinth reflexes play a special role when other sources are reduced or absent [8].

The absence of the sense of vision can affect the motor skills of the individual, although this does not prevent individuals who have visual impairment from being physically active. The practice of sporting activity can bolster their sense of independence and autonomy, generating multiple benefits of a physiological, sociological, and psychological nature. It is therefore important to find new ways to encourage social relations between individuals with visual impairment and their peers as well as other people outside their homes [20].

Various studies have reported the positive impact of vestibular rehabilitation on postural stability in individuals experiencing dizziness [21–23], as well as patients who have disorders such as multiple sclerosis (MS) [24]. A review [25] covering 27 studies involving 1668 individuals with unilateral peripheral dysfunction of the vestibular organ who had undergone vestibular rehabilitation—a distinctive form of therapy that aims to limit dizziness, balance disruptions, and improve gaze stability, postural stability, and general physical condition—concluded that such therapy is safe and effective in such cases. These findings suggest the potential viability of developing a similar experimental exercise program for individuals without such vestibular disorders.

The goal of our study, therefore, was to assess the impact of training meant to stimulate the vestibular system in individuals with visual impairment. We hypothesized that such training would improve the effective use of this system, creating new motor patterns and thereby improving postural stability. Such research may help us better understand the mechanisms involved in balance maintenance in persons with visual impairment and encourage the practice of proper exercise.

2. Method

2.1. Participants. A total of 70 people took part in the study, including 28 individuals with visual impairment as the experimental group and 42 individuals without visual impairment as the control group. The inclusion criteria for the experimental group were visual impairment from birth or in the first months of life, the absence of any disease affecting the sense of balance, and no previous performance of motor activity aimed at improving the vestibular system. Qualification was conducted by a medical doctor and included a medical history interview (on history of falls, fractures, stumbling, and dizziness), analysis of medical records, clinical examination, assessment of eye damage, electrocardiography (ECG) and assessment of the cranial nerves for any possible meningeal symptoms, cerebellar testing (finger-to-nose test, diadochokinesis, and pronator drift test), and the results of static and dynamic tests which assess the correctness of posture and gait (Romberg's test, Unterberger's test, the Babinski-Weil test, the Fukuda test, and the straight line test).

The participants with visual impairment were all adolescent pupils at a vocational school for the blind in Laski (Poland), categorized for total or legal blindness according to the US Association for Blind Athletes (USABA) sport classification: blind (B1s; 22) and low vision (B3s; 10). Twenty-four of the 32 adolescents were congenitally blind; the other eight experienced onset in early childhood.

The individuals with visual impairment who qualified for the training program were preliminarily divided by the qualifying physician into 2 groups based on health criteria: blind without damage to the vestibular system (20 patients) and low vision (i.e., able to sense light) without damage to the vestibular system (8 people). However, in view of the fact that there was no difference in stabilometric results between these groups (much like in the study of Haibach et al. [18]), we considered them together as a single "group with visual impairment."

The members of the control group, in turn, selected on the basis of age and physical traits (height and weight), were either students of the Academy of Physical Education in Warsaw (majoring in physiotherapy, a course of study that does not involve any particularly intense regime of physical activity) or high school students, in whom the qualifying physician found no damaged sense of balance or postural disorders. This group did not participate in the vestibular training program.

The general level of physical activity was similar for both groups of participants (attending physical education classes three times a week for 45 minutes). Approval was obtained from the Institute's Research Ethics Commission and additional informed consent was obtained from all participants for whom any identifying information is included in this paper.

2.2. Measurements. Postural stability was measured using a Biodex Balance System SD (BBS), an instrument designed to measure and train postural stability on static or unstable surfaces. The BBS consists of a circular platform that is free to move in the anterior-posterior and medial lateral axes simultaneously, offering the ability to control the movement degree of the platform by 12 levels. The BSS device is interfaced with dedicated software (Biodex Medical Systems, Inc., version 1.3.4), allowing the BSS to measure the degree of tilt in each axis, providing an average sway score. Eight springs located underneath the outer edge of the platform provide resistance to movement (stability level of the platform), with resistance levels ranging from 12 (most stable) to 1 (least stable). The BBS has a display to give feedback in real time about the posture and is calibrated before use.

The participants stood on the BBS supported on their two legs, facing towards the display, for all time trials. All trials were conducted without shoes and foot position was recorded using coordinates on the platform's grid to ensure the same stance and, therefore, consistency with future tests. In this research three measurement protocols were used: the Postural Stability Test (PST) under two different conditions (stable platform, unstable platform level 8) and the Fall Risk Test (FRT) with eyes closed. The unstable platform level 8 was chosen for our purposes after a pilot study indicated that this would be an appropriate level to test for dynamic balance, not too difficult for most of the participants. In the FRT, the platform is unstable and thus permits investigators to obtain the fall risk index (FRI). This test was conducted using the standard software configuration: three trials of 20 seconds each, with ten seconds rest between tests, and platform levels varying from 12 to 6. In the PST the platform is static in the anterior-posterior stability index (APSI) and medial-lateral stability index (MLSI) axes and can measure the overall stability index (OSI). This test consisted of three trials, each 20 seconds long, with one minute between trials. These indexes represent fluctuations around a zero point established prior to testing when the platform was stable [26].

For all participants, basic somatic measurements were taken—body weight and height using a standard electronic scale and anthropometer. On the basis of these anthropometric measurements the Body Mass Index (BMI) was calculated. For participants with visual impairment, measurements were performed twice—once before training and again immediately after the training cycle.

The recorded data were analyzed statistically using the STATISTICA software (v.10). A 2 (group) × 2 analysis of variance (ANOVA) was performed with repeated measures on the OSI, APSI, MLSI, and FRI tests, for significant differences between the groups and pretraining baseline measures. Significant group-by-test interactions were further broken down by comparing pretraining differences separately for each group using Tukey's post hoc HSD test for differing sample sizes. Normality of distribution was analyzed by the Shapiro-Wilk test. Because the tested features have normal distributions, variables were compared between the first and second test using parametric Student's t-test ($p < 0.05$).

2.3. Intervention. The group of participants with visual impairment underwent training sessions (approximately 20 minutes each), twice a week, over the course of 3 months, with the aim of encouraging them to engage in more physical exercise, improve their physical fitness, and particularly improve their balance control. In particular, stimulation of the vestibular organ [27] was achieved by having these participants rotate their heads and bodies in the sagittal, transverse, and frontal planes.

In designing this exercise program, we anticipated that, by triggering sensory conflict, such activity would force the correct responses from the vestibular and proprioceptor organs, leading to an improved relationship between them, which in turn should improve balance control. Stimulation caused by movements of the head and body is perceived as a nonspecific impulse. When the stimuli are repeated regularly and not followed by a loss of balance, the response is inhibited, which in turn leads to the formation of a new image of the vestibular position in the CNS. This method aims to improve the perception and verification of information originating from the external and internal environment.

The exercises started with gentle, slow movements of the head without visual control in various low positions (lying position), such as body bent at 30°, with participants lying on their backs or sides. For example, after first lying on their back, participants shifted to lying on their right side and then to their left side, repeating this approximately 10 times (in keeping with their individual capabilities). After all the repetitions were performed, participants stopped moving for approximately 3 minutes while lying on their right side. The cycle was then repeated, this time stopping on the left side. Another exercise involved sitting with legs extended straight and then performing side bends to rest the arms on the mat behind the body, alternatingly on the left and right side. After several repetitions were performed, movement was halted for about three minutes in the side-bend position with upper limbs bent and with head held motionless, alternatingly on the left side and the right side. Following relaxation to music, the next exercise involved gentle, slow movements of the head with the participants keeping their eyes closed and maintaining the final position for approximately 3 minutes. Once in the final positions, participants performed breathing and relaxation exercises with their heads kept motionless. Such exercises without visual control, supplemented with additional auditory impulses (music), aimed to improve kinaesthesia by stimulating signals from the vestibular organ through head and body movements. In the final phase of each movement, the participant was stopped in the corrective position and asked to maintain the set position. The purpose of this halting of movement for about three minutes after performing each simple exercise movement was to equalize the flow (following head movement) of endolymph in the semicircular canals and the otoliths of the saccule and utricle of the vestibular organ. As the exercise program progressed, the number of repetitions was gradually increased and the support plane was reduced. These exercises were further supplemented with additional auditory stimuli (soothing music), intended to help create an appropriately positive mood, calm the subjects, and make the motor exercises more attractive. The relevance of such music in attempting to boost vestibular performance may be justified by the close proximity and close interrelatedness of the auditory and vestibular organs [28] and the fact that receptor cells for the both organs are situated in the inner ear and transmit information along the same, eighth cranial nerve (the vestibulocochlear nerve) [29]. The maximum heart rate during exercise did not exceed 100 bpm (a fact that underscores the low intensity of the exercise regime).

3. Results

No significant differences were found (Tukey's post hoc HSD test for different sample sizes) among participants of the same sex with respect to age, body height, weight, and Body Mass Index (BMI) (Table 1). But because there were significant statistical differences found in traits such as body height and weight between participants of different sexes, the male and female groups were considered separately, thus yielding four subgroups (with/without visual impairment, male/female). According to the BMI standards adopted by the WHO [30], the arithmetic mean of this variable for each group fell within the normal range.

The training program was completed by 13 men and 15 women in the group with visual impairment. Four participants were excluded from the study because of illness-related

Table 1: Participants' demographics (mean ± SD).

Group	n	AGE [years]	Body height [cm]	Body mass [kg]	BMI
Female 1 (without visual impairment)	21	20.0 ± 0.89	163.14 ± 3.26	57.43 ± 4.51	21.58 ± 1.83
Female 2 (with visual impairment)	15	19.14 ± 1.56	159.4 ± 6.45	53.64 ± 9.15	21.14 ± 3.45
Male 3 (without visual impairment)	21	18.85 ± 1.35	175.43 ± 2.56	69.18 ± 6.22	23.01 ± 2.85
Male 4 (with visual impairment)	13	19.08 ± 1.6	169.6 ± 6.57	62.73 ± 12.46	21.86 ± 4.82

Table 2: Female participants' baseline measures for stability index (mean ± SD): women without (female 1) and with (female 2) visual impairment.

Stability index (SI)	Female 1 (n = 21)	Female 2 (n = 15)	"Effect Size"	p
OSI CE, static	1.9 ± 0.78	2.17 ± 0.89	0.32	"NS"
APSI CE, static	1.39 ± 0.83	1.45 ± 0.74	0.08	"NS"
MLSI CE, static	0.99 ± 0.59	1.35 ± 0.77	0.53	"NS"
OSI CE, 8	2.24 ± 0.48	3.28 ± 0.59	**1.94**	$\boldsymbol{p < 0.033}$
APSI CE, 8	1.57 ± 0.42	2.57 ± 0.68	**1.82**	$\boldsymbol{p < 0.004}$
MLSI CE, 8	1.3 ± 0.22	1.58 ± 0.42	0.87	"NS"

OSI: overall stability index, APSI: anterior-posterior stability index, MLSI: medial-lateral stability index, OE: open eyes for participants fully sighted, closed for those with visual impairment, CE: closed eyes, static: stable platform, 8: dynamic balance level 8, and "Effect Size": mean of female 2 − mean of female 1/standard deviation.
p: Tukey's HSD test.

Table 3: Male participants' baseline measures for stability index (mean ± SD): men without (male 3) and with (male 4) visual impairment.

Stability index (SI)	Male 3 (n = 21)	Male 4 (n = 13)	"Effect Size"	p
OSI CE, static	1.42 ± 0.61	2.58 ± 1.5	**1,09**	**0.000**
APSI CE, static	1.13 ± 0.57	1.72 ± 1.06	**0,72**	**0.049**
MLSI CE, static	0.66 ± 0.36	1.48 ± 1.5	**0,88**	**0.002**
OSI CE, 8	3.59 ± 1.69	3.67 ± 0.89	0,08	"NS"
APSI CE, 8	2.24 ± 1.01	2.65 ± 0.71	0,48	"NS"
MLSI CE, 8	2.3 ± 1.18	1.99 ± 0.79	−0,31	"NS"

OSI: overall stability index, APSI: anterior-posterior stability index, MLSI: medial-lateral stability index, OE: open eyes for participants fully sighted, closed for those with visual impairment, CE: closed eyes, static: stable platform, 8: dynamic balance level 8, and "Effect Size": mean of male 4 − mean of male 3/standard deviation.
p: Tukey's HSD test.

absences, due to an adopted rule whereby participants were excluded from the program if they missed two sessions.

Tables 2 and 3 show the stabilometric parameters obtained in the first test by the group with visual impairment as compared to the control group. No statistically significant differences were noted in these tests between the sexes (comparing group 1 to group 3 and group 2 to group 4). No significant differences were observed in any of the static parameters for women under closed-eye conditions, while men performing the same test did show significantly worse results in the group with visually impairment than the control group. In unstable-platform tests, women with visual impairment fared worse than women without visual impairment, except in the ML index.

For the men with visual impairment, there were no significant differences in the unstable platform protocol test; however, this test was not performed for 30% of the respondents (n = 5), for whom it proved too difficult and who were excluded from the study as a result.

Analysis of the average values for the overall stability index after training showed statistically significant differences for both women with visual impairment (Table 4), with a value of 2.17 ± 0.89 to 1.43 ± 0.75, and men with visual impairment (Table 5), with a value of 2.58 ± 1.5 to 1.42 ± 0.73, although the OSI derivatives, the AP and ML parameters, do not show the same changes for the two genders.

In women there are no changes, whereas among men the AP index significantly improved. A similar pattern was found for the dynamic test, with the OSI changing significantly at the $p < 0.01$ level in individuals of both sexes and the AP index improving only in women. Noteworthy progress was made on the FRT protocol (platform levels varying from 12 to 6) by both groups. The test using variable platform stability settings is therefore the most difficult test for dynamic postural stability.

After the training cycle was completed, the results were compared between the participants in the groups with and without visual impairment, finding no statistically significant

Table 4: Comparison of pre- and postexercise values for stability index (mean ± SD) in women with visual impairment.

Stability index (SI)	Female 2 preexercise (n = 15)	Female 2 postexercise (n = 15)	"Effect Size"	p
OSI CE, static	2.17 ± 0.89	1.43 ± 0.75	0.9	0.016
APSI CE, static	1.45 ± 0.74	1.11 ± 0.77	0.45	"NS"
MLSI CE, static	1.35 ± 0.77	0.88 ± 0.58	0.7	"NS"
OSI CE, 8	3.28 ± 0.59	2.67 ± 0.6	1.02	0.003
APSI CE, 8	2.57 ± 0.68	1.96 ± 0.51	1.02	0.007
MLSI CE, 8	1.58 ± 0.42	1.41 ± 0.5	0.37	"NS"
FRI (12–6)	2.83 ± 0.45	2.37 ± 0.45	1.02	0.029

Note. OSI: overall stability index, APSI: anterior-posterior stability index, MLSI: medial-lateral stability index, FRI: fall risk index, CE: closed eyes, 8: dynamic balance level 8, and static: platform stable.
p: Student's t-test.

Table 5: Comparison of pre- and postexercise values for stability index (mean ± SD) in men with visual impairment.

Stability index (SI)	Male 4 preexercise (n = 13)	Male 4 postexercise (n = 13)	"Effect Size"	p
OSI CE, static	2.58 ± 1.5	1.42 ± 0.73	**1.04**	**0.008**
APSI CE, static	1.72 ± 1.06	1.0 ± 0.56	**0.89**	**0.048**
MLSI CE, static	1.48 ± 1.5	0.83 ± 0.49	0.65	"NS"
OSI CE, 8	3.67 ± 0.89	3.25 ± 0.96	**0.44**	**0.006**
APSI CE, 8	2.65 ± 0.71	2.3 ± 0.8	0.46	"NS"
MLSI CE, 8	1.99 ± 0.79	1.83 ± 0.62	0.23	"NS"
FRI (12–6)	3.35 ± 0.72	2.71 ± 0.65	**0.93**	**0.002**

Note. OSI: overall stability index, APSI: anterior-posterior stability index, MLSI: medial-lateral stability index, FRI: fall risk index, CE: closed eyes, 8: dynamic balance level 8, and static: platform stable.
p: Student's t-test.

differences between the groups of the same sex in closed-eye conditions. Women without visual impairment had a general stability index in the static test of 1.9 ± 0.78, whereas women with visual impairment after training showed 1.43 ± 0.75; for men the respective values were 1.42 ± 0.61 and 1.42 ± 0.73. For the dynamic test, the stability indexes also did not differ significantly between the group of women without visual impairment (2.24 ± 0.48) and with visual impairment after training (2.67 ± 0.6), or men without visual impairment (3.59 ± 1.69) and men with visual impairment after training (3.25 ± 0.96).

4. Discussion

Balance is an indispensable factor for individuals with visual impairment, helping to encourage their integration in space [20]. Balance disorders have been the subject of numerous studies, and the results of these have been used in developing forms of exercise designed to address such disorders. Studies have shown that persons with visual impairment perform worse on Postural Stability Tests than those without visual impairment [28–30].

In the study reported herein, prior to training, no significant differences were noted between the two groups of women in static tests, whereas in unstable-platform tests such differences are observed in the general index and in anterior-posterior deviations. Male participants without impaired vision, on the other hand, were found to differ significantly in terms of the stabilometric parameters with the static tests. In tests on an unstable platform (level 8), without visual control, men in the group with visual impairment did not differ significantly from men in the group without visual impairment. This may be due to the fact that 5 men were unable to perform the unstable-platform test (some individuals with visual impairment have trouble maintaining balance even on a slightly unstable base, and these individuals were excluded from the study as a result).

After the training regime, however, the results of participants (men and women) with impaired vision became approximate to those of participants without impaired vision operating with eyes closed and did not differ significantly in terms of any of the parameters evaluated. Overall, then, our study indicates that individuals with impaired vision can significantly improve their balance, although it was found that even specialized training is not capable of fully compensating for the impact of vision loss on balance retention. Different results in stability measurements between blind and sighted individuals (in tests with eyes closed) were obtained by Melzer et al. [31]. They evaluated stability in static tests with no additional auditory-memory tasks and found no differences between the groups. In contrast, when their participants performed additional auditory-memory tasks, in sighted participants the stability parameters dropped, in contrast to participants who were blind. This may suggest that individuals with/without visual impairment use different strategies to maintain stability.

Giagazoglou et al. [11] presented results that differed from those of the above authors but which agreed with the results in our study. They found that the ability to control balance in both the AP and ML directions was significantly worse in female participants who were blind than in those who were fully sighted. When sighted women performed the tests blindfolded, their COP deviations increased significantly in both directions; however, the blind participants still consistently achieved significantly inferior results. The results of Blomqvist and Rehn's [9] study using the DOLS (Dynamic One-Leg Stance) test, in turn, indicate that individuals who are blind are not able to compensate for loss of vision by utilizing other sensory extravisual information during dynamic motor tasks on the right and left limb.

Aydog et al. [32] found no differences between the results of individuals playing goalball who are blind and sighted, in all three indices (SI) on the BBS platform. The athletes' postural stability was better than that of blind people who engaged in little physical activity. Specific exercise, in this case playing goalball, causes adaptation in the form of greater system tolerance to imbalances. In Colaka et al.'s study [20] goalball players showed a significant advantage over their respective control groups on the Flamingo balance test. The better result of goalball players on the balance test perhaps suggests that the training program may exert an impact on motor skills [20]. Visual impairment need not be a factor differentiating predispositions towards tolerating imbalances in the body. This is confirmed by the findings of Marini et al. [33], which found persons who are blind practicing a variety of Italian basketball to be in better control of postural stability than those not doing so, as evaluated by functional tests (Fukuda and Tinetti).

The positive effects of exercise (better results of stabilometric parameters) have been repeatedly demonstrated. The authors of many studies have sought to identify an optimal exercise program for improving balance in order to minimize the future risk of falls. Gioftsidou et al. [34] used the BBS platform to study the effect of one year of football training on maintaining balance. The authors observed no statistically significant changes. Our study, in turn, provides evidence that specially designed training, dominated by balance exercises, must be conducted for postural control to be improved.

In our study, individuals with visual impairment participated in an easily implemented training program which actuates the vestibular system through motions of the head and body in all planes. This proposed form of activity has the advantage of not requiring great physical exertion, not involving a wide range of abilities, and predominantly involving relaxation exercises that also stimulate cognitive functions. It therefore can serve to activate individuals with visual impairment who do not engage in sporting activity for physical exercise; the relatively easy activity may also yield positive results in terms of improved balance and also other benefits of a psychological and sociological nature, through group work and integration. Our results showed encouraging improvements in the overall stability parameters in all test protocols used. A positive effect of vestibular rehabilitation on improving postural stability has been found in patients with dizziness [21–23] and in patients with multiple sclerosis [24].

The assumption of all these studies is that vestibular-stimulating physical activity triggers sensory conflicts which cause appropriate stimulation of the vestibular system and proprioceptors, leading to improved connections between them, which we assumed would compensate for the blindness and which in turn would improve the postural stability. Repeated linear and angular movements are sensed as a nonspecific stimulus, but when these operations are repeated regularly and there is no danger of losing balance because of them, this reaction is impeded, leading to the formation in the central nervous system of a new image of the vestibular situation.

Morozetti et al. [35] showed that rehabilitation for dizziness consisting of specific movements of the head resulted in a significant improvement in otoneurological clinical evaluation and in participants' perceptions of the head. Their test group obtained an improvement in balance assessed by the overall index of stability, while the anterior-posterior and mediolateral indices did not always differ significantly from the initial data, although reductions in the average were noted. Vestibular and labyrinth reflexes play a special role when other sources are reduced or absent [8]. Thus, the observed progress is important because stability decreases together with the lack of visual control. Radvay et al. [36] also observed a greater decrease in the stability results in dynamic tests, as confirmed by our own study presented in this paper.

In closing, we conclude that an appropriately designed regime of simple exercise can improve balance in individuals with visual impairment, although of course the lack of visual sensitivity in these patients cannot be fully replaced by activation of other receptors or nerve pathways.

5. Limitations

Two specific points bear mentioning here: firstly, the control group did not participate in the training program and were not also tested for posturography after three months. Secondly, 30% of the participants did not complete the unstable-platform test because they fell off, being unable to maintain balance in three attempts of 20 seconds each. Both of these factors could impact the interpretation of the results.

In the more general sense, we should also point out that the regime tested herein, although performed in group exercises, might be seen as less social or well-known than the reviewed and compared interventions involving football, goalball, and so forth. As such, future research may look into how similar results can be achieved through a more social, functional intervention.

6. Implications for Practitioners

Exercises designed to stimulate the vestibular system through head and body movements should be recommended for patients with visual impairment to achieve better balance retention, although a congenital dysfunction of vision cannot be fully compensated for by activating other neural pathways.

Conflict of Interests

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

Acknowledgment

The study was supported by the Polish Ministry of Science and Higher Education (AWF-DM 146).

References

[1] J. M. D. Greve, M. Cuğ, D. Dülgeroğlu, G. C. Brech, and A. C. Alonso, "Relationship between anthropometric factors, gender, and balance under unstable conditions in young adults," *BioMed Research International*, vol. 2013, Article ID 850424, 5 pages, 2013.

[2] V. Anand, J. G. Buckley, A. Scally, and D. B. Elliott, "Postural stability in the elderly during sensory perturbations and dual tasking: the influence of refractive blur," *Investigative Ophthalmology and Visual Science*, vol. 44, no. 7, pp. 2885–2891, 2003.

[3] T. Mergner, G. Schweigart, C. Maurer, and A. Blümle, "Human postural responses to motion of real and virtual visual environments under different support base conditions," *Experimental Brain Research*, vol. 167, no. 4, pp. 535–556, 2005.

[4] M. Bove, A. Nardone, and M. Schieppati, "Effects of leg muscle tendon vibration on group Ia and group II reflex responses to stance perturbation in humans," *The Journal of Physiology*, vol. 550, no. 2, pp. 617–630, 2003.

[5] M. Schieppati, A. Nardone, and M. Schmid, "Neck muscle fatigue affects postural control in man," *Neuroscience*, vol. 121, no. 2, pp. 277–285, 2003.

[6] A. Kavounoudias, R. Roll, and J.-P. Roll, "The plantar sole is a 'dynamometric map' for human balance control," *NeuroReport*, vol. 9, no. 14, pp. 3247–3252, 1998.

[7] B. L. Day, A. Séverac Cauquil, L. Bartolomei, M. A. Pastor, and I. N. Lyon, "Human body-segment tilts induced by galvanic stimulation: a vestibularly driven balance protection mechanism," *Journal of Physiology*, vol. 500, no. 3, pp. 661–672, 1997.

[8] A. M. Bacsi and J. G. Colebatch, "Evidence for reflex and perceptual vestibular contributions to postural control," *Experimental Brain Research*, vol. 160, no. 1, pp. 22–28, 2005.

[9] S. Blomqvist and B. Rehn, "Validity and reliability of the Dynamic One Leg Stance (DOLS) in people with vision loss," *Advances in Physiotherapy*, vol. 9, no. 3, pp. 129–135, 2007.

[10] W.-L. Hsu, J. P. Scholz, G. Schöner, J. J. Jeka, and T. Kiemel, "Control and estimation of posture during quiet stance depends on multijoint coordination," *Journal of Neurophysiology*, vol. 97, no. 4, pp. 3024–3035, 2007.

[11] P. Giagazoglou, I. G. Amiridis, A. Zafeiridis, M. Thimara, V. Kouvelioti, and E. Kellis, "Static balance control and lower limb strength in blind and sighted women," *European Journal of Applied Physiology*, vol. 107, no. 5, pp. 571–579, 2009.

[12] C. Sforza, L. Eid, and V. F. Ferrario, "Sensorial afferents and center of foot pressure in blind and sighted adults," *Journal of Visual Impairment and Blindness*, vol. 94, no. 2, pp. 97–107, 2000.

[13] S. R. Lord and H. B. Menz, "Visual contributions to postural stability in older adults," *Gerontology*, vol. 46, no. 6, pp. 306–310, 2000.

[14] R. Legood, P. Scuffham, and C. Cryer, "Are we blind to injuries in the visually impaired? A review of the literature," *Injury Prevention*, vol. 8, no. 2, pp. 155–160, 2002.

[15] M. Friedrich, H.-J. Grein, C. Wicher et al., "Influence of pathologic and simulated visual dysfunctions on the postural system," *Experimental Brain Research*, vol. 186, no. 2, pp. 305–314, 2008.

[16] M. Schmid, A. Nardone, A. M. De Nunzio, M. Schmid, and M. Schieppati, "Equilibrium during static and dynamic tasks in blind subjects: no evidence of cross-modal plasticity," *Brain*, vol. 130, no. 8, pp. 2097–2107, 2007.

[17] V. Juodžbaliene and K. Muckus, "The influence of the degree of visual impairment on psychomotor reaction and equilibrium maintenance of adolescents," *Medicina*, vol. 42, no. 1, pp. 49–56, 2006.

[18] P. Haibach, L. Lieberman, and J. Pritchett, "Balance in adolescents with and without visual impairments," *Insight Journal*, vol. 4, no. 3, pp. 112–123, 2011.

[19] C. V. Portfors-Yeomans and C. L. Riach, "Frequency characteristics of postural control of children with and without visual impairment," *Developmental Medicine and Child Neurology*, vol. 37, no. 5, pp. 456–463, 1995.

[20] T. Colaka, B. Bamaca, M. Aydinb, B. Mericb, and A. Ozbek, "Physical fitness level of blind and visually impaired goalball team players," *Science*, vol. 12, no. 4, pp. 247–252, 2004.

[21] M. B. Badke, J. A. Miedaner, C. R. Grove, T. A. Shea, and G. M. Pyle, "Effects of vestibular and balance rehabilitation on sensory organization and dizziness handicap," *Annals of Otology, Rhinology and Laryngology*, vol. 114, no. 1 I, pp. 48–54, 2005.

[22] M. B. Badke, T. A. Shea, J. A. Miedaner, and C. R. Grove, "Outcomes after rehabilitation for adults with balance dysfunction," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 2, pp. 227–233, 2004.

[23] D. M. Wrisley, M. J. Stephens, S. Mosley, A. Wojnowski, J. Duffy, and R. Burkard, "Learning effects of repetitive administrations of the sensory organization test in healthy young adults," *Archives of Physical Medicine and Rehabilitation*, vol. 88, no. 8, pp. 1049–1054, 2007.

[24] J. R. Hebert, J. R. Corboy, M. M. Manago, and M. Schenkman, "Effects of vestibular rehabilitation on multiple sclerosis-related fatigue and upright postural control: a randomized controlled trial," *Physical Therapy*, vol. 91, no. 8, pp. 1166–1183, 2011.

[25] S. L. Hillier and M. McDonnell, "Vestibular rehabilitation for unilateral peripheral vestibular dysfunction," *Clinical Otolaryngology*, vol. 36, no. 3, pp. 248–249, 2011.

[26] B. L. Arnold and R. J. Schmitz, "Examination of balance measures produced by the biodex stability system," *Journal of Athletic Training*, vol. 33, no. 4, pp. 323–327, 1998.

[27] I. Wiszomirska, K. Kaczmarczyk, L. Ilnicka, M. Słón, A. Frysztak, and T. Marciniak, "The effect of training stimulation vestibular organ on postural stability in elderly," *Advances in Rehabilitation*, vol. 24, no. 4, pp. 5–10, 2010.

[28] F. H. Netter, *Atlas of Human Anatomy*, Urban & Partner, 2011.

[29] S. King, J. Wang, A. J. Priesol, and R. F. Lewis, "Central integration of canal and otolith signals is abnormal in vestibular migraine," *Frontiers in Neurology*, vol. 10, no. 5, article 233, 2014.

[30] WHO Expert Consultation, "Appropriate body-mass index for Asian populations and its implications for policy and intervention strategies," *The Lancet*, vol. 363, no. 9403, pp. 157–163, 2004.

[31] I. Melzer, E. Damry, A. Landau, and R. Yagev, "The influence of an auditory–memory attention-demanding task on postural control in blind persons," *Clinical Biomechanics*, vol. 26, no. 4, pp. 358–362, 2011.

[32] E. E. Aydog, S. T. Aydog, A. A. Cakci, and M. N. Doral, "Dynamic postural stability in blind athletes using the Biodex Stability System," *International Journal of Sports Medicine*, vol. 27, no. 5, pp. 415–418, 2006.

[33] M. Marini, E. Sarchielli, M. F. Portas et al., "Can baseball improve balance in blind subjects?" *The Journal of Sports Medicine and Physical Fitness*, vol. 51, no. 2, pp. 227–232, 2011.

[34] A. Gioftsidou, P. Malliou, G. Pafis, A. Beneka, and G. Godolias, "Effects of a soccer training session fatigue on balance ability," *Journal of Human Sport and Exercise*, vol. 6, no. 3, pp. 521–527, 2011.

[35] P. G. Morozetti, C. F. Ganança, and B. M. Chiari, "Comparison of different protocols for vestibular rehabilitation in patients with peripheral vestibular disorders," *Jornal da Sociedade Brasileira de Fonoaudiologia*, vol. 23, no. 1, pp. 44–50, 2011.

[36] X. Radvay, S. Duhoux, F. Koenig-Supiot, and F. Vital-Durand, "Balance training and visual rehabilitation of age-related macular degeneration patients," *Journal of Vestibular Research: Equilibrium and Orientation*, vol. 17, no. 4, pp. 183–193, 2007.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 958450, 9 pages
http://dx.doi.org/10.1155/2015/958450

Research Article

Evaluation of Myoelectric Activity of Paraspinal Muscles in Adolescents with Idiopathic Scoliosis during Habitual Standing and Sitting

Garcia Kwok,[1] Joanne Yip,[1] Mei-Chun Cheung,[2] and Kit-Lun Yick[1]

[1] *Institute of Textiles and Clothing, The Hong Kong Polytechnic University, Hung Hom, Kowloon, Hong Kong*
[2] *Department of Social Work, The Chinese University of Hong Kong, Shatin, New Territories, Hong Kong*

Correspondence should be addressed to Joanne Yip; tcjyip@polyu.edu.hk

Received 24 April 2015; Revised 6 July 2015; Accepted 21 July 2015

Academic Editor: Massimiliano Pau

There is a number of research work in the literature that have applied sEMG biofeedback as an instrument for muscle rehabilitation. Therefore, sEMG is a good tool for this research work and is used to record the myoelectric activity in the paraspinal muscles of those with AIS during habitual standing and sitting. After the sEMG evaluation, the root-mean-square (RMS) sEMG values of the paraspinal muscles in the habitual postures reflect the spinal curvature situation of the PUMC Type Ia and IIc subjects. Both groups have a stronger average RMS sEMG value on the convex side of the affected muscle regions. Correction to posture as instructed by the physiotherapist has helped the subjects to achieve a more balanced RMS sEMG ratio in the trapezius and latissimus dorsi regions; the erector spinae in the thoracic region and/or erector spinae in the lumbar region. It is, therefore, considered that with regular practice of the suggested positions, those with AIS can use motor learning to achieve a more balanced posture. Consequently, the findings can be used in less intrusive early orthotic intervention and provision of care to those with AIS.

1. Introduction

Adolescent idiopathic scoliosis (AIS) is a multifactorial, three-dimensional deformity of the spine and trunk. It can appear and sometimes progress during any of the rapid periods of growth in children [1]. Noninvasive brace therapy is usually recommended for spinal curvatures between 21 and 40 degrees while surgery is suggested for curvatures over 41 degrees [2, 3]. Conventional orthoses are used to apply passive forces onto the human body to support the trunk alignment and control the deformity of the spine. However, the use of an external support is affected by factors such as poor appearance, bulkiness, and physical constraint that could lead to low acceptance and compliance [4, 5]. Regardless of current clinical practices, treatment is nothing more than just observation if the curve is less than 20 degrees, even if the child is at high risk of progressive spinal deformity during puberty, which is between the ages of 10–16 [6, 7]. The reason is that the prediction of curve progression is not available for untreated AIS patients (a Risser stage less than 1 is 22% and larger than 1 is only 2–4%) [8] and alternative treatment options are very limited. However, with biofeedback as an area that is eliciting growing interest in the medical and psychology fields, and its proven effectiveness for a number of physical, psychological, and psychophysical problems [9–11], it is possible that biofeedback can be one of the new techniques that will provide an alternative type of noninvasive treatment for AIS patients.

Biofeedback is a nonmedical process that involves the measuring of specific and quantifiable bodily functions of a subject, such as the brain wave activity, blood pressure, heart rate, skin temperature, sweat gland activity, and muscle tension, thus conveying the information to the patient in real-time. The basic aim of biofeedback therapy is to support a patient in realizing his/her self-ability to control specific psychophysiological processes [12]. The literature has consistently indicated that surface electromyography (sEMG) biofeedback is effective for muscle rehabilitation. A review of the sEMG studies on upper extremity dysfunctions in the physically disabled [13] concluded that sEMG is a valuable method for increasing upper extremity muscle activity and most effective when used in conjunction with physiotherapy. In a meta-analysis of sEMG biofeedback studies applied to hemiplegic stroke subjects, Schleenbaker and Mainous III concluded that the use of sEMG improves functional outcomes in both the upper and the lower extremities and that sEMG should be included in therapeutic regimes [14]. Therefore, a thorough evaluation of the myoelectric activity in the muscles of those with AIS would be extremely important for the formulation of a database towards sEMG biofeedback training.

Different studies have also been done on the paraspinal muscle activity of adolescents with idiopathic scoliosis by using sEMG. Avikainen et al. recorded the force-time and the EMG-time curves of the paraspinal muscles during maximal isometric trunk extensions. The isometric force-time curves as well as the maximal integrated EMG activity that were recorded from both sides of the thoracic and lumbar spines did not show any significant differences between the normal and scoliotic groups [15]. Chwała et al. also conducted an EMG assessment of the paraspinal muscles during static exercise in those with AIS. They found that, during symmetric and asymmetric exercises, the muscle tension patterns significantly differ for both the normal and scoliotic groups in comparison with the examination at rest, which in most cases generated positive corrective patterns [16]. Farahpour et al. investigated the EMG activity of the erector spinae and external oblique muscles during lateral bending and axial rotation between AIS patients and a healthy control group. Their findings showed that asymmetric muscle activity is not obvious in all of the tested postures; therefore, asymmetric muscle activity is not a necessary characteristic of AIS patients [17]. Odermatt et al. examined the EMG signals of the trunk muscles in braced and unbraced conditions under four specific trunk exercises. The results showed that the tested muscle area under the braced condition has a significant increase of 43% in EMG activity for three out of four exercises [18]. The mentioned studies aimed to test the paraspinal muscles under specific motions or exercises. Although the motions and exercises tested in these studies can reflect how the muscle activity of AIS patients is different from that of healthy subjects, two daily postures that are commonly performed, that is, standing and sitting, have not been included. Nault et al. investigated the difference in the standing stability between 71 able-bodied girls and subjects with AIS. The scoliotic group had a larger number of correlations between standing stability and posture parameters than the nonscoliotic group which indicates the standing imbalance of the scoliotic group [19]. The study showed that scoliotic patients have issues in maintaining standing stability. However, it was compared by using center of pressure displacements. The performance of the paraspinal muscles was not included in the study. Our approach is different. We have conducted a thorough evaluation of the myoelectric activity of those with AIS during habitual standing and sitting by using sEMG. Also, we have studied the changes in the myoelectric activity of the paraspinal muscles after a position has been suggested by a physiotherapist so as to investigate the muscle activity of scoliosis subjects during their most performed motions, that is, standing and sitting, and the difference between the habitual and suggested positions. The findings can thus be used to provide a less intrusive type of early intervention and serve as a means of care for adolescents with mild idiopathic scoliosis (i.e., Cobb's angle less than 20 degrees) and therefore reduce the possible need to prescribe brace wear treatment due to its associated psychological issues and negative impacts on adolescents.

2. Methods

A screening program was carried out in Hong Kong during 2014 with 2 schools and the target population was 10–13-year-old females. During the examination process, the subjects were invited to perform Adam's forward bending test and an OSI scoliometer was employed to measure the rib hump which is directly related to spinal rotation and rib deviations. The angle of trunk rotation (ATR) in the spine of the subjects was measured while lying prone in order to preliminarily assess their spinal conditions (as shown in Figure 1). The participants were assigned to the normal subject group (N group) if they had an ATR 0–2° without any posture problems. They were assigned to the group with signs of scoliosis (P group) if they had an ATR ≥ 3°. This is because an ATR > 3° might be an early sign of scoliosis and the concerned individuals were recommended to undergo checkups more frequently [20]. In total, out of the 185 participants who were screened, 26 were found to have an ATR ≥ 3° (14.1%).

Participants from the P group accepted the invitation to take lateral 3D images through ultrasound by using the Scolioscan [21, 22]. Clinicians use the Scolioscan to measure the spine deformity angle and rotation through manually assigned markers on 3D images. This method is considered to be potentially compatible with the traditional Cobb's angle measurement which uses X-rays and yet the subjects do not face a radiation hazard [23]. After the evaluation, a total of 21 participants with a curve angle of 6 to 20 degrees, without any previous surgical or orthotic treatment for AIS, were recruited for the study. The study was approved by the Human Ethics Committee of the Hong Kong Polytechnic University. All of the subjects signed an informed consent form along with their parents, and both were informed about the purpose of the study.

Curve type in this study is defined on the basis of the guidelines from the Peking Union Medical College (PUMC) classification system [24]. The subjects ($N = 21$) were divided into 3 groups: PUMC type Ia, a single thoracic curve with apex between the T2 and T11-T12 disc; PUMC type Ib, a single thoracolumbar curve with apex at T12, T12-L1 disc, and L1; and PUMC type IIc (double curves) thoracic and thoracolumbar/lumbar curves, with a curve magnitude difference less than 10 degrees°. The concave and convex sides of the paraspinal muscle region were identified based on ultrasound images obtained from the Scolioscan. The demographic data of the subjects are shown in Table 1.

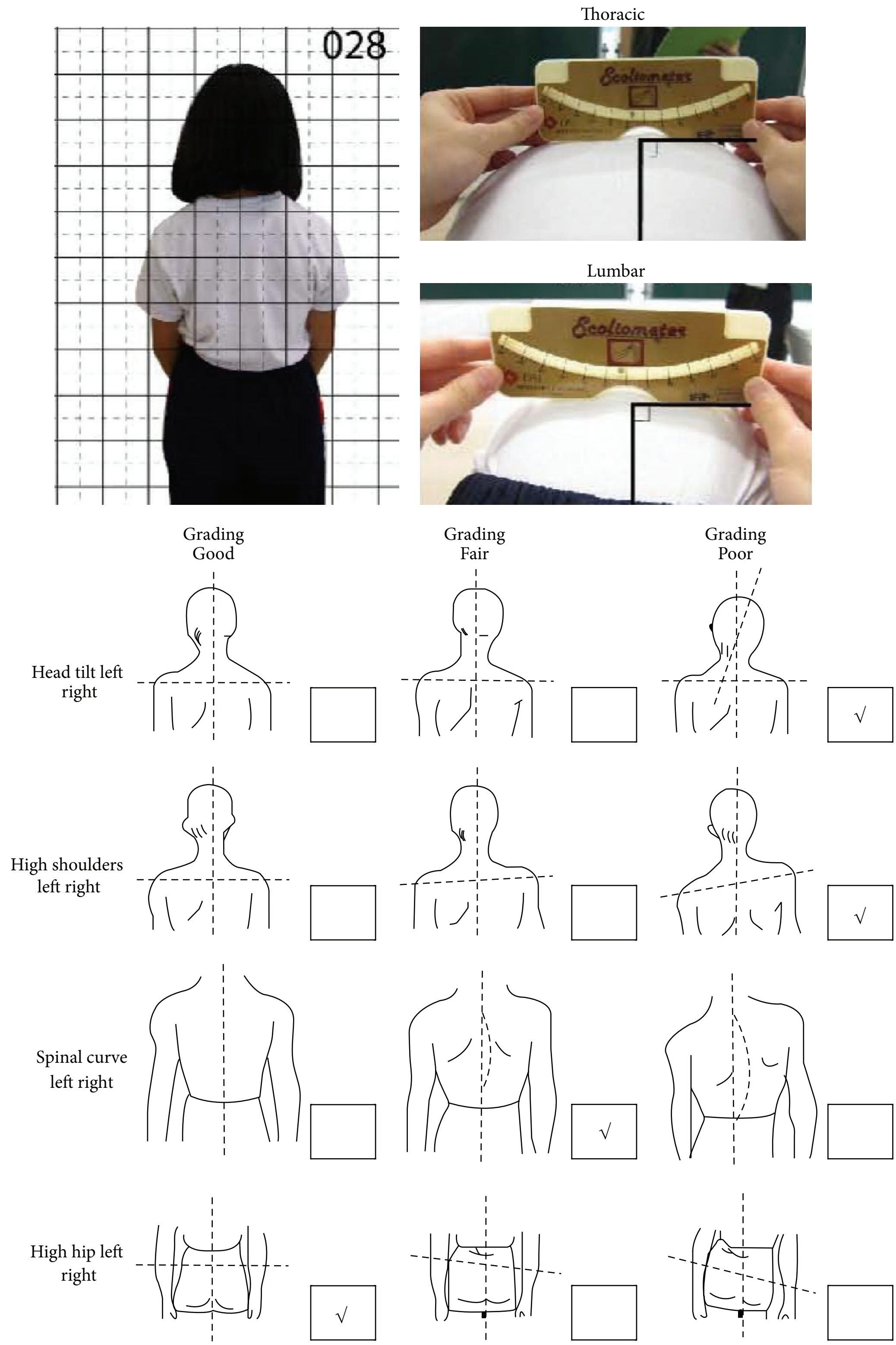


Figure 1: Adam's forward bending test and posture record form.

Table 1: Demographic data of subjects with ATI ≥ 3°.

PUMC type	Convex side	N =		Mean (S.D.)
Ia	Right (thoracic)	4	Height (cm)	156.5 (5.00)
			Weight (kg)	47.8 (5.34)
			Thoracic curve angle (°)	17.6 (8.02)
			Lumbar curve angle (°)	0
Ib	Left (thoracolumbar)	2	Height (cm)	153.5 (3.54)
			Weight (kg)	45.4 (11.7)
			Thoracic curve angle (°)	13.6 (0.85)
			Lumbar curve angle (°)	0
IIc	Right (thoracic)	7	Height (cm)	150.1 (5.98)
			Weight (kg)	42.89 (7.72)
	Left (lumbar)		Thoracic curve angle (°)	15.8 (5.21)
			Lumbar curve angle (°)	16.7 (4.03)

The parameters of the sEMG assessment were formulated based on the Surface EMG for Noninvasive Assessment of Muscles (SENIAM) standards [25]. The sEMG activity was acquired with the use of a preamplified sensor, MyoScan (model T9503M), and a data acquisition system, Flexcomp (model T7555M), both from Thought Technology (Montreal, Canada). The sEMG electrodes with ground reference from the same company (Triode T3402M) were placed onto the paraspinal muscles, namely, the trapezius, latissimus dorsi, erector spinae at the thoracic region, and erector spinae at the lumbar region (Figure 2) in pairs to determine the muscle activity along the whole spine. The back of the subject was shaved and cleaned with alcohol as suggested by the SENIAM standards. The electrodes were placed by a physiotherapist based on the SENIAM instructions and the sEMG signals were verified by an impedance test found in the BioGraph Infiniti software (Thought Technology Ltd.). The EMG assessment was only performed when the impedance check indicated that the data received were under 50 khms/s.

Figure 3 shows the habitual standing postures of some of the selected participants. It can be observed that the common posture problems are kyphosis, flat back, rounded and elevated shoulders, and a pushed-forward head position.

During the collection of the sEMG data for habitual standing, the study participants were barefoot with arms relaxed and lightly clasped in front of their body and feet positioned 20 cm apart. They were instructed to focus straight ahead and look at a designated point [26]. The habitual postures are the natural postures of the subjects performed without any instructions from the physiotherapist. An adjustable height treatment table was used for all the habitual sitting positions. The hips and knees were flexed to 90°. Under standardized instructions, the participants were positioned by the same investigator for all of the trials.

The requirements of the standing and sitting positions were used per recommendations in McKenzie [27] and Cheung et al. [28], and the subjects were guided to perform the posture accordingly by the physiotherapist (as shown in Figure 4).

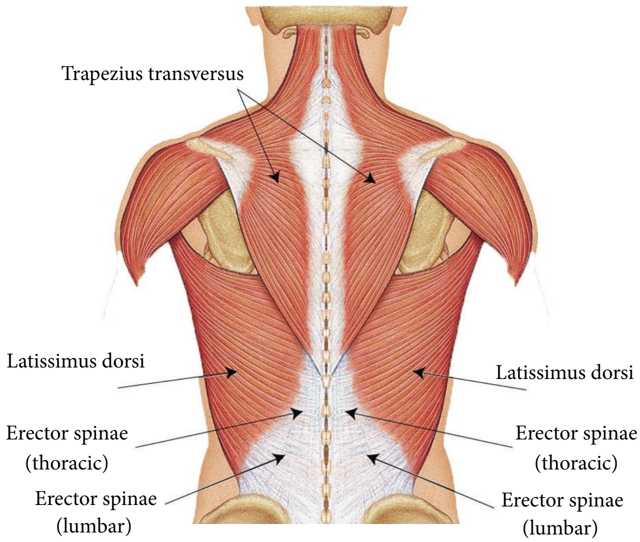


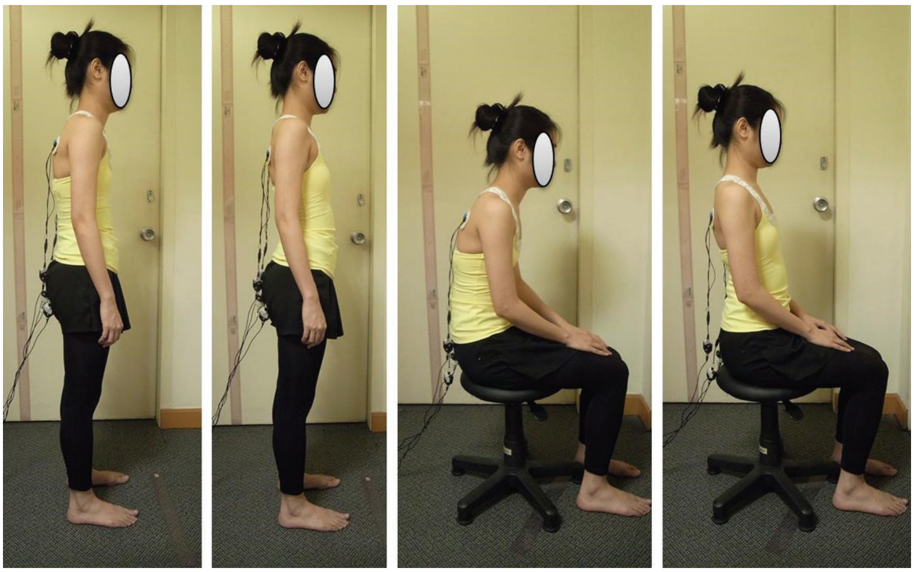
Figure 2: Placement of the EMG electrodes at the targeted paraspinal muscle regions.

(1) Suggested standing position: the head and ankles should be straight, shoulders and hip are level, kneecaps face the front, the head and knees are straight, and the chin should be parallel to the floor and aligned with the ears. The lower back should be slightly bent forward with the aid of the chest, stomach, and buttock muscles.

(2) Suggested sitting position: the head and ankles should be straight, shoulders and hips are level, kneecaps face the front, and the chin should be parallel to the floor and aligned with the ears. The lower back should be slightly bent forward to support the body with no extra weight distributed onto the spine.

The measurements of the EMG activity of the paraspinal muscles of the subjects were taken during the habitual postures and suggested positions of standing and sitting for a

FIGURE 3: Habitual standing postures of participants.



Habitual standing Suggested standing Habitual sitting Suggested sitting

FIGURE 4: Subject whose posture has been "corrected" per instructions from a physiotherapist.

duration of 1 minute and repeated twice. A band pass filter that ranged from 10 to 500 Hz was applied to eliminate undesired artifacts, such as sudden movement, and a 60 Hz notch filter was used to eliminate noise. The sEMG signals were sampled at a rate of 2048 Hz. The EMG raw data were averaged by using root mean square (RMS) to obtain the average amplitude of the EMG signal. The RMS sEMG ratio of the subjects was calculated based on the following equation [29]:

$$\text{RMS sEMG Ratio} = \frac{\text{RMS sEMG (convex)}}{\text{RMS sEMG (concave)}}. \quad (1)$$

The ratio is an index of the symmetric sEMG activity of the tested muscles, in which when the ratio is 1, the tested pairs of muscles have identical sEMG activity from the concave and convex sides of the tested muscle. If the ratio is less than 1, the concave side of the muscle has stronger sEMG activity than the convex side. If the ratio is larger than 1, the concave side of the muscle has weaker sEMG activity than the convex side. The equation was applied to assess the effectiveness of the suggested positions for the scoliosis subjects. The suggested positions are effective if the ratio is closer to 1 compared to the ratio recorded for the habitual postures.

3. Statistics

Statistical analyses were conducted by using the SPSS 19 program for Windows. The difference between the convex and concave sides during habitual standing and sitting was compared by t-testing with the significance level at $p < 0.05$. Besides, the difference between habitual and the suggested standing and sitting sEMG ratios was compared through the significance of a one-sample t-test with a test value of 1. The level of significance is $p < 0.05$.

Table 2: Result of mean RMS sEMG values (S.D.) of the paraspinal muscles of subjects during habitual standing and sitting.

PUMC type	Convex side	$N =$	Muscle region	Mean RMS sEMG (S.D.) (μV) Habitual standing (left)	Habitual standing (right)	Habitual sitting (left)	Habitual sitting (right)
Ia	Right (thoracic)	4	Trapezius	3.00 (2.21)	**4.90 (2.30)**	2.65 (1.23)	**5.73 (4.24)**
			Latissimus dorsi	2.75 (1.70)	**4.22 (1.12)**	3.13 (1.55)	**5.16 (1.79)**
			Erector spinae thoracic	2.68 (0.62)	**6.06 (6.88)**	3.95 (1.71)	**11.5 (13.4)**
			Erector spinae lumbar	3.24 (3.17)	**5.30 (5.91)**	4.95 (5.11)	**6.02 (4.56)**
Ib	Left (thoracolumbar)	2	Trapezius	2.62 (0.81)	*7.93 (1.29)*	2.54 (1.39)	*6.37 (2.33)*
			Latissimus dorsi	6.15 (2.41)	*7.30 (1.62)*	6.95 (4.14)	*7.99 (3.02)*
			Erector spinae thoracic	4.94 (2.44)	*5.50 (1.91)*	10.3 (1.04)	5.82 (1.93)
			Erector spinae lumbar	4.06 (1.57)	*19.5 (13.1)*	5.77 (2.04)	*11.9 (10.4)*
IIc	Right (thoracic)	7	Trapezius	1.71 (1.04)	**4.94 (6.40)**	2.40 (1.55)	**5.06 (5.13)**
			Latissimus dorsi	3.57 (2.19)	**4.58 (2.72)**	4.03 (2.30)	**4.44 (1.90)**
	Left (lumbar)		Erector spinae thoracic	**3.06 (1.99)**	2.83 (1.17)	**9.39 (7.25)**	8.02 (4.67)
			Erector spinae lumbar	**3.68 (2.38)**	3.31 (1.75)	4.71 (2.72)	5.15 (3.19)

4. Results

Table 2 shows the results of the mean RMS sEMG values (S.D.) of the paraspinal muscles of the subjects during habitual standing and sitting. The data are categorized based on PUMC type and occurrence of spinal curvature on the convex side of the subjects.

Table 3 shows the mean of the RMS sEMG ratio values ($\pm s$) of the paraspinal muscles of the subjects during habitual standing and sitting and the suggested standing and sitting positions. The highlighted parts indicate posture improvement under the guidance of the physiotherapist, in which the RMS sEMG ratio of the suggested positions is closer to 1 as opposed to that obtained by the same habitual posture.

5. Discussion

Based on the acquired data in Table 2, it can be observed that, for the habitual postures, the convex side of the paraspinal muscles tends to have stronger RMS sEMG values than the concave side at certain regions where spinal curvature has occurred (in bold and italic in Table 2). This situation is found to be true for both the PUMC type Ia and IIc subjects. This indicates that the curvature of the spine has affected the paraspinal muscle activity and caused muscle impairment. For example, for the PUMC type IIc subjects, the spinal curve is found at the thoracic and lumbar regions, with the convex to the right side at the thoracic region and to the left side at the lumbar region. The average RMS sEMG values at the right side of the trapezius and latissimus dorsi are stronger than those at the left side, and the average RMS sEMG values at the left side of the erector spinae thoracic are stronger than those at the right side. For the PUMC type Ia subjects, the convex side of their spinal curvature is on the right side, and the average RMS EMG values on the right side for all tested muscle regions are stronger than those on the left side. In terms of the latissimus dorsi region during habitual standing, the convex and concave sides have a significant difference ($p = 0.043$). This is consistent with the findings of Mannion et al. where the concave side of the paraspinal muscles has lower bioelectric activity which caused muscle impairment at the convex side [30].

However, a contradictory situation is found in the PUMC type Ib subjects for most of the tested regions; the RMS EMG values show a different result than that for the PUMC type Ia and IIc subjects. The RMS EMG values at the muscle region on the concave side are larger than those on the convex side. This may be due to the comparatively smaller degree of spinal curvature of the PUMC type Ib subjects (mean = 13.6); therefore, the reduction of bioelectric activity on the concave side was not reflected from the sEMG data. It is important to note that the convex side where the spinal deformity has taken place does not necessarily incur stronger sEMG values as opposed to the concave side. An influencing factor to take into consideration could be due to the degree of the spinal curvature.

The treatment for adolescents with mild idiopathic scoliosis is often passive, with only periodical observation by orthopaedists. It is possible that scoliotic adolescents suffer from progressive spinal curvature in a short period of time during puberty [31]. Adolescents with an idiopathic scoliosis curve that is over 21° are usually recommended to wear a brace made of rigid materials, such as that with a plastic or metal frame, to limit the progression of the spinal deformity. The brace causes irritation to the wearer, thus resulting in deterioration in the quality of life [32]. In some of the more severe cases, surgery may be needed to rebuild the spine by fusing bone grafts to the spine discs. This type of surgery has significant impacts on patients, which results in their immobility and therefore inability to carry out daily activities.

Despite the suggested therapy in which patients with mild idiopathic scoliosis should be periodically observed rather than prescribed with any type of treatment or exercise during the early stages, unless a rapid change takes place in the curvature angle or there is spine rotation, adolescents with

Table 3: Mean of RMS sEMG ratio values ($\pm s$) of the paraspinal muscles of subjects during habitual standing and sitting.

PUMC type	Concave side	$N =$	Muscle region	Mean RMS EMG ratio ($\pm s$) Habitual standing	Suggested standing	Habitual sitting	Suggested sitting
Ia	Right (thoracic)	4	Trapezius	2.54 ± 1.95	4.22 ± 5.87	3.84 ± 3.95	**3.16 ± 2.16**
			Latissimus dorsi	2.00 ± 1.22	2.41 ± 1.64	2.22 ± 1.66	2.32 ± 2.06
			Erector spinae thoracic	1.87 ± 1.70	2.00 ± 1.49	2.83 ± 2.69	**2.51 ± 2.68**
			Erector spinae lumbar	1.70 ± 0.85	3.05 ± 2.45	3.50 ± 4.88	**2.66 ± 2.92**
Ib	Left (thoracolumbar)	2	Trapezius	0.34 ± 0.16	**0.48 ± 0.47**	0.38 ± 0.08	**0.53 ± 0.19**
			Latissimus dorsi	0.90 ± 0.53	0.60 ± 0.15	0.83 ± 0.20	0.75 ± 0.28
			Erector spinae thoracic	0.87 ± 0.14	**0.99 ± 0.67**	1.84 ± 0.43	**1.34 ± 0.74**
			Erector spinae lumbar	0.30 ± 0.29	**0.91 ± 0.43**	0.91 ± 0.97	**1.03 ± 0.97**
IIc	Right (thoracic)	7	Trapezius	2.73 ± 1.76	3.76 ± 4.87	2.39 ± 1.63	1.96 ± 0.94
			Latissimus dorsi	2.46 ± 2.90	**1.79 ± 1.76**	1.97 ± 2.63	2.50 ± 3.88
	Left (lumbar)		Erector spinae thoracic	1.36 ± 1.14	**1.10 ± 0.67**	1.48 ± 1.35	**1.35 ± 1.26**
			Erector spinae lumbar	1.21 ± 0.59	1.01 ± 0.65	1.21 ± 1.25	1.22 ± 1.55

mild scoliosis could be treated with the use of exercise, as indicated by work in the literature; see [33–35]. In 1984, Dickson [36] provided a critical review on the use of exercise for the treatment of scoliosis. In the article, examples of successful cases in which scoliosis was treated with exercise were provided [37], and by adding loads to recover the postural balance of the patients, "the spinal deformity can be completely eliminated" [36]. As mentioned, Chwała et al. [16] performed an EMG assessment of adolescents with idiopathic scoliosis, and the results suggested that patients with a single curve show a beneficial corrective factor during asymmetric load-free and symmetric exercises.

In this study, the subjects are asked to perform habitual standing and sitting and given suggested standing and sitting positions, which are recorded with the EMG method. The suggested positions as guided by a physiotherapist with the aim of retaining postural balance of the subjects during standing and sitting are considered to be common daily positions. By restricting the scoliotic adolescents to a balanced posture, the paraspinal muscles between the two sides of the spine were able to achieve a more balanced state.

The results in Table 3 show that the PUMC type Ib subjects, with a single thoracolumbar curve, benefit relatively more from the suggested positions. During the suggested sitting position, they are able to achieve a more balanced RMS EMG ratio (closer to 1) at the trapezius, erector spinae thoracic, and erector spinae lumbar regions versus in their habitual sitting posture. During the suggested sitting position, for PUMC types Ia and Ib subjects, the RMS EMG ratio is closer to 1 at the trapezius, erector spinae thoracic, and erector spinae lumbar regions versus when in the habitual sitting posture. The result suggests a similar finding as that in Chwała et al. [16] in that the single curve patients with idiopathic scoliosis benefit most during static exercise compared to those with double curve scoliosis.

Overall, during habitual standing, the RMS sEMG ratio for the trapezius region has a significant difference with 1 ($p = 0.023$) while none of the tested muscle regions have a ratio with a significant difference with 1 during the suggested standing posture. During habitual sitting, the ratio for both the trapezius area and erector spinae at the thoracic region tends to have a significant difference of 1 ($p = 0.053$ and $p = 0.076$, resp.) while during the suggested sitting posture, only the ratio of the trapezius has a significant difference with 1 ($p = 0.025$). The results show that the subjects can achieve a more balanced sEMG signal during the suggested posture while standing and sitting.

Based on the presented findings, it is considered that, through the motor learning ability of those with AIS, postural balance can be permanently maintained by practicing the suggested positions on a regular basis as a new motor task [38]

6. Conclusions

The aim of this study is to conduct a thorough evaluation of the myoelectric activity of those with AIS during habitual standing and sitting and comprises part of our research work for the formulation of a database towards sEMG biofeedback training. The major findings of this study are as follows.

(1) The results from the PUMC type Ia and IIc subjects reflect consistency with stronger RMS sEMG values from the convex side of the paraspinal muscles as opposed to the concave side.

(2) The correction of posture per instructions from a physiotherapist can reduce muscle impairment as evidenced by the RMS sEMG ratios, and the result indicates that the PUMC type Ia and Ib subjects with a single thoracic or thoracolumbar curve benefit relatively more from the suggested positions. The result also echoes a similar finding in the literature on treating scoliosis patients with static exercises. The single curve patients benefit more and the paraspinal muscles between the two sides of their spine are better

balanced than their counterparts with a double curve during symmetrical exercise.

The findings can be used to provide a less intrusive type of early intervention and serve as a means of care for adolescents with mild idiopathic scoliosis (i.e., Cobb's angle less than 20 degrees) and therefore reduce the possible need to prescribe brace wear treatment due to its associated psychological stress and negative impacts on adolescents. It is considered that, through motor learning of the suggested positions as a new type of motor task, the concave and convex sides of the paraspinal muscles will be more balanced.

Conflict of Interests

The authors declare that there is no conflict of interests with regard to the publication of this paper.

Acknowledgment

This work was supported by funding from the Innovation and Technology Commission through the ITF project (ITS/283/13) entitled An Innovative Body-Mapping Tank Top Equipped with Biofeedback System for Adolescents with Early Scoliosis.

References

[1] M. D. Rigo, M. Villagrasa, and D. Gallo, "A specific scoliosis classification correlating with brace treatment: description and reliability," *Scoliosis and Spinal Disorders*, vol. 5, no. 1, article 1, 2010.

[2] USC Center for Spinal Surgery, *Scoliosis, Children's Spine Surgery*, USC Center for Spinal Surgery, 2005, http://www.usc-spine.com/conditions/childrens-scoliosis.cfm.

[3] L. A. Dolan, M. J. Donnelly, K. F. Spratt, and S. L. Weinstein, "Professional opinion concerning the effectiveness of bracing relative to observation in adolescent idiopathic scoliosis," *Journal of Pediatric Orthopaedics*, vol. 27, no. 3, pp. 270–276, 2007.

[4] W. R. Frontera, J. K. Silver, and T. D. Rizzo, *Essentials of Physical Medicine and Rehabilitation: Musculoskeletal Disorders, Pain, and Rehabilitation*, Saunders Elsevier, Philadelphia, Pa, USA, 2nd edition, 2008.

[5] E. M. Bunge, E. W. de Bekker-Grob, F. C. van Biezen, M.-L. Essink-Bot, and H. J. de Koning, "Patients' preferences for scoliosis brace treatment: a discrete choice experiment," *Spine*, vol. 35, no. 1, pp. 57–63, 2010.

[6] P. Liu, J. Yip, K. Yick et al., "Effects of a tailor-made girdle on posture of adolescents with early scoliosis," *Textile Research Journal*, vol. 85, no. 12, pp. 1234–1246, 2015.

[7] J. E. Lonstein and J. M. Carlson, "The prediction of curve progression in untreated idiopathic scoliosis during growth," *The Journal of Bone and Joint Surgery—American Volume*, vol. 66, no. 7, pp. 1061–1071, 1984.

[8] J. Lonstein, "Scoliosis: surgical versus nonsurgical treatment," *Clinical Orthopaedics and Related Research*, vol. 443, pp. 248–259, 2006.

[9] P. Lehrer, A. Smetankin, and T. Potapova, "Respiratory sinus arrhythmia biofeedback therapy for asthma: a report of 20 unmedicated pediatric cases using the Smetankin method," *Applied Psychophysiology Biofeedback*, vol. 25, no. 3, pp. 193–200, 2000.

[10] Y. Nestoriuc, A. Martin, W. Rief, and F. Andrasik, "Biofeedback treatment for headache disorders: a comprehensive efficacy review," *Applied Psychophysiology Biofeedback*, vol. 33, no. 3, pp. 125–140, 2008.

[11] M. U. Ahmed, S. Begum, P. Funk, N. Xiong, and B. Scheele, "A multi-module case-based biofeedback system for stress treatment," *Artificial Intelligence in Medicine*, vol. 51, no. 2, pp. 107–115, 2011.

[12] B. Kappes, "Biofeedback therapy: training or treatment," *Applied Psychophysiology and Biofeedback*, vol. 33, pp. 173–179, 2008.

[13] G. M. Lyons, P. Sharma, M. Baker, S. O'Malley, and A. Shanahan, "A computer game-based EMG biofeedback system for muscle rehabilitation," in *Proceddings of the 25th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 2, pp. 1625–1628, September 2003.

[14] R. E. Schleenbaker and A. G. Mainous III, "Electromyographic biofeedback for neuromuscular reeducation in the hemiplegic stroke patient: a meta-analysis," *Archives of Physical Medicine and Rehabilitation*, vol. 74, no. 12, pp. 1301–1304, 1993.

[15] V. J. Avikainen, A. Rezasoltani, and H. A. Kauhanen, "Asymmetry of paraspinal EMG-time characteristics in idiopathic scoliosis," *Journal of Spinal Disorders*, vol. 12, no. 1, pp. 61–67, 1999.

[16] W. Chwała, A. Koziana, T. Kasperczyk, R. Walaszek, and M. Płaszewski, "Electromyographic assessment of functional symmetry of paraspinal muscles during static exercises in adolescents with idiopathic scoliosis," *BioMed Research International*, vol. 2014, Article ID 573276, 7 pages, 2014.

[17] N. Farahpour, H. Younesian, and F. Bahrpeyma, "Electromyographic activity of erector spinae and external oblique muscles during trunk lateral bending and axial rotation in patients with adolescent idiopathic scoliosis and healthy subjects," *Clinical Biomechanics*, vol. 30, no. 5, pp. 411–417, 2015.

[18] D. Odermatt, P. A. Mathieu, M. Beauséjour, H. Labelle, and C. É. Aubin, "Electromyography of scoliotic patients treated with a brace," *Journal of Orthopaedic Research*, vol. 21, no. 5, pp. 931–936, 2003.

[19] M.-L. Nault, P. Allard, S. Hinse et al., "Relations between standing stability and body posture parameters in adolescent idiopathic scoliosis," *Spine*, vol. 27, no. 17, pp. 1911–1917, 2002.

[20] W. P. Bunnell, "An objective criterion for scoliosis screening," *The Journal of Bone & Joint Surgery: American Volume*, vol. 66, no. 9, pp. 1381–1387, 1984.

[21] C.-W. J. Cheung, S.-Y. Law, and Y.-P. Zheng, "Development of 3-D ultrasound system for assessment of adolescent idiopathic scoliosis (AIS): and system validation," in *Proceedings of the 35th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '13)*, pp. 6474–6477, Osaka, Japan, July 2013.

[22] C.-W. J. Cheung and Y. Zheng, "Development of 3-D ultrasound system for assessment of adolescent idiopathic scoliosis (AIS)," in *6th World Congress of Biomechanics (WCB 2010). August 1–6, 2010 Singapore*, vol. 31 of *IFMBE Proceedings*, pp. 584–587, Springer, Berlin, Germany, 2010.

[23] G. X. Qiu, J. G. Zhang, and Y. P. Wang, "The PUMC operative classification system for idiopathic scoliosis," *Chinese Journal of Orthopaedics*, vol. 23, pp. 1–9, 2013.

[24] H. Hermens, B. Freriks, R. Merletti et al., *European Recommendations for Surface Electromyography: Results of the SENIAM Project*, Roessingh Research and Development, Enschede, The Netherlands, 2nd edition, 1999.

[25] P. B. O'Sullivan, K. M. M. Grahamslaw, M. M. Kendell, S. C. M. Lapenskie, N. E. M. Möller, and K. V. M. Richards, "The effect of different standing and sitting postures on trunk muscle activity in a pain-free population," *Spine*, vol. 27, no. 11, pp. 1238–1244, 2002.

[26] K. Lau, *Health in Your Hands: Your Plan for Natural Scoliosis Prevention and Treatment*, Health in Your Hands Pte, Singapore, 2nd edition, 2011.

[27] R. McKenzie, *Treat Your Own Back*, Spinal Publications New Zealand, Raumati Beach, New Zealand, 9th edition, 2011.

[28] J. Cheung, J. P. K. Halbertsma, A. G. Veldhuizen et al., "A preliminary study on electromyographic analysis of the paraspinal musculature in idiopathic scoliosis," *European Spine Journal*, vol. 14, no. 2, pp. 130–137, 2005.

[29] A. F. Mannion, M. Meier, D. Grob, and M. Müntener, "Paraspinal muscle fibre type alterations associated with scoliosis: an old problem revisited with new evidence," *European Spine Journal*, vol. 7, no. 4, pp. 289–293, 1998.

[30] A. Dimeglio and F. Canavese, "Progression or not progression? How to deal with adolescent idiopathic scoliosis during puberty," *Journal of Children's Orthopaedics*, vol. 7, no. 1, pp. 43–49, 2013.

[31] J. Deceuninck and J.-C. Bernard, "Quality of life and brace-treated idiopathic scoliosis: a cross-sectional study performed at the Centre des Massues on a population of 120 children and adolescents," *Annals of Physical and Rehabilitation Medicine*, vol. 55, no. 2, pp. 93–102, 2012.

[32] K. Lau, *Your Natural Scoliosis Treatment Journal: The Essential Companion for Your 12 Weeks to a Straighter and Stronger Spine*, Health In Your Hands Pte, Singapore, 1st edition, 2011.

[33] M. Monroe, *Yoga and Scoliosis: A Journey to Health and Healing*, Demos Health, New York, NY, USA, 2012.

[34] M. H. Mehta, "Pain provoked scoliosis. Observations on the evolution of the deformity," *Clinical Orthopaedics and Related Research*, vol. 135, pp. 58–65, 1978.

[35] M. C. Hawes, "The use of exercises in the treatment of scoliosis: an evidence-based critical review of the literature," *Pediatric Rehabilitation*, vol. 6, no. 3-4, pp. 171–182, 2003.

[36] R. A. Dickson, *Management of Spinal Deformities*, Butterworths, London, UK, 1984.

[37] Y.-T. Chen, M. Kwon, E. J. Fox, and E. A. Christou, "Altered activation of the antagonist muscle during practice compromises motor learning in older adults," *Journal of Neurophysiology*, vol. 112, no. 4, pp. 1010–1019, 2014.

[38] A. B. Vallbo and N. A. Al-Falahe, "Human muscle spindle response in a motor learning task," *The Journal of Physiology*, vol. 421, pp. 553–568, 1990.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 916478, 8 pages
http://dx.doi.org/10.1155/2015/916478

Research Article

Effectiveness and Limitations of Unsupervised Home-Based Balance Rehabilitation with Nintendo Wii in People with Multiple Sclerosis

Massimiliano Pau,[1] Giancarlo Coghe,[2] Federica Corona,[1,2] Bruno Leban,[1] Maria Giovanna Marrosu,[2] and Eleonora Cocco[2]

[1] *Department of Mechanical, Chemical and Materials Engineering, University of Cagliari, 09123 Cagliari, Italy*
[2] *Multiple Sclerosis Centre, Department of Public Health, Clinical and Molecular Medicine, University of Cagliari, 09126 Cagliari, Italy*

Correspondence should be addressed to Massimiliano Pau; massimiliano.pau@dimcm.unica.it

Received 25 February 2015; Revised 15 May 2015; Accepted 9 June 2015

Academic Editor: Jacob J. Sosnoff

Balance training represents a critical part of the rehabilitation process of individuals living with multiple sclerosis (MS) since impaired postural control is a distinctive symptom of the disease. In recent years, the use of the Nintendo Wii system has become widespread among rehabilitation specialists for this purpose, but few studies have verified the effectiveness of such an approach using quantitative measures of balance. In this study, we analyzed the postural sway features of a cohort of twenty-seven individuals with MS before and after 5 weeks of unsupervised home-based balance training with the Wii system. Center of pressure (COP) time-series were recorded using a pressure platform and processed to calculate sway area, COP path length, displacements, and velocities in mediolateral (ML) and anteroposterior (AP) directions. Although the results show a significant reduction in sway area, COP displacements, and velocity, such improvements are essentially restricted to the ML direction, as the Wii platform appears to properly stimulate the postural control system in the frontal plane but not in the sagittal one. Available Wii games, although somewhat beneficial, appear not fully suitable for rehabilitation in MS owing to scarce flexibility and adaptability to MS needs and thus specific software should be developed.

1. Introduction

In persons living with multiple sclerosis (MS), the reduction of the physical impairments associated with the disease represents one of the main goals of the rehabilitation process [1, 2]. In particular, since people with MS often show impaired postural control with consequent increased risk of falls [3], it is important to include specific balance training routines in the rehabilitation plan.

While different approaches (strength, resistance and aerobic training, hippotherapy, kinesio taping, whole body vibration, sensory integration balance training, etc.) have been tested for this purpose with encouraging results [4–12], in recent years the use of devices originally designed for entertainment purposes has become widespread (see the recent exhaustive review by Taylor and Griffin for further details [13]). In fact, several studies report significant improvements in balance of MS patients following both fully or partly supervised training periods (as part of a rehabilitation program) performed using the Nintendo Wii console in addition to the Balance Board and Wii-Fit software [14–18]. It is believed that the positive effects observed are due to biofeedback mechanisms, activated mainly through the visual system, even though the actual modalities of stimulation of the postural control system induced by this device have been scarcely investigated.

It is noteworthy that the outcome measures employed to assess the effectiveness of the Wii training program are mixed since balance has been assessed using either clinically rated (e.g., Berg Balance Scale, Equiscale, and the Dynamic Gait Index) or patient-reported outcome measures (such as the Activities-Specific Balance Confidence Scale) or objective instrumental measurements. Thus, there are difficulties in correctly evaluating the actual impact of this approach, as well as in generalizing the results obtained.

Few studies have evaluated changes in balance due to Wii training through postural sway analysis obtained from center-of-pressure (COP) position time-series acquired by a force platform during a quiet standing test [16–18]. This technique has been proven reliable in characterizing the performance of the postural control system in a variety of conditions, which include either suppression of the visual input or alteration of proprioceptive information [19]. Moreover, in a prospective study on people with MS, it was observed that abnormal sway values (particularly as regards the COP path length) are associated with a greater risk of falls [20].

Brichetto et al. [16] analyzed the sway area (i.e., 95% confidence ellipse) in presence and absence of visual input in two groups of individuals with MS. One of them performed 12 sessions (articulated in three 60-minute sessions/week) of supervised Wii activity, while the other (matched for age and Expanded Disability Status Scale (EDSS) score) underwent conventional balance training exercises. Although significant reductions in sway area were detected in both individuals, the statistical analysis revealed that there was a greater effect over time for the Wii-group versus the control group as regards closed-eye stabilometry. Similarly, Prosperini et al. [18] found a significant reduction in the COP path length (i.e., the overall length of the trajectory followed by the COP during the quiet standing trial) in a crossover study that analyzed balance before and after a period of 12 weeks (48 30-minutes sessions) of partly supervised home-based Wii activity in 36 patients with MS divided into two groups. They also observed that in one of the groups analyzed the training effects on static balance were substantially maintained even 12 weeks after completion of training.

Apart from the limited, not homogeneous, and somewhat incomplete data on postural sway changes originated by Wii training, there are some issues related to the use of this platform that have not yet been fully clarified. For example, the results of tests to characterize the postural strategies adopted by a patient during an actual Wii rehabilitation session showed that the Wii-Fit games commonly used for this purpose induce an unbalanced activity in sagittal and frontal planes (i.e., significantly higher COP displacements and velocities in the mediolateral direction) [21]. This phenomenon is not dependent on the pathology since it is evident with similar magnitude even in healthy individuals. Moreover, while in most previous studies the patient's activity was supervised, the great advantage of the use of the Wii platform in the home environment appears to be that the balance training can be self-administered, thus reducing the costs of the whole rehabilitation plan. Nevertheless, there is no information on possibly reduced effectiveness of this approach in absence of continuous training guidance.

Thus, on the basis of the aforementioned considerations, this study intends to assess the effect of 5 weeks of unsupervised home-based balance rehabilitation performed with Nintendo Wii, Balance Board, and Wii-Fit software, by means of postural sway analysis. In particular, the set of sway parameters considered for the evaluation of treatment effectiveness will be expanded in comparison with previous investigations including COP displacements and velocities (not previously considered). The hypothesis to be tested is that the balance training performed with the Nintendo Wii has different effects in the anteroposterior (AP) and mediolateral (ML) directions.

2. Methods

2.1. Participants. In the period of January-February 2014, a convenience sample of 38 outpatients with MS followed at the Regional Multiple Sclerosis Centre of Cagliari (Sardinia, Italy) was informed about the study by the neurologists of the center and assessed. Individuals who met the following criteria were considered eligible for the study: diagnosis of MS according to the 2005 McDonald criteria [22] and the ability to sustain a stable upright posture for at least 15 minutes, which was the minimum time required for a Wii typical balance training session. Moreover, to avoid the influence of confounding factors in the assessment of the effectiveness of the Wii-assisted balance training program, patients who were already engaged in systematic physical activity programs were excluded. The screening for eligibility criteria and the Expanded Disability Status Scale (EDSS) score attribution were performed by a neurologist experienced in MS (GC, EC, and MGM). A subgroup of 30 individuals was enrolled in the study (Figure 1). Three of them declined to participate after a few days because of lack of time.

Kick-start meetings were organized and held between March and May 2014 by a team of neurologists, physiotherapists, and engineers. In that circumstance the participants (divided into three groups) were given a Nintendo Wii Mini console, a Balance Board, the Wii-Fit software suite, and a written memorandum with the schedule of the training protocol. Detailed instructions and a practical demonstration on the basic use of the platform and the type of games to play were also delivered.

The local ethics committee approved the study and all participants signed an informed consent agreeing to participate in the study.

2.2. Training Sessions. The training was planned over a 5-week period, with 5 compulsory sessions per week and a minimum of 30 minutes per day of exercise, dividable (at participants' discretion depending on their fatigue state) into two 15-minute sessions. Instead, no limits were imposed as regards the maximum training time: participants were allowed to use the console as much as they liked thus self-administering a possible surplus of training. In any case, the daily time spent playing was recorded in the Wii console log and subsequently analyzed to verify adherence to the protocol.

We expected participants to cumulate at least 12.5 hours of training during the period of study. This value was selected after considering all previous studies involving Wii use by people with MS, in which total training time ranged from 6 to 24 hours, with single-session durations from 15 to 60 minutes, and who reported balance improvements due to

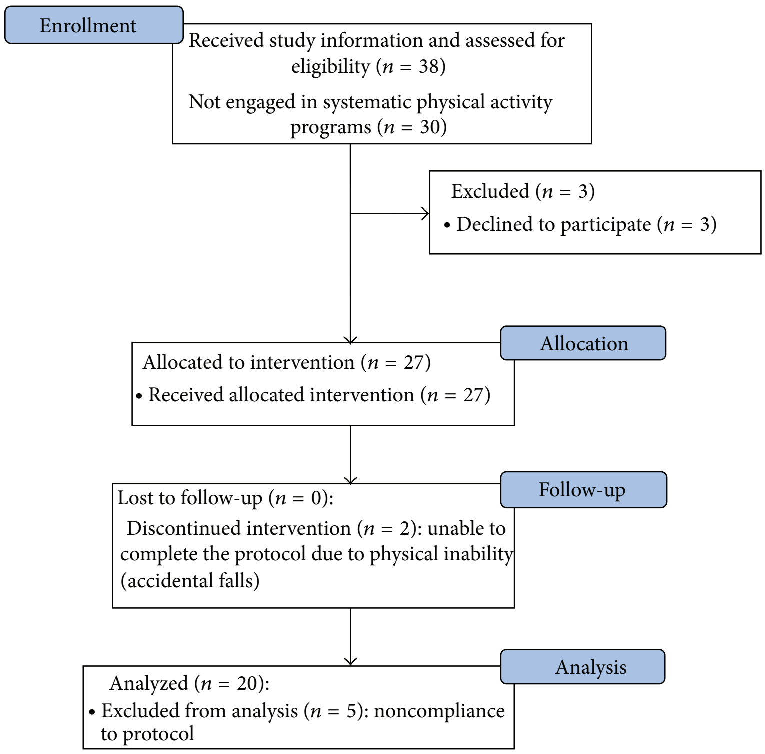


Figure 1: Flow of participants through the study.

the training program [16–18]. All the games selected for the training, namely, Penguin Slide, Table Tilt, and Balance Bubble, belong to the Wii-Fit suite and are among the most commonly employed in similar studies on individuals with MS [15, 16, 18]. Participants were instructed to equally divide the training time with the three games. For safety reasons, although no participants needed mobility aids to sustain the upright posture, we allowed them to arrange the presence of supports around the Balance Board location to use if/when needed.

2.3. Postural Sway Measurements. Postural sway was assessed before and after the training period on the basis of the analysis of COP time-series acquired using a digital force platform (BTS P6000, BTS Bioengineering, Italy) with acquisition frequency set at 480 Hz. Participants were asked to stand barefoot as still as possible for 30 seconds on the platform, having the feet placed on two 30°-oriented footprints (intermalleolar distance of 8 cm) drawn on a paper sheet placed on the force platform, while maintaining a stable and relaxed position with the arms freely positioned by the sides and the gaze fixed on a target image placed at a distance of 3 m, adjusted to each participant's eye level. These procedures ensured a common reference position, and foot placement followed recommendations of the International Society of Posturography [23]. Tests were repeated in two conditions, namely, with eyes open and closed. Three trials for each condition were acquired, allowing a suitable rest time between them.

The raw COP time-series were low-pass filtered (10 Hz cutoff, 4th-order Butterworth, bidirectional) and then post-processed with a custom-developed MATLAB (The MathWorks, Inc., Natick, MA, USA) routine to calculate the following parameters:

(i) sway area (SA, 95% confidence ellipse, mm^2),

(ii) COP path length (the overall distance travelled by the COP during the trial, mm),

(iii) COP maximum displacement (the difference between the maximum and minimum values of the selected coordinate recorded during the trial, mm) in the AP and ML directions,

(iv) COP velocity (calculated as the average of the instantaneous values recorded during the trial, $mm\,s^{-1}$) in the AP and ML directions.

2.4. Statistical Analysis. A two-way (vision × time) repeated measures ANOVA was conducted using SPSS software (v.20, IBM, Armonk, NY, USA) to examine the effect of balance training on the aforementioned sway parameters. The level of significance was set at $p = 0.05$. When necessary, a post hoc Holm-Sidak test for pairwise comparison was carried out to assess intra- and intergroup differences. Data were preliminarily checked for normality and equal variance using the Shapiro-Wilk and Levene tests.

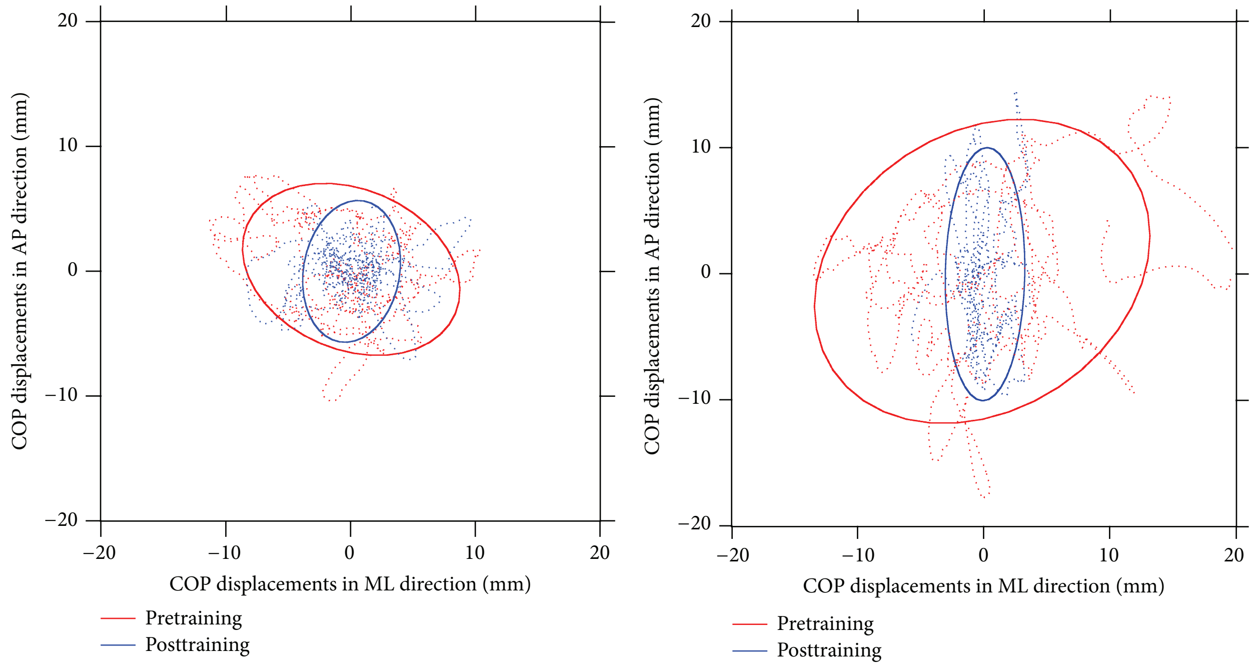


FIGURE 2: Example of sway path and areas acquired before and after Wii training.

TABLE 1: Baseline demographics and clinical characteristics of the participants.

Variable	Mean values	Range
Age (years)	44.6 ± 10.6	18.3–60.2
Body mass (kg)	62.6 ± 10.4	46.0–88.0
Stature (cm)	166.2 ± 7.7	150–178
BMI (kg m^{-2})	22.9 ± 3.8	18.6–31.2
EDSS score	3.4 ± 1.3	1.5–6.0
Total Wii training time (h)	24.23 ± 12.82	12.6–52.9

3. Results

Of the 30 patients assessed for eligibility to enter the study, 3 declined to participate; thus 27 people started the balance training program. Two persons dropped out due to injuries consequent to accidental falls (not related to the use of the Nintendo Wii) and thus data of 25 individuals were available. Moreover, the analysis of the activity logs recorded by the Wii console revealed that 5 other participants did not fully comply with the predefined schedule (e.g., they skipped some sessions or trained less than 30 minutes per day) and thus they were excluded from the analysis. These subjects were interviewed after the log analysis and when requested to explain the poor compliance, they generically reported "lack of time" and "scheduling problems" as the main reasons for the incomplete training.

Descriptive statistics for the 20 patients who correctly and completely followed the training program are reported in Table 1.

The values of the selected postural sway parameters measured before and after the training protocol are summarized in Table 2.

ANOVA revealed main effect of training as regards sway area ($F_{(1,19)} = 10.96$, $p = 0.004$, and Wilks $\lambda = 0.63$) and COP displacements and velocity in the ML direction ($F_{(1,19)} = 12.11$, $p = 0.003$, and Wilks $\lambda = 0.61$) for displacements and ($F_{(1,19)} = 4.83$, $p = 0.041$, and Wilks $\lambda = 0.79$) for velocity.

Main effects of vision were also observed for all the sway parameters: sway area ($F_{(1,19)} = 13.27$, $p = 0.002$, and Wilks $\lambda = 0.59$), COP path length ($F_{(1,19)} = 28.25$, $p < 0.001$, and Wilks $\lambda = 0.40$), COP displacements in ML direction ($F_{(1,19)} = 11.65$, $p = 0.003$, and Wilks $\lambda = 0.61$), COP displacements in AP direction ($F_{(1,19)} = 52.29$, $p < 0.001$, and Wilks $\lambda = 0.27$), COP velocity in ML direction ($F_{(1,19)} = 17.37$, $p < 0.001$, and Wilks $\lambda = 0.52$), and COP velocity in AP direction ($F_{(1,19)} = 35.27$, $p < 0.001$, and Wilks $\lambda = 0.35$).

In particular, the post hoc analysis revealed that the sway area decreased by 19% only for the eyes open condition ($p = 0.006$), the COP displacements in ML direction decreased in both visual conditions (eyes open: −11%, $p = 0.049$; eyes closed: −18%, $p < 0.001$), and COP velocity in ML direction significantly decreased only in presence of visual input (−10%, $p = 0.048$). No significant time-per-vision interactions were found.

Some examples of the sway paths and areas calculated before and after the Wii training are shown in Figure 2.

4. Discussion

Our purpose was to characterize postural sway changes in individuals with MS following 5 weeks of unsupervised

Table 2: Variation of postural sway parameters before and after the Wii training for bipedal stance in eyes-open and eyes-closed conditions.

	Eyes open			Eyes closed			Time effect	Vision effect	Time × vision interaction
	Pre-Wii	Post-Wii	Pre-post change	Pre-Wii	Post-Wii	Pre-post change	p value	p value	p value
Sway area (mm^2)	340.68 ± 197.82	275.85 ± 195.73	−64.83 ± 75.62	502.08 ± 337.78	389.44 ± 339.54	−97.96 ± 199.44	0.004†	0.002†	0.482
COP path length (mm)	469.85 ± 142.91	463.52 ± 163.92	−6.33 ± 84.54	700.95 ± 308.83	668.12 ± 330.90	−26.52 ± 159.63	0.502	<0.001†	0.521
COP max disp. ML (mm)	25.12 ± 9.54	22.36 ± 9.00	−2.76 ± 5.43	31.82 ± 13.66	25.98 ± 11.15	−5.13 ± 6.67	0.003†	0.003†	0.131
COP max disp. AP (mm)	29.38 ± 6.68	29.18 ± 10.08	−0.21 ± 6.66	39.02 ± 9.67	37.26 ± 13.98	−1.30 ± 11.47	0.629	<0.001†	0.704
COP velocity ML ($mm\,s^{-1}$)	8.94 ± 2.89	8.07 ± 3.20	−0.86 ± 1.83	12.29 ± 5.60	11.09 ± 5.78	−0.96 ± 2.72	0.041†	<0.001†	0.932
COP velocity AP ($mm\,s^{-1}$)	10.78 ± 3.86	8.11 ± 3.20	−2.16 ± 5.19	17.02 ± 7.87	16.72 ± 8.77	−0.25 ± 4.77	0.892	<0.001†	0.507

All values are expressed as mean ± SD. Pre-Wii: baseline values, Post-Wii: after 5 weeks of balance training with Nintendo Wii. The symbol † denotes statistical significance ($p < 0.05$).

home-based balance rehabilitation performed using the Nintendo Wii system.

Although several previous studies reported the usefulness of such an approach in improving balance abilities in individuals with MS, few of them specifically employed objective quantitative techniques (i.e., use of force platforms) to assess the quantitative changes in postural sway associated with the balance training [16–18].

In particular, our results are consistent with those of Guidi et al. [17] and Brichetto et al. [16] who found a reduction in sway area after 10 and 12 hours of Wii training. In particular, Brichetto et al. [16], similar to what was observed here, reported sway area reduction in presence and absence of visual input, although more markedly with respect to the present investigations (40% versus 20%). On the contrary, we found no significant improvements in COP path length as reported by Prosperini et al. [18] who observed a reduction in this parameter in the range of 15–17%, even though the baseline values of the COP path length were similar, thus indicating that, despite the differences in equipment, such COP measurements are very robust and reliable for balance assessment in individuals with MS. Such discrepancies are probably due either to the different duration/intensity of the training or to the existence of supervision by a therapist. As regards the latter aspect, several studies in different fields of neuro- and physical rehabilitation observed that although in both supervised and unsupervised exercise programs benefits for the patients were observed, supervised programs are associated with greater improvements in the investigated function [24, 25]. Also to be noted is that some differences may have been introduced by the foot position on the platform during the posturographic tests, as this variable is able to influence the sway values [26].

It is noteworthy that some improvements in postural control performances (i.e., reduction of COP displacements in ML direction) were found even when participants were tested in absence of visual input, despite the fact that the training was mainly based on visual feedback. This phenomenon was observed in similar previous studies [27, 28] and explained with the fact that the use of visual feedback during training is probably able to induce some kind of recalibration process of the sensory inputs, thus globally ameliorating the effectiveness of the postural control system.

The main novelty of the present study is represented by a more refined analysis of sway parameters that includes assessment of COP maximum displacements and velocities subdivided by anatomical plane. Our data show that the improvements associated with Wii training are restricted to the ML direction. This would indicate that the commercially available software commonly used to train individuals with MS (i.e., the Wii-Fit suite) differentially stimulates postural control activity in the sagittal and frontal planes.

Previous experiments performed to characterize the postural strategies of individuals with MS during actual rehabilitation sessions [21] showed that COP displacements and velocities in the ML direction were significantly higher with respect to the AP direction. Moreover, such a phenomenon is present even in the case of healthy individuals, and thus such unbalanced activity is likely due to the structure of the Wii-Fit games, which mainly require subjects to shift their weight from one limb to the other while reduced movements are required in the AP direction.

Indeed, the focal clinical point of our results is as follows: are we sure that we are saying that the Wii balance system leads to the best performance achievable? In fact, even though a "global" beneficial effect is introduced by the Wii training in people with MS (expressed through sway area or COP path length reduction), the Wii-Fit exercises lead to unbalanced activity in AP and ML directions in terms of COP displacements and velocities. This is likely due to the fact that this system is designed for healthy people and recreational purposes and thus may not be fully suitable for achieving controlled improvement in postural control performances.

One possible way to overcome such difficulties would be the use of dedicated software that exploits Balance Board capabilities as the input device and is also designed to meet the specific training requirements and allows therapists to define the type, duration, and difficulty of exercises depending on patients' needs and degree of impairment. Such an approach was used, for example, by Young et al. [29] and Gil-Gómez et al. [30] who built and validated customized games to train balance in older adults and individuals affected by brain injury. It is noteworthy that in the study by Young et al. [29] similar reductions of sway after training were observed in both the ML and AP directions, thus indicating that balance training with Nintendo Wii was able to stimulate postural control in a well-balanced manner.

Some limitations of the study are to be acknowledged: firstly, our design study did not include a control group, and this fact limits the possibility of attributing the changes in postural sway we observed exclusively to the effect of Wii training. Secondly, we tested only three games of the Wii-Fit suite, selecting those most widely used in previous studies on people with MS. Further studies with larger samples and a wider variety of games would be desirable to see if the unbalanced postural activity in ML direction detected here is actually to be associated with intrinsic features of the Wii system. Finally, it is still unclear whether the effect of a therapist's supervision may or may not mitigate (or remove completely) possible confusing factors affecting the results due to incorrect performance of the training which is possible in the home setting, as well as reducing the number of noncompliant participants.

5. Conclusion

The results of the present study confirm that 5 weeks of unsupervised home-based balance rehabilitation training performed using the Nintendo Wii with Balance Board and Wii-Fit games significantly reduced several postural sway parameters of individuals living with MS, thus indicating an improvement in the performance of the postural control system.

Nevertheless, there are some issues associated with such an approach that suggest caution in its use or, at the very least, further in-depth analyses. In particular, balance training appears to be much more intense in the ML than in the AP direction: in fact, the significant changes observed here as regards COP displacements and velocities involve in practice only the frontal plane. Moreover, the scarce flexibility of the Wii-Fit software in terms of exercise difficulty and scoring system make it difficult for physiotherapists to administer and customize training in accordance with the patient's initial conditions and improvements. Thus, also considering the possibilities of exploiting the interesting features of the Balance Board as a training tool even when not used in conjunction with a Wii console (as it can be connected to a common Personal Computer), future studies should address the development and testing of software specifically designed for MS needs.

Conflict of Interests

The authors report no conflict of interests.

Acknowledgments

The authors wish to thank the MS patients and Ms. Patrizia Melis and Ms. Ilde Carrus for their support in patients' assessment and data acquisition. This study was partly supported by the Fondazione Banco di Sardegna (Grant 2013.1301) and by the University of Cagliari (Grant INNOVA.RE P.O. Sardegna FESR 2007–2013 CUP F25C10001420008 DREAMS Devices for Rehabilitation in Multiple Sclerosis).

References

[1] F. Khan, L. Turner-Stokes, L. Ng, and T. Kilpatrick, "Multidisciplinary rehabilitation for adults with multiple sclerosis," *Cochrane Database of Systematic Reviews*, vol. 18, no. 2, Article ID CD006036, 2007.

[2] S. Beer, F. Khan, and J. Kesselring, "Rehabilitation interventions in multiple sclerosis: an overview," *Journal of Neurology*, vol. 259, no. 9, pp. 1994–2008, 2012.

[3] M. H. Cameron and S. Lord, "Postural control in multiple sclerosis: implications for fall prevention," *Current Neurology and Neuroscience Reports*, vol. 10, no. 5, pp. 407–412, 2010.

[4] A. Romberg, A. Virtanen, J. Ruutiainen et al., "Effects of a 6-month exercise program on patients with multiple sclerosis: a randomized study," *Neurology*, vol. 63, no. 11, pp. 2034–2038, 2004.

[5] L. S. DeBolt and J. A. McCubbin, "The effects of home-based resistance exercise on balance, power and mobility in adults with multiple sclerosis," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 2, pp. 290–297, 2004.

[6] T. Kjølhede, K. Vissing, and U. Dalgas, "Multiple sclerosis and progressive resistance training: a systematic review," *Multiple Sclerosis*, vol. 18, no. 9, pp. 1215–1228, 2012.

[7] C. Bronson, K. Brewerton, J. Ong, C. Palanca, and S. J. Sullivan, "Does hippotherapy improve balance in persons with multiple sclerosis: a systematic review," *European Journal of Physical and Rehabilitation Medicine*, vol. 46, no. 3, pp. 347–353, 2010.

[8] M. Cortesi, D. Cattaneo, and J. Jonsdottir, "Effect of kinesio taping on standing balance in subjects with multiple sclerosis: a pilot study," *NeuroRehabilitation*, vol. 28, no. 4, pp. 365–372, 2011.

[9] O. Schuhfried, C. Mittermaier, T. Jovanovic, K. Pieber, and T. Paternostro-Sluga, "Effects of whole-body vibration in patients with multiple sclerosis: a pilot study," *Clinical Rehabilitation*, vol. 19, no. 8, pp. 834–842, 2005.

[10] R. R. Mason, D. J. Cochrane, G. J. Denny, E. C. Firth, and S. R. Stannard, "Is 8 weeks of side-alternating whole-body vibration a safe and acceptable modality to improve functional performance in multiple sclerosis?" *Disability & Rehabilitation*, vol. 34, no. 8, pp. 647–654, 2012.

[11] D. Cattaneo, J. Jonsdottir, M. Zocchi, and A. Regola, "Effects of balance exercises on people with multiple sclerosis: a pilot study," *Clinical Rehabilitation*, vol. 21, no. 9, pp. 771–781, 2007.

[12] M. Gandolfi, D. Munari, C. Geroin et al., "Sensory integration balance training in patients with multiple sclerosis: a randomized, controlled trial," *Multiple Sclerosis Journal*, 2015.

[13] M. J. D. Taylor and M. Griffin, "The use of gaming technology for rehabilitation in people with multiple sclerosis," *Multiple Sclerosis Journal*, vol. 21, no. 4, pp. 355–371, 2015.

[14] M. Plow and M. Finlayson, "Potential benefits of Nintendo Wii Fit among people with multiple sclerosis. A longitudinal pilot study," *International Journal of MS Care*, vol. 13, no. 1, pp. 21–30, 2011.

[15] Y. E. Nilsagård, A. S. Forsberg, and L. von Koch, "Balance exercise for persons with multiple sclerosis using Wii games: a randomised, controlled multi-centre study," *Multiple Sclerosis*, vol. 19, no. 2, pp. 209–216, 2013.

[16] G. Brichetto, P. Spallarossa, M. L. L. de Carvalho, and M. A. Battaglia, "The effect of Nintendo Wii on balance in people with multiple sclerosis: a pilot randomized control study," *Multiple Sclerosis Journal*, vol. 19, no. 9, pp. 1219–1221, 2013.

[17] I. Guidi, T. Giovannelli, and M. Paci, "Effects of Wii exercises on balance in people with multiple sclerosis," *Multiple Sclerosis Journal*, vol. 19, no. 7, article 965, 2013.

[18] L. Prosperini, D. Fortuna, C. Giannì, L. Leonardi, M. R. Marchetti, and C. Pozzilli, "Home-based balance training using the wii balance board: a randomized, crossover pilot study in multiple sclerosis," *Neurorehabilitation and Neural Repair*, vol. 27, no. 6, pp. 516–525, 2013.

[19] A. Ruhe, R. Fejer, and B. Walker, "The test-retest reliability of centre of pressure measures in bipedal static task conditions—a systematic review of the literature," *Gait and Posture*, vol. 32, no. 4, pp. 436–445, 2010.

[20] L. Prosperini, D. Fortuna, C. Giannì, L. Leonardi, and C. Pozzilli, "The diagnostic accuracy of static posturography in predicting accidental falls in people with multiple sclerosis," *Neurorehabilitation and Neural Repair*, vol. 27, no. 1, pp. 45–52, 2013.

[21] G. Coghe, M. Pau, F. Corona et al., "Postural adaptation during a Nintendo Wii balance training," *Multiple Sclerosis Journal*, vol. 20, no. 1, supplement, p. 434, 2014.

[22] C. H. Polman, S. C. Reingold, G. Edan et al., "Diagnostic criteria for multiple sclerosis: 2005 revisions to the 'McDonald Criteria'," *Annals of Neurology*, vol. 58, no. 6, pp. 840–846, 2005.

[23] T. S. Kapteyn, W. Bles, C. J. Njiokiktjien, L. Kodde, C. H. Massen, and J. M. Mol, "Standardization in platform stabilometry being a part of posturography," *Agressologie*, vol. 24, no. 7, pp. 321–326, 1983.

[24] S. J. Olney, J. Nymark, B. Brouwer et al., "A randomized controlled trial of supervised versus unsupervised exercise programs for ambulatory stroke survivors," *Stroke*, vol. 37, no. 2, pp. 476–481, 2006.

[25] E.-Y. Kim, S.-Y. Kim, and D.-W. Oh, "Pelvic floor muscle exercises utilizing trunk stabilization for treating postpartum

urinary incontinence: randomized controlled pilot trial of supervised versus unsupervised training," *Clinical Rehabilitation*, vol. 26, no. 2, pp. 132–141, 2012.

[26] L. Chiari, L. Rocchi, and A. Cappello, "Stabilometric parameters are affected by anthropometry and foot placement," *Clinical Biomechanics*, vol. 17, no. 9-10, pp. 666–677, 2002.

[27] C. Walker, B. J. Brouwer, and E. G. Culham, "Use of visual feedback in retraining balance following acute stroke," *Physical Therapy*, vol. 80, no. 9, pp. 886–895, 2000.

[28] M. Schwenk, G. S. Grewal, B. Honarvar et al., "Interactive balance training integrating sensor-based visual feedback of movement performance: a pilot study in older adults," *Journal of NeuroEngineering and Rehabilitation*, vol. 11, article 164, 2014.

[29] W. Young, S. Ferguson, S. Brault, and C. Craig, "Assessing and training standing balance in older adults: a novel approach using the 'Nintendo Wii' Balance Board," *Gait & Posture*, vol. 33, no. 2, pp. 303–305, 2011.

[30] J.-A. Gil-Gómez, R. Lloréns, M. Alcañiz, and C. Colomer, "Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: a pilot randomized clinical trial in patients with acquired brain injury," *Journal of NeuroEngineering and Rehabilitation*, vol. 8, article 30, 2011.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 878472, 7 pages
http://dx.doi.org/10.1155/2015/878472

Research Article

Crossover versus Stabilometric Platform for the Treatment of Balance Dysfunction in Parkinson's Disease: A Randomized Study

G. Frazzitta,[1] F. Bossio,[1] R. Maestri,[2] G. Palamara,[1] R. Bera,[1] and D. Ferrazzoli[1]

[1] *Department of Parkinson Disease and Brain Injury Rehabilitation, "Moriggia-Pelascini" Hospital, Gravedona ed Uniti, 22015 Como, Italy*
[2] *Department of Biomedical Engineering, Scientific Institute of Montescano, S. Maugeri Foundation IRCCS, Montescano, 27040 Pavia, Italy*

Correspondence should be addressed to G. Frazzitta; frazzittag62@gmail.com

Received 23 February 2015; Revised 17 April 2015; Accepted 4 May 2015

Academic Editor: Luis-Millán González

Balance dysfunctions are a major challenge in the treatment of Parkinson's disease (PD). Previous studies have shown that rehabilitation can play a role in their treatment. In this study, we have compared the efficacy of two different devices for balance training: stabilometric platform and crossover. We have enrolled 60 PD patients randomly assigned to two groups. The first one (stabilometric group) performed a 4-week cycle of balance training, using the stabilometric platform, whereas the second one (crossover group) performed a 4-week cycle of balance training, using the crossover. The outcome measures used were Unified Parkinson's Disease Rating Scale (UPDRS) part II, Berg Balance Scale (BBS), Timed Up and Go (TUG), and Six Minutes Walking Test (6MWT). Results showed that TUG, BBS, and UPDRS II improved in both groups. There was not difference in the efficacy of the two balance treatments. Patients in both groups improved also the meters walked in the 6MWT at the end of rehabilitation, but the improvement was better for patients performing crossover training. Our results show that the crossover and the stabilometric platform have the same effect on balance dysfunction of Parkinsonian patients, while crossover gets better results on the walking capacity.

1. Introduction

Balance dysfunction (BD) in Parkinson's disease (PD) is a disabling symptom that contributes to falls and it impairs the ability to perform the activities of daily living, such as walking, turning, and rising to a standing position [1].

Even with optimal pharmacological and surgical management, these deficits cannot be controlled satisfactorily and have a negative impact on the quality of life [2]. There is an increasing evidence that physical therapy, especially highly challenging balance exercises, can improve BD and reduce the risk of falls, though the long-term effects of physical therapy interventions need to be further explored [3, 4]. Previous studies have shown the efficacy of training with stabilometric platform and visual feedback for balance in Parkinsonian patients in stage 3 of Hoehn-Yahr scale [5]. Unfortunately, the stabilometric platforms are not widespread and they are hard to find in the normal gymnasiums, while the crossover is widely used. Crossover is a device that simulates the movement of a skater, broadening the support base and improving the functionality of paravertebral muscles. Therefore, this device can act on several fundamental aspects of PD such as the narrow-based gait [6], the loss of automatisms (e.g., upper arms swinging) [7], and the altered functionality of the trunk muscles [8]. With its automatic and coordinate movement of upper and lower limbs, crossover could help patients in the unconscious relearning of upper limb swinging and it could also improve the narrow base support [9]. Furthermore, the wide dissemination of crossover could allow Parkinsonian patients an early, continuous, and cheap treatment that could have beneficial effects both on the development and on the control of BD.

The aim of this study was to assess the efficacy of crossover compared to the stabilometric platform in the treatment of BD in patients with PD in order to propose the crossover as a useful device for their treatment.

2. Methods

From January to October 2014, we screened one hundred Parkinsonian patients referred to the Department of Parkinson's disease Rehabilitation of the "Moriggia-Pelascini" Hospital (Gravedona ed Uniti, Italy).

Inclusion criteria were (1) probable diagnosis of PD according to Gelb et al. [10], (2) stage 3 of Hoehn-Yahr scale, (3) ability to walk without any assistance, and (4) mini mental state examination > 25.

Exclusion criteria were (1) presence of dyskinesias, (2) presence of other neurological diseases, (3) visual and/or auditory dysfunctions that impair gait and balance, (4) postural hypotension, and (5) severe orthopaedic and/or rheumatic diseases.

Patients were randomly assigned to two groups: group 1 (stabilometric group) underwent a 4-week cycle of stabilometric platform training for 6 days per week and group 2 (crossover group) underwent a 4-week cycle of crossover training for 6 days per week.

For the patients' randomization, a computer-generated list of binary random numbers was used. Both groups were evaluated at admission and at discharge, using the following outcome measures: Berg Balance Scale (BBS), Timed Up and Go (TUG) Test, Six Minutes Walking Test (6MWT), and Unified Parkinson's Disease Rating Scale part two (UPDRS II), daily activities.

Patients were evaluated at 9 A.M., one hour after taking the dopaminergic replacement therapy (including levodopa, dopamine agonists, and monoamine oxidase-B inhibitors). A neurologist, expert in movement disorders, and a physiotherapist, both blind to the purpose of the study, evaluated all patients. The study design and protocol were approved by the local Scientific Committee and Institutional Review Board (General Hospital Moriggia Pelascini, Gravedona e Uniti-Como) and followed the ethical principles outlined by the Helsinki Declaration.

2.1. Sample Size Computation. Published studies report a standard error of measurement (SEM) equal to 1.8, 0.59 s, and 30 m for BBS, TUG, and 6MWT, respectively [11, 12]. We wanted to detect a difference in improvement between groups of 2, 1 s, and 35 m, respectively. To detect these differences with a two-tailed type I error of 0.05 and a power of 80%, the estimated sample size was of 27, 12, and 24 patients per group, respectively. Taking into account the possibility of a small rate of dropout, we set our sample size to 30 + 30 patients.

2.2. Outcome Measures. The efficacy of the two different treatments was evaluated with the following outcome measures.

2.2.1. Berg Balance Scale (BBS). The Berg Balance Scale (BBS) is the gold standard scale for balance tests. It is a 14-item test designed to measure the balance during specific functional tasks. Each task is scored from 0 to 4 for a maximum of 56 points (normal subject).

2.2.2. Six Minutes Walking Test (6MWT). In the 6MWT, subjects walk as far as they can for 6 minutes. Patients are allowed to take a rest or slow down during the test (if they need it). We use for the test an unimpeded hallway length 15 meters in order to increase the number of turnings during walking, being the turning one of the major problems of Parkinsonian patients.

2.2.3. Timed Up and Go Test. This test measures mobility in elderly people and it is a useful tool to quantify locomotion performance in individuals with PD. The patients have to stand up from a chair, walk 3 meters, turn around, walk back to the chair, and sit down again. The whole test is repeated 3 times and the average time is calculated.

2.2.4. UPDRS II. The UPDRS is a common scale used to follow the longitudinal course of PD. Part II of this section assesses the activities of daily life including speech, swallowing, handwriting, dressing, hygiene, falling, salivating, turning in bed, walking, and cutting food.

2.3. Characteristics of Devices

2.3.1. The Stabilometric Platform. For the training of PD patients, we have used a stabilometric platform (Prokin 254 (Pro-Kin Software Stability), TecnoBody S.R.L., Dalmine, 24044 Bergamo, Italy). This device is a force platform with a flat and regular surface fixed to four force-transduction systems (Figure 1). The platform sends the signals to a computer for offline analysis and for detecting the position of the centre of pressure (CoP). The CoP represents the point of application of forces concerning feet and ground. The CoP area is an index of the effectiveness of the tonic postural system in keeping the centre of gravity closer to the intermediate position of balance.

2.3.2. Crossover. Crossover is a type of cross-trainer, designed for cardiovascular exercises. It has two platforms and two interconnected levers that move simultaneously. It engages the arms in a converging motion and the legs in extending, rotating, and flexing actions so that multiple muscles are working at the same time (Figure 2).

This device does not have feedback. There are 25 levels of difficulty based mainly on the progressive increase of resistance of the footboards.

2.4. Rehabilitative Treatment. All patients underwent a front-to-front treatment of 20 minutes with a physiotherapist, performing cardiovascular exercises, stretching exercises, and a passive mobilization of four limbs. Then patients underwent a stabilometric platform training or a crossover training for 15 minutes.

2.4.1. Stabilometric Group. Subjects underwent a 4-week cycle of balance training using stabilometric platform (Prokin

Figure 1: Stabilometric platform.

Figure 2: Crossover.

254, TecnoBody S.R.L., Dalmine, 24044 Bergamo, Italy) for 6 days per week. The patients performed five "stabilometric track" exercises (see Figure 3). The patients had to stand still on a force plate with their feet comfortably positioned within a box whose dimensions are equal to their foot length. They looked straight ahead at a screen surface placed 80 cm away, putting their arms on two handles. Using a visual feedback sensitive to the displacement of the centre of gravity, patients had to move their CoP within a red track in order to reach two yellow circles placed at the end of the track. Patients were not supposed to leave the track and an auditory feedback signalled any possible deflection. The five tracks allowed different positions in the space (vertical, horizontal, oblique (left/right), and circular) (Figure 3). The duration of each exercise increased from $30''$ on the first week to $60''$ on the second week and to $90''$ on the third week until it reached $120''$ on the last week. Each exercise was performed only one time on the 1st week and on the 2nd week and 2 times on the 3rd week and on the 4th week. A pause between the 4th and 5th exercise was programmed on the first two weeks and between the 3rd and 4th exercise on the two final weeks. The exercises on the platform increased in difficulty each week and were studied in order to progressively stress the patient's limit of stability (i.e., every week the length of the tracks was increased until a designed target in different space directions—antero-posterior, medio-lateral direction, etc.—was chosen) (Figure 4).

2.4.2. Crossover Group. Each patient underwent a 4-week cycle of crossover training (Technogym Crossover 700) for 6 days per week. In our study, the crossover training involved only the first 4 of the 25 levels of difficulty/resistance: first week, level 1; second week, level 2; third week, level 3; and fourth week, level 4. Patients performed a 5-minute treatment and repeated it 3 times with a 5-minute break between each repetition. The use of the first 4 levels on the crossover is due to the need to maintain the exercises on aerobic conditions.

2.5. Statistical Analysis. Descriptive statistics are reported as mean (SD). The normality of the distribution of all variables was assessed by the Shapiro-Wilk statistic. For each outcome variable considered, the effect of the two different rehabilitation protocols was assessed by a two-factor analysis of variance: the first factor was the rehabilitation protocol (stabilometric platform versus crossover) and the second factor was time (end of treatment versus baseline) with repeated measures of the time factor. If a significant interaction effect for time × treatment was found, two separate paired t-tests (one for each group of patients) were carried out to compare end of rehabilitation and baseline values.

Between-group comparisons for continuous data were assessed with unpaired t-test or with Mann-Whitney U test in case of violation of the normality assumption. Comparisons for categorical variables were carried out by the Chi-square test or Fisher's exact test when appropriate.

A p value < 0.05 was considered statistically significant. When multiple comparisons were carried out, the Bonferroni correction was applied. Accordingly, when couples of comparisons were considered, the significance level was set to 0.025. All analyses were carried out using the SAS/STAT statistical package, release 9.2 (SAS Institute Inc., Cary, NC, USA).

3. Results

We enrolled sixty patients in the study: 30 were assigned to the stabilometric platform group (group 1) and 30 were assigned to the crossover group (group 2).

Baseline demographic and clinical characteristics of all patients are reported in Table 1. No variable violated the normality assumption. No statistically significant differences were observed between the two groups in any variable at the baseline.

Results from repeated measurements analysis of variance are summarized in Table 2. A significant ($p = 0.0337$) time × group interaction was found for 6MWT, indicating a difference in the effects of stabilometric platform versus crossover rehabilitation strategies. Due to this significant interaction, two separate paired t-tests (one for each group of

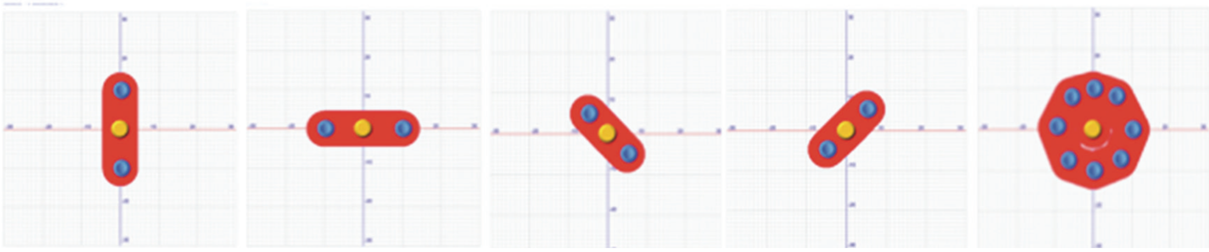


Figure 3: Examples of the five "stabilometric track" exercises.

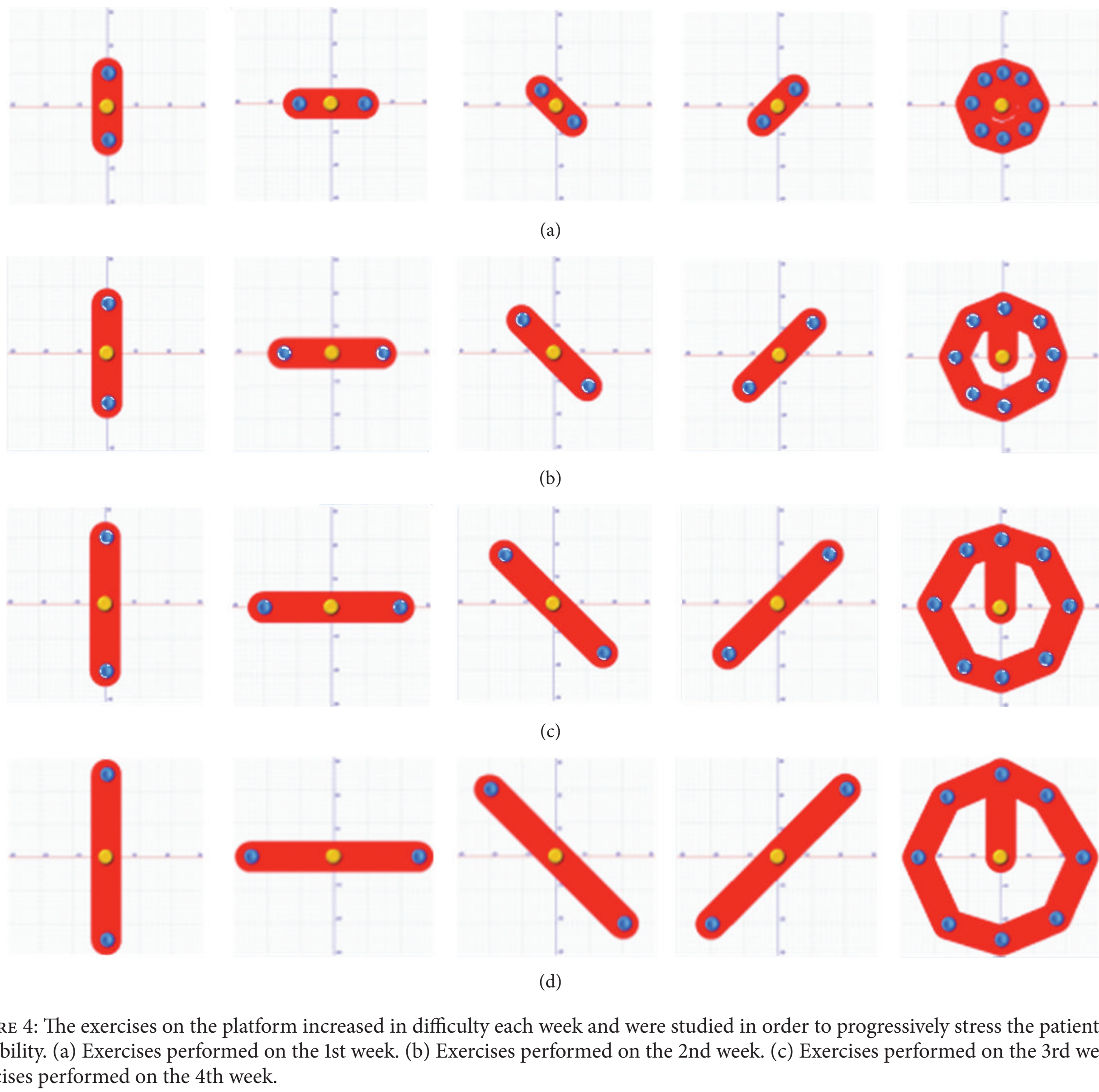


Figure 4: The exercises on the platform increased in difficulty each week and were studied in order to progressively stress the patient's limit of stability. (a) Exercises performed on the 1st week. (b) Exercises performed on the 2nd week. (c) Exercises performed on the 3rd week. (d) Exercises performed on the 4th week.

patients) were carried out, demonstrating that both rehabilitation strategies improved the distance walked in 6 minutes ($p = 0.0172$ for group 1 and $p < 0.0001$ for group 2), but the improvement was better in group 2 from a statistical point of view. A different result was found for BBS, TUG, and UPDRS II, where a largely nonsignificant time × group interaction ($p > 0.7$ all) was found, revealing no differences in the effect of treatment. All these outcome variables significantly improved after the rehabilitation period (time effect $p < 0.0001$ all).

Table 1: Demographic characteristics and basal clinical data in both groups.

Variable	Group 1 (stabilometric)	Group 2 (crossover)	p value
Age	66.6 (10.0)	65.0 (8.8)	0.500
Sex (male/female)	13/17	17/13	0.30
Height (cm)	165.7 (10.5)	166.4 (8.6)	0.76
L DOPA eq	608.7 (307.6)	740.9 (297.8)	0.059
HY	2.8 (0.4)	2.8 (0.4)	0.627
UPDRS II	15.1 (5.2)	13.9 (4.4)	0.313
BBS	45.5 (7.8)	46.3 (5.1)	0.917
TUG	12.5 (6.4)	11.0 (3.5)	0.337
6MWT	319.3 (115.1)	340.3 (87.3)	0.333

Table 2: Delta: value at discharge − baseline value (absolute effect size).

Variable	Group 1 Delta	Group 2 Delta	Group effect		Time effect		Time × group interaction	
			$F(1,58)$	p	$F(1,58)$	p	$F(1,58)$	p
BBS	7.3	7.6	0.53	0.47	170	<0.0001	0.05	0.82
UPDRS II	−5.2	−5.1	1.19	0.28	220	<0.0001	0.04	0.85
TUG	−3.3	−3.0	2.12	0.15	42	<0.0001	0.13	0.72
6MWT	68.8	103.3	2.44	0.12	118	<0.0001	4.7	0.0337

4. Discussion

The purpose of this study was to evaluate the effectiveness of a balance training using the crossover and to compare the results with those obtained from patients who underwent a balance treatment with a stabilometric platform.

Our results show that the crossover improves balance in Parkinsonian patients and that there are no differences in the BBS, TUG, and UPDRS improvement in comparison with the results obtained from patients who underwent a stabilometric platform treatment. Moreover, patients in the crossover group show better results in 6MWT in comparison with patients who underwent stabilometric platform training.

This is the first study that shows the efficacy of crossover on BD in Parkinsonian patients.

BD is a relevant challenge for the treatment of PD. In a precedent study, we showed the efficacy of a stabilometric platform treatment for balance disturbances in PD [5]. However, the stabilometric platforms are expensive and are not widespread devices. The possibility to find a common device as the crossover could allow the Parkinsonian patients an early, continuous, and cheap treatment that could have beneficial effects both on the development and on the control of BD. Using the crossover, the centre of mass moves continuously on three different directions. This movement determines a continuous shift of the direction of lower body force resulting in a possible stabilization and coordination benefits. The benefits of the crossover training may be also related to an implicit learning. Motor explicit and implicit learning strategies could improve rehabilitation outcomes in PD [13], and the use of feedback and cues improve the repetition of correct movements using a voluntary control which goes beyond the degenerated automatic motor mechanisms [14]. Moreover, several authors suggest that exercises in PD must incorporate specific characteristics in terms of intensity, repetition, specificity, and difficulty [9]. These characteristics are important to enhance cognitive engagement for the consolidation of learned behavior and for the changes of the dysfunctional motor circuits. Aerobic training could also restore neuroplasticity in striato-thalamic-cortical motor circuit, system involved in the automatic movements [9]. Through coordination and repetition, the crossover training combines the cognitive engagement with the aerobic training, both fundamental for motor learning mechanisms.

The biomechanics of crossover simultaneously involves legs in extending, abducting, flexing, and rotating actions improving the support base and the functionality of paravertebral muscles. This is a further reason that could explain its beneficial effect on BD. In fact, other aspects involved in BD in PD are the stooped posture, characterized by a flexion of the thoracolumbar spine due to degeneration of the paravertebral muscles [8], and a narrow base of support [6]. Both are responsible for body misalignment beyond the limit of stability, poor balance, and falls.

The repetitive and voluntary-induced movement that patients perform using crossover may determine a form of unconscious learning able to bypass the diseased basal ganglia and could act indirectly on the proprioceptive and vestibular systems. Moreover, this type of exercise counteracts the progressive course of the paravertebral muscles degeneration [15].

We have seen that patients who underwent a crossover treatment showed better results in 6MWT in comparison with those treated with stabilometric platform. The dopaminergic deficit in PD leads to an alteration in the striato-cortical pathways [16], which can affect motor unit recruitment and results in muscle weakness [17–19]. Progressive resistance exercises have been suggested as a treatment option to preserve function and health-related quality of life in PD [20, 21]. Our crossover training consists of an aerobic exercise with an increasing intensity (from low to moderate). Lima et al. have just reported an increase in muscle strength and an improvement of gait after progressive resistance training programmes in Parkinsonian patients [22]. The crossover involves the upper and lower limbs and previous studies have shown that whole body intervention programs might improve multiple specific PD impairments. In particular, Farley and Koshland [23] found an improvement in bradykinesia. It has been previously described [24] that bradykinesia/hypokinesia is the major specific PD impairment that limits the walking capacity assessed by the 6MWT [25]. We hypothesize that an improvement of bradykinesia might explain the improvement of 6MWT in crossover group.

The most important limitation of our study is the lack of a follow-up period in order to evaluate the persistence of beneficial effects and further studies are necessary to assess this issue. Moreover, it will be useful to quantify the results not only using scales but also with quantitative measurements, for example, a force plate.

In conclusion, our study shows that the crossover and the stabilometric platform have the same efficacy on BD and that crossover training might also improve walking capacity. Thus, we believe that the crossover can be proposed as a rehabilitative device for Parkinsonian patients in early and medium stage of disease because it is cheap and easily available in a common gym.

Conflict of Interests

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

References

[1] H. Suarez, D. Geisinger, A. Suarez, X. Carrera, R. Buzo, and I. Amorin, "Postural control and sensory perception in patients with Parkinson's disease," *Acta Oto-Laryngologica*, vol. 129, no. 4, pp. 354–360, 2009.

[2] N. Smania, E. Corato, M. Tinazzi et al., "Effect of balance training on postural instability in patients with idiopathic Parkinson's disease," *Neurorehabilitation and Neural Repair*, vol. 24, no. 9, pp. 826–834, 2010.

[3] S. D. Kim, N. E. Allen, C. G. Canning, and V. S. C. Fung, "Postural instability in patients with Parkinson's disease: epidemiology, pathophysiology and management," *CNS Drugs*, vol. 27, no. 2, pp. 97–112, 2013.

[4] G. Frazzitta, G. Bertotti, G. Riboldazzi et al., "Effectiveness of intensive inpatient rehabilitation treatment on disease progression in parkinsonian patients: a randomized controlled trial with 1-year follow-up," *Neurorehabilitation and Neural Repair*, vol. 26, no. 2, pp. 144–150, 2012.

[5] G. Frazzitta, G. Bertotti, D. Uccellini et al., "Short- and long-term efficacy of intensive rehabilitation treatment on balance and gait in parkinsonian patients: a preliminary study with a 1-year followup," *Parkinson's Disease*, vol. 2013, Article ID 583278, 5 pages, 2013.

[6] A. Charlett, C. Weller, A. G. Purkiss, S. M. Dobbs, and R. J. Dobbs, "Breadth of base whilst walking: effect of ageing and parkinsonism," *Age and Ageing*, vol. 27, no. 1, pp. 49–54, 1998.

[7] P. Redgrave, M. Rodriguez, Y. Smith et al., "Goal-directed and habitual control in the basal ganglia: implications for Parkinson's disease," *Nature Reviews Neuroscience*, vol. 11, no. 11, pp. 760–772, 2010.

[8] N. G. Margraf, A. Wrede, A. Rohr et al., "Camptocormia in idiopathic Parkinson's disease: a focal myopathy of the paravertebral muscles," *Movement Disorders*, vol. 25, no. 5, pp. 542–551, 2010.

[9] G. M. Petzinger, B. E. Fisher, S. McEwen, J. A. Beeler, J. P. Walsh, and M. W. Jakowec, "Exercise-enhanced neuroplasticity targeting motor and cognitive circuitry in Parkinson's disease," *The Lancet Neurology*, vol. 12, no. 7, pp. 716–726, 2013.

[10] D. J. Gelb, E. Oliver, and S. Gilman, "Diagnostic criteria for Parkinson disease," *Archives of Neurology*, vol. 56, no. 1, pp. 33–39, 1999.

[11] D. M. Hamilton and R. G. Haennel, "Validity and reliability of the 6-minute walk test in a cardiac rehabilitation population," *Journal of Cardiopulmonary Rehabilitation*, vol. 20, no. 3, pp. 156–164, 2000.

[12] T. Steffen and M. Seney, "Test-retest reliability and minimal detectable change on balance and ambulation tests, the 36-item short-form health survey, and the unified Parkinson disease rating scale in people with parkinsonism," *Physical Therapy*, vol. 88, no. 6, pp. 733–746, 2008.

[13] A. Nieuwboer, L. Rochester, L. Müncks, and S. P. Swinnen, "Motor learning in Parkinson's disease: limitations and potential for rehabilitation," *Parkinsonism and Related Disorders*, vol. 15, supplement 3, pp. S53–S58, 2009.

[14] M. E. Morris, R. Iansek, T. A. Matyas, and J. J. Summers, "Stride length regulation in Parkinson's disease. Normalization strategies and underlying mechanisms," *Brain*, vol. 119, no. 2, pp. 551–568, 1996.

[15] M. Nallegowda, U. Singh, G. Handa et al., "Role of sensory input and muscle strength in maintenance of balance, gait, and posture in Parkinson's disease: a pilot study," *American Journal of Physical Medicine and Rehabilitation*, vol. 83, no. 12, pp. 898–908, 2004.

[16] A. E. Lang and A. M. Lozano, "Parkinson's disease," *The New England Journal of Medicine*, vol. 339, no. 15, pp. 1044–1053, 1998.

[17] G. Frazzitta, R. Maestri, G. Bertotti et al., "Rehabilitation in Parkinson's disease: assessing the outcome using objective metabolic measurements," *Movement Disorders*, vol. 25, no. 5, pp. 609–614, 2010.

[18] R. Cano-de-la-Cuerda, M. Pérez-de-Heredia, J. C. Miangolarra-Page, E. Muñoz-Hellín, and C. Fernández-de-Las-Peñas, "Is there muscular weakness in Parkinson's disease?" *American Journal of Physical Medicine & Rehabilitation*, vol. 89, no. 1, pp. 70–76, 2010.

[19] L. M. Inkster, J. J. Eng, D. L. MacIntyre, and A. Jon Stoessl, "Leg muscle strength is reduced in Parkinson's disease and relates to the ability to rise from a chair," *Movement Disorders*, vol. 18, no. 2, pp. 157–162, 2003.

[20] L. E. Dibble, O. Addison, and E. Papa, "The effects of exercise on balance in persons with parkinson's disease: a systematic review across the disability spectrum," *Journal of Neurologic Physical Therapy*, vol. 33, no. 1, pp. 14–26, 2009.

[21] M. J. Falvo and G. M. Earhart, "Reference equation for 6-minute walk in individuals with Parkinson disease," *Journal of*

Rehabilitation Research and Development, vol. 46, no. 9, pp. 1121–1126, 2009.

[22] L. O. Lima, A. Scianni, and F. Rodrigues-de-Paula, “Progressive resistance exercise improves strength and physical performance in people with mild to moderate Parkinson’s disease: a systematic review,” *Journal of Physiotherapy*, vol. 59, no. 1, pp. 7–13, 2013.

[23] B. G. Farley and G. F. Koshland, “Training BIG to move faster: the application of the speed-amplitude relation as a rehabilitation strategy for people with Parkinson’s disease,” *Experimental Brain Research*, vol. 167, no. 3, pp. 462–467, 2005.

[24] C. G. Canning, L. Ada, J. J. Jhonson, and S. McWhirter,, “Walking capacity in mild to moderate Parkinson’s disease,” *Archives of Physical Medicine and Rehabilitation*, vol. 87, no. 3, pp. 371–375, 2006.

[25] M. J. Falvo and G. M. Earhart, “Six-minute walk distance in persons with Parkinson disease: a hierarchical regression model,” *Archives of Physical Medicine and Rehabilitation*, vol. 90, no. 6, pp. 1004–1008, 2009.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 621845, 7 pages
http://dx.doi.org/10.1155/2015/621845

Clinical Study

Effects of Unilateral Cochlear Implantation on Balance Control and Sensory Organization in Adult Patients with Profound Hearing Loss

Cécile Parietti-Winkler,[1,2,3] Alexis Lion,[3,4] Bettina Montaut-Verient,[1] Rémy Grosjean,[1] and Gérome C. Gauchard[1,3,5]

[1]*Department of Otorhinolaryngology, Head and Neck Surgery, University Hospital, 29 avenue du Maréchal de Lattre de Tassigny, 54035 Nancy Cedex, France*
[2]*Faculty of Medicine, Université de Lorraine, 9 avenue de la Forêt de Haye, CS 50184, 54505 Vandoeuvre-lès-Nancy, France*
[3]*EA 3450 DevAH, Development, Adaptation and Disadvantage, Faculty of Medicine, Université de Lorraine, CS 50184, 54505 Vandoeuvre-lès-Nancy, France*
[4]*Sports Medicine Research Laboratory, Luxembourg Institute of Health, 1460 Luxembourg, Luxembourg*
[5]*UFR STAPS, Faculty of Sport Sciences, Université de Lorraine, 30 rue du Jardin Botanique, CS 30156, 54603 Villers-lès-Nancy, France*

Correspondence should be addressed to Gérome C. Gauchard; gerome.gauchard@univ-lorraine.fr

Received 20 April 2015; Revised 30 June 2015; Accepted 9 July 2015

Academic Editor: Luis-Millán González

Many studies were interested in the consequence of vestibular dysfunction related to cochlear implantation on balance control. This pilot study aimed to assess the effects of unilateral cochlear implantation on the modalities of balance control and sensorimotor strategies. Posturographic and vestibular evaluations were performed in 10 patients (55 ± 20 years) with profound hearing loss who were candidates to undergo unilateral multichannel cochlear implantation. The evaluation was carried out shortly before and one year after surgery. Posturographic tests were also performed in 10 age-matched healthy participants (63 ± 16 years). Vestibular compensation was observed within one year. In addition, postural performances of the patients increased within one year after cochlear implantation, especially in the more complex situations, in which sensory information is either unavailable or conflicting. Before surgery, postural performances were higher in the control group compared to the patients' group. One year after cochlear implantation, postural control was close to normalize. The improvement of postural performance could be explained by a mechanism of vestibular compensation. In addition, the recovery of auditory information which is the consequence of cochlear implantation could lead to an extended exploration of the environment possibly favoring the development of new balance strategies.

1. Introduction

Cochlear implantation aims to restore hearing ability and to improve the quality of life (QoL) of patients with severe deafness [1]. The surgical procedure consists in inserting a multielectrode array into the cochlea. Vestibular damage following the surgery is possible due to the anatomical proximity between the vestibular system and the cochlea. Therefore, this insertion may alter the inner ear and may induce vestibular disorders. Indeed, up to 75% of patients undergoing cochlear implant surgery report postoperative vestibular symptoms such as vertigo, dizziness, or imbalance [2–8].

Maintaining equilibrium in upright stance requires the central processing of input signals from the visual, somatokinesthetic, and vestibular systems, leading to a context-specific motor response through the adjustments of static and dynamic postures [9, 10]. Therefore, the alteration of the inner ear by the cochlear implantation may generate postural disorders just after the surgery and after the activation of the cochlear implant [11]. Despite possible inner ear damage induced by the surgery, improvement of the postural stability has been observed even two years after cochlear implantation [2, 12]. Postural improvement after implantation may be induced by a vestibular compensation of a previously uncompensated vestibular lesion [2]. However, several studies showed that postural control of teenagers and adults remains impaired after implantation compared to healthy control subjects, without any effect of hearing restoration even 5 years after implantation [13–16]. The effect on the postural control of the activation of the cochlear implant (CI) has been evaluated in adults between 6 and 8 weeks after surgery and shows that postural control is improved in the more demanding postural situations when the CI is switched on [17].

The characterization of the possible vestibular effects induced by cochlear implantation is well performed by the classical tests (e.g., caloric test, rotary test, and video head impulse test) [6, 7]. The current absence of consensus concerning the outcome of cochlear implantation on postural control does not allow drawing any conclusions for their potential effects. This pilot study aimed to assess the effects of cochlear implantation on the modalities of balance control and sensorimotor strategies.

2. Material and Methods

2.1. Participants. This prospective study (one-year follow-up study) was conducted at the Nancy University Hospital (France) and involved 10 patients (CI group, median age = 55 ± 20 years, age range = 27 to 72 years) with profound hearing loss scheduled to receive unilateral multichannel cochlear implant (Oticon Medical-Neurelec, Vallauris, France/Cochlear Macquarie University, New South Wales, Australia) and 10 healthy participants (control group, age median = 63 ± 16 years, age range = 24 to 71 years). The indication for unilateral cochlear implantation was bilateral profound sensorineural hearing loss with no benefit from hearing aids. All participants were free from any central nervous system disease and presented no orthopedic disorders either of the trunk or the lower limbs that could affect postural performance. All participants gave written consent prior to this pilot study. This pilot study was conducted to examine the feasibility of this approach used in a larger scale study which was approved by the local ethics committee (*Comité de Protection des Personnes de Lorraine*).

2.2. Data Collection. All patients were submitted to hearing, gaze control, and posturographic evaluations two days before and one year after unilateral cochlear implantation. Healthy participants, who were free from ENT disorders, were only submitted to posturographic testing.

2.2.1. Hearing Performance Evaluation. Lafon's cochlear lists, which were used to evaluate hearing ability [1], are composed of 17 triphonemic French words; the percentage of phonemes recognized among the 51 phonemes present in the lists gives an intelligibility score (hearing performance) for each subject. Scores without lip-reading were obtained in quiet at 70 dB SPL. One year after surgery, CI users were tested with their own processor programmed with their normal everyday processing strategy and electrode configuration.

2.2.2. Visuooculomotor and Vestibular Evaluation. The visuooculomotor and vestibular assessments were performed with videonystagmography (Synapsys, Marseille, France) (see details in [18, 19]). The visuooculomotor tests included the evaluation of smooth pursuit and saccades. Spontaneous nystagmus without visual fixation (in darkness) was also recorded. The caloric test, which remains the gold standard to evaluate the degree of vestibular asymmetry, was performed according to the bithermal caloric test protocol, with water infusion of the ear canal for 30 seconds. The subject was in the supine position with the head flexed 30°. Vestibular areflexia was characterized by an absence of caloric response on the side of the affected labyrinth, while vestibular hyporeflexia was determined by a decreased response of more than 20% on the side of the affected labyrinth [20]. For the rotary test, the patients sat in a rotary chair (MED4, Synapsys) with opened eyes in a dark room, and the plane of the lateral semicircular canal was positioned perpendicular to the axis of rotation. The rotary chair test protocol was a pendular test, consisting in seven sinusoidal oscillations around an earth-vertical axis at 0.22 Hz frequency, with increasing then decreasing amplitude. The highest instantaneous velocity of the stimulus was about 30°/s and the highest oscillation amplitude about 30°. Fourier analyses were performed to calculate both the slow phase eye velocity and the chair velocity. The gain measurements of the vestibuloocular reflex were determined by the ratio of the amplitudes of the eye velocity to that of the chair velocity. The directional preponderance measurements were determined by the mean slow phase eye velocity over the duration of the stimulus.

2.2.3. Postural Control Evaluation. The sensory organization test (SOT, EquiTest, Clackamas, OR) aims to evaluate a subject's ability to maintain balance control in six different conditions which are repeated three times during 20 seconds. During these trials, the displacements of the center of foot pressure are recorded. Postural control is challenged by using a technique commonly referred to as sway-referenced, which involves tilting the support surface and/or the visual surround to directly follow the anterior-posterior sways of the subject's center of gravity [21]. Vision is not challenged in conditions 1 (C1) and 4 (C4). Eyes are closed in conditions 2 (C2) and 5 (C5). The visual surround may move in conditions 3 (C3) and 6 (C6). The support surface may move in conditions 4 to 6. These six conditions increase in difficulty and were not randomized. The theoretical limit of stability is based on the individual's height and size of the base of support. The following formula was used to calculate the equilibrium score: $[12.5^\circ - ((\theta_{max} - \theta_{min})/12.5^\circ)] \times 100$, where θ_{max} indicates the greatest anteroposterior center of gravity sway angle displayed by the subject while θ_{min} indicates the lowest anteroposterior center of gravity sway angle. Lower sways lead to a higher score, indicating a better balance control performance (a score of 100 represents no sway, while 0 indicates sway that exceeds the limit of stability, resulting in a fall). Equilibrium scores (ES) were calculated for every condition ($C1^{ES}$ to $C6^{ES}$). A composite equilibrium score (C^{ES}) was calculated by adding the average scores from conditions

Table 1: Characteristics of the ten CI patients.

Patient number	Gender	Etiology of deafness	Age when implanted (in years)	Deafness duration (in years)	CI side	Caloric test		Hearing performance (% of phonemes recognized)	
						Presurgery	Postsurgery	Presurgery	Postsurgery
1	Female	Unknown	58	55	R	N	N	0	90
2	Female	Unknown	39	35	R	RH	RH	0	100
3	Female	Unknown	42	0.5	L	LH	LH	0	50
4	Female	Unknown	52	22	L	N	N	0	60
5	Female	Unknown	69	2	R	N	N	0	80
6	Female	Genetic	59	2	R	RA	RA	0	100
7	Female	Cogan's syndrome	27	2	L	N	N	10	100
8	Male	Unknown	52	18	R	N	N	10	60
9	Male	Otosclerosis	72	39	R	RH	RH	0	90
10	Male	Otosclerosis	66	8	L	RH	RH	0	100

CI side: L, left; R, right. Caloric test: LH, left hyporeflexia; N, normoreflexia; RA, right areflexia; RH, right hyporeflexia.

1 and 2 and the ES from each trial of conditions 3 to 6, and finally dividing that sum by the total number of trials [21–23].

The participants were requested to stand upright and barefoot at mark level on the support surface to control stance width, remaining as still as possible, breathing normally, and being with their arms at their sides. They were instructed to look straight ahead at a picture located on the visual surround. To protect against falls, an operator stood within reaching distance of the participant and all wore a safety harness connected to the ceiling by two suspension straps in all test conditions.

2.3. Statistical Analysis. Qualitative data were expressed as percentages (%) and compared using Fisher's exact test. Quantitative data were expressed as median with interquartile range and compared using nonparametric statistical tests. The Wilcoxon test was performed to compare the results observed before and one year after cochlear implantation. A Mann-Whitney test was performed to compare the results between the CI and control groups. Statistically significant differences were accepted for a probability level of $P < 0.05$.

3. Results

3.1. Participants. Gender, etiology of deafness, age at implantation, deafness duration, side of implantation, hearing performances before and after cochlear implantation, and vestibular status before and after cochlear implantation are presented in Table 1. The implantation was performed on the right side in 7 patients and on the left side in 3 patients by the same surgeon. The CI was inserted unilaterally via the round window surgical technique for one patient and via a cochleostomy procedure for the nine remaining patients. Data on age, gender, height, weight, and body mass index for patients and controls are presented in Table 2 and no significant difference was observed between the two groups for all these parameters.

Table 2: Age and anthropometric characteristics, expressed in median associated with interquartile range (IQR), observed in cochlear implant patients (CI group) and in healthy participants (control group).

	CI group ($n = 10$)	Control group ($n = 10$)	P values*
Women, n (%)	7 (70%)	7 (70%)	NS
Age (years), median (IQR)	55.0 (20.0)	63.0 (16.0)	NS
Height (m), median (IQR)	1.67 (0.05)	1.65 (0.05)	NS
Weight (kg), median (IQR)	64.0 (30.0)	57.5 (9.0)	NS
Body mass index (kg/m^2), median (IQR)	23.5 (7.2)	21.1 (3.0)	NS

* P values for Fisher exact test or Mann-Whitney tests. NS: non-significant.

3.2. Visuooculomotor and Vestibular Evaluation. No participant declared vertigo or dizziness. The visuooculomotor tests showed that smooth pursuit and saccades were normal for all the patients. For the gaze test, only one patient presented a spontaneous nystagmus before cochlear implantation. One year after surgery, no spontaneous nystagmus was observed. Before surgery, the caloric test showed that 1 patient had vestibular areflexia and 4 patients had vestibular hyporeflexia (Table 1). The visual fixation inhibited the nystagmus induced by the caloric test in all these patients. The surgery did not modify the vestibular status of the patients (Figure 1(a)). For the rotatory chair test, the gain (Figure 1(b)) was higher one year after cochlear implantation than before surgery ($P = 0.005$), whereas no difference was observed for the directional preponderance (Figure 1(c)).

3.3. Evolution of Postural Control within One Year in CI Patients. The evolution of postural control within one year in CI patients is showed in Figure 2. Postural control improved one year after surgery compared to before surgery. Indeed, C^{ES} was higher one year after surgery ($P = 0.021$), mainly

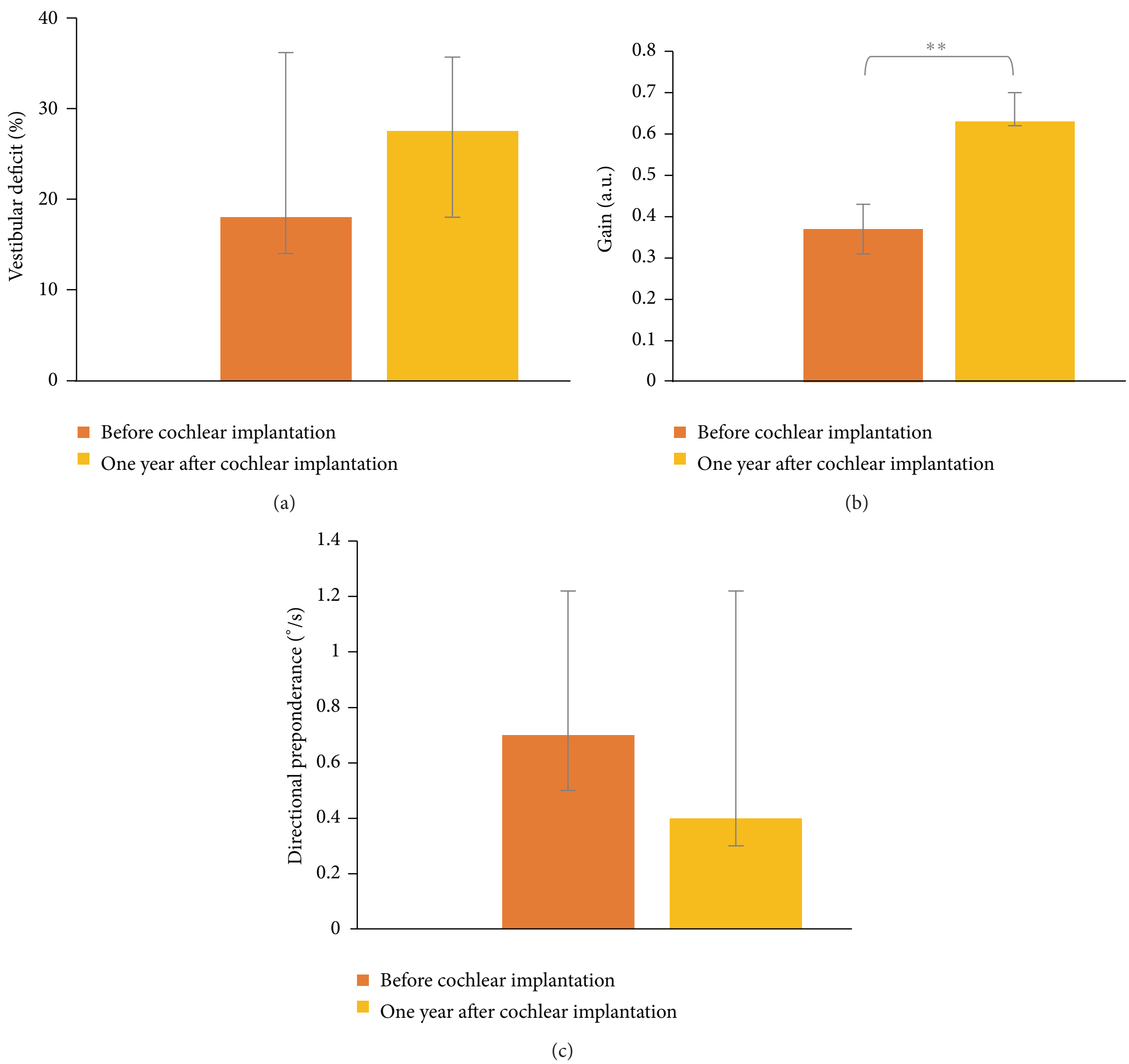


Figure 1: Median values and interquartile ranges of the vestibular status from the caloric test (a) and of the gain (b) and directional preponderance (c) from the rotatory chair test observed in CI group before (orange bars) and one year after cochlear implantation (yellow bars). $^{**}P < 0.01$.

because of higher performances in C3 ($P = 0.025$), C4 ($P = 0.033$), and C5 ($P = 0.043$). In addition, postural improvement was also observed for the patient who had vestibular areflexia (C^{ES} varying from 27 to 65% within one year).

3.4. Comparison of Postural Control between CI and Control Groups. Before surgery (Figure 2), CI patients had altered postural performances. Indeed, C^{ES} was lower in the CI group before surgery than in the control group ($P = 0.010$), which mainly resulted from lower performances in C1 ($P = 0.008$), C3 ($P = 0.013$), C5 ($P = 0.015$), and C6 ($P = 0.008$). One year after surgery, only one difference was observed between the CI group and the control group. Indeed, postural performances normalized in all postural conditions, except for C6 which remained significantly different between the two groups ($P = 0.019$).

4. Discussion

This prospective study showed that patients with unilateral cochlear implants displayed an improvement of postural performance one year after implantation compared to before surgery (even for the patient who had unilateral vestibular areflexia). The gain at the rotatory chair test, which was low before surgery, increased considerably one year after cochlear implantation. In addition, postural performances, which were altered before surgery especially in the more complex conditions, increased one year after cochlear implantation and reached the performances observed in the control group. However, a difference is still observed one year after cochlear implantation between the CI and control groups in the most challenging condition characterized by the possible simultaneous movements of the visual surrounding and the support surface.

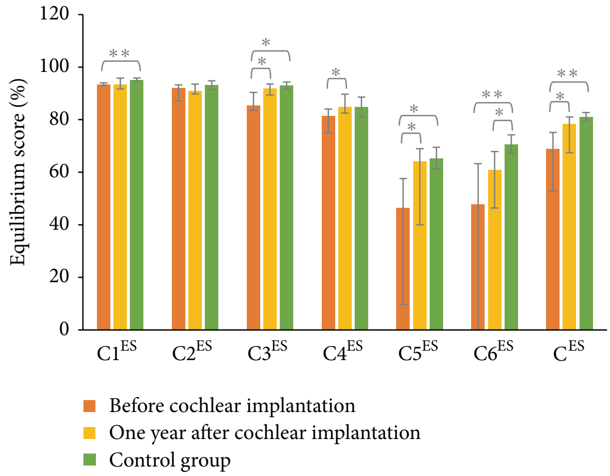


Figure 2: Sensory organization test: median values and interquartile ranges of the equilibrium scores (ES, in %) for the six conditions ($C1^{ES}$ to $C6^{ES}$) and the composite equilibrium score (C^{ES}) observed in CI group before (orange bars) and one year after cochlear implantation (yellow bars) and in control group (green bars). $^{*}P < 0.05$, $^{**}P < 0.01$.

Deaf patients, who are candidates to cochlear implantation, have normal or decreased vestibular function, vestibular dysfunction in some deaf patients being related to a joint pathology of the posterior and anterior labyrinths [3, 24, 25]. Indeed, CI candidates have often preoperative vertigo symptoms [3]. In our study, CI patients had low preoperative gain at the rotatory chair test, suggesting that vestibular function was altered before surgery and highlighting a low efficiency of vestibular compensation [26]. One hypothesis may be that this low degree of vestibular compensation is one explanatory factor of low preoperative postural performance, which is in accordance with previous observations by Magnusson et al. [25]. According to the theoretical framework of perception-action, the brain receives information from the various sensory afferents to produce movement, and the action determines the perception [27]. Applied to vestibular pathology, two things are required in order to compensate. First, the brain must receive signals from the balance organs. Thus, movements must not be avoided, because movements create the signals which are required by the brain to compensate for the injury. Second, the balance areas of the brain must be capable of change. In CI candidates, the brain is usually not damaged and is capable of plasticity. Therefore, the failure of preoperative compensation is probably induced by the isolation and the restriction of activity related to the deafness [28]. One year after cochlear implantation, patients showed no degradation of their postural performances in spite of the introduction of the electrode array in the cochlea which may increase the risk of the vestibular asymmetry to the side of the implantation. The higher values of the gain and the lower values of directional preponderance, observed in the rotatory chair test one year after cochlear implantation, demonstrate the efficiency of the vestibular compensation. This vestibular compensation is associated with a postural compensation highlighted by the absence of degradation of postural performance one year after cochlear implantation. As it had been suggested in other kinds of vestibular dysfunction, for example, in the case of vestibular schwannoma, the time-course of implementation of central adaptive mechanisms, characterized by substitution by other sensory afferences and new behavioral strategies, could lead to a recovery of balance control with an improvement in balance performance [29, 30], which are close to normal and are difficult to decompensate [31].

One year after cochlear implantation, the postural performances did not decrease. Conversely, the performances improved significantly and tended to reach normal performances. The vestibular compensation could not explain alone the major improvement of the postural performances. The restoration of the auditory information could contribute to balance regulation in CI patients according to two complementary hypotheses. On the one hand, the recovery of hearing may lead to a reorientation of CI patients in their surroundings. Indeed, equilibrium function relies on the control of posture but also on spatial orientation. The spatial orientation of the body allows a subject to perceive his/her environment and to act particularly in case of movement or destabilization [32]. Posture can thus be considered as a primary support for action [9]. Although auditory information is not considered as a fundamental signal involved in balance control, the auditory system is a perceptual system which, with vision and touch, is involved in the perception of the dynamic environment and in complex representations of 3D space. In the same way as sensory inputs are redundant and complementary to fine-tune postural control during sensory conflict situations [18], the ability to use multiple senses to perceive environmental characteristics allows a more relevant detection of objects and events, leading to an accurate orientation behaviour [33]. Indeed, recent studies showed that multisensory cues were more effective at capturing spatial attention than unimodal cues under conditions of concurrent perceptual tasks [34, 35]. The recovery of auditory function in environmental perception and orientation references after cochlear implantation could initiate motor learning and, in this way, allow the implementation of new neural networks, leading to new sensorimotor and behavioral strategies. These new sensorimotor and behavioral strategies could explain the improvement of postural control efficiency observed here, especially in the more complex postural situations. CI patients displayed new balance abilities. This could be related to the recovery of auditory information, which may participate in spatial orientation through interactions with the visual signals [36, 37]. On the other hand, cochlear implantation is known to induce an improvement in the quality of life (QoL). A multiple stepwise regression analysis showed that, after implantation, improvements in communication abilities, reduced psychological problems, and improvements of daily life abilities were key determinants to QoL improvements for CI patients [1, 38]. The QoL increase reaches a plateau at about 1.5 to 3 years after implantation [39]. The recovery of hearing may therefore reduce isolation, depressive state, restriction of activity, and the breakdown of familial or other social occupational activities. These factors could influence vestibular and postural compensations. Therefore, CI patients may interact more with the environment which may lead to enhanced sensory-motor stimulations.

The minor remaining perturbations of postural control observed one year after surgery in CI patients indicate that postural compensation is close to be completed and could justify a longer follow-up. Indeed, balance performances still improved one year after unilateral vestibular deafferentation and this improvement could be related to a reinforcement of the newly elaborated sensorimotor and behavioral strategies due to daily life activities [19]. Thus, postural performances might need more than one year to reach age-matched control level in CI patients, who have to perform simultaneously several learning processes, such as auditory rehabilitation with speech recognition learning, vestibular compensation after vestibular function degradation related to the introduction of the electrode array in the cochlea, and finally new orienting behaviour learning process after recovery of auditory perception of dynamic environment.

The main limitation of this study is the possible postural learning effect that could be observed between the two measurements in the CI group. The learning effect has been observed to last at least three months [40, 41]. However, this hypothesis is not convincing according to the sensorimotor modifications induced by surgical approach and the delay (one year) between the two posturographic evaluations. Therefore, the learning effect is probably limited. In addition, the test-retest reliability of the composite and equilibrium scores has been shown to be fairly good [42]. Another limitation of this study is the sound from the engines and from the movements of the screen or the platform during the sway-referenced conditions. It cannot be excluded that the CI patients have used these sounds to limit their movements. In addition, the EquiTest used the most extreme data samples to evaluate the postural control; this is therefore flawed as it is not fully representative of the average stability. However, this device is fairly convenient to use in a clinical setting. Finally, the study did not investigate the orientation ability of the CI patients. The results of this study should be interpreted in light of these limitations and further prospective studies are needed especially with the standardization of the CI patients according to the origin of the hearing loss and the recovery of hearing after cochlear implantation.

5. Conclusions

Unilateral cochlear implantation is not harmful for postural performances. Conversely, postural stability improves within one year after the cochlear implantation. This improvement could be an indirect consequence of the recovery of auditory information. Indeed, patients may be less dependent and move more, strengthening postural control. The increase of postural control performances could also take an important part in the improvement of the quality of life observed in CI patients. Vestibular and postural evaluations are important in the follow-up of the cochlear implantation. The vestibular and postural improvements should be taken into account in the decision-making process of the cochlear implantation.

Conflict of Interests

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

Acknowledgment

The authors would like to thank the Clinical and Scientific Research Department of Oticon Medical for editorial help during the finalization of the paper.

References

[1] C. Rumeau, J. Frère, B. Montaut-Verient, A. Lion, G. Gauchard, and C. Parietti-Winkler, "Quality of life and audiologic performance through the ability to phone of cochlear implant users," *European Archives of Oto-Rhino-Laryngology*, 2014.

[2] C. A. Buchman, J. Joy, A. Hodges, F. F. Telischi, and T. J. Balkany, "Vestibular effects of cochlear implantation," *Laryngoscope*, vol. 114, no. 10, supplement 103, pp. 1–22, 2004.

[3] E. Krause, J. P. R. Louza, J.-M. Hempel, J. Wechtenbruch, T. Rader, and R. Gürkov, "Prevalence and characteristics of preoperative balance disorders in cochlear implant candidates," *Annals of Otology, Rhinology and Laryngology*, vol. 117, no. 10, pp. 764–768, 2008.

[4] E. Krause, J. P. R. Louza, J.-M. Hempel, J. Wechtenbruch, T. Rader, and R. Gürkov, "Effect of cochlear implantation on horizontal semicircular canal function," *European Archives of Oto-Rhino-Laryngology*, vol. 266, no. 6, pp. 811–817, 2009.

[5] E. Krause, J. P. R. Louza, J. Wechtenbruch, and R. Gürkov, "Influence of cochlear implantation on peripheral vestibular receptor function," *Otolaryngology—Head and Neck Surgery*, vol. 142, no. 6, pp. 809–813, 2010.

[6] D. Vibert, R. Häusler, M. Kompis, and M. Vischer, "Vestibular function in patients with cochlear implantation," *Acta Oto-Laryngologica, Supplementum*, vol. 545, pp. 29–34, 2001.

[7] A. Batuecas-Caletrio, M. Klumpp, S. Santacruz-Ruiz, F. B. Gonzalez, E. G. Sánchez, and M. Arriaga, "Vestibular function in cochlear implantation: correlating objectiveness and subjectiveness," *The Laryngoscope*, 2015.

[8] T.-A. N. Melvin, C. C. D. Santina, J. P. Carey, and A. A. Migliaccio, "The effects of cochlear implantation on vestibular function," *Otology and Neurotology*, vol. 30, no. 1, pp. 87–94, 2009.

[9] J. Massion and M. H. Woollacott, "Posture and equilibrium," in *Balance, Posture and Gait*, A. M. Bronstein, T. Brandt, and M. H. Woollacott, Eds., pp. 1–18, Arnold, London, UK, 1996.

[10] C. Maurer, T. Mergner, and R. J. Peterka, "Multisensory control of human upright stance," *Experimental Brain Research*, vol. 171, no. 2, pp. 231–250, 2006.

[11] R. Filipo, M. Patrizi, R. La Gamma, C. D'Elia, G. La Rosa, and M. Barbara, "Vestibular impairment and cochlear implantation," *Acta Oto-Laryngologica*, vol. 126, no. 12, pp. 1266–1274, 2006.

[12] P. A. Abramides, R. S. Bittar, R. K. Tsuji, and R. F. Bento, "Caloric test as a predictor tool of postural control in CI users," *Acta Oto-Laryngologica*, vol. 135, no. 7, pp. 685–691, 2015.

[13] H.-D. Kluenter, R. Lang-Roth, and O. Guntinas-Lichius, "Static and dynamic postural control before and after cochlear implantation in adult patients," *European Archives of Oto-Rhino-Laryngology*, vol. 266, no. 10, pp. 1521–1525, 2009.

[14] L. Bernard-Demanze, J. Léonard, M. Dumitrescu, R. Meller, J. Magnan, and M. Lacour, "Static and dynamic posture control in postlingual cochlear implanted patients: effects of dual-tasking, visual and auditory inputs suppression," *Frontiers in Integrative Neuroscience*, vol. 7, article 111, 2014.

[15] H. Suarez, S. Angeli, A. Suarez, B. Rosales, X. Carrera, and R. Alonso, "Balance sensory organization in children with profound hearing loss and cochlear implants," *International Journal of Pediatric Otorhinolaryngology*, vol. 71, no. 4, pp. 629–637, 2007.

[16] M.-W. Huang, C.-J. Hsu, C.-C. Kuan, and W.-H. Chang, "Static balance function in children with cochlear implants," *International Journal of Pediatric Otorhinolaryngology*, vol. 75, no. 5, pp. 700–703, 2011.

[17] B. Schwab, M. Durisin, and G. Kontorinis, "Investigation of balance function using dynamic posturography under electrical-acoustic stimulation in cochlear implant recipients," *International Journal of Otolaryngology*, vol. 2010, Article ID 978594, 7 pages, 2010.

[18] C. Parietti-Winkler, G. C. Gauchard, C. Simon, and P. P. Perrin, "Visual sensorial preference delays balance control compensation after vestibular schwannoma surgery," *Journal of Neurology, Neurosurgery and Psychiatry*, vol. 79, no. 11, pp. 1287–1294, 2008.

[19] C. Parietti-Winkler, G. C. Gauchard, C. Simon, and P. P. Perrin, "Long-term effects of vestibular compensation on balance control and sensory organisation after unilateral deafferentation due to vestibular schwannoma surgery," *Journal of Neurology, Neurosurgery and Psychiatry*, vol. 81, no. 8, pp. 934–936, 2010.

[20] J. Bergenius and M. Magnusson, "The relationship between caloric response, oculomotor dysfunction and size of cerebello-pontine angle tumours," *Acta Oto-Laryngologica*, vol. 106, no. 5-6, pp. 361–367, 1988.

[21] L. M. Nashner and J. F. Peters, "Dynamic posturography in the diagnosis and management of dizziness and balance disorders," *Neurologic Clinics*, vol. 8, no. 2, pp. 331–349, 1990.

[22] F. O. Black, W. H. Paloski, D. D. Doxey-Gasway, and M. F. Reschke, "Vestibular plasticity following orbital spaceflight: recovery from postflight postural instability," *Acta Oto-Laryngologica*, vol. 115, no. 520, pp. 450–454, 1995.

[23] G. C. Gauchard, G. Vançon, P. Meyer, D. Mainard, and P. P. Perrin, "On the role of knee joint in balance control and postural strategies: effects of total knee replacement in elderly subjects with knee osteoarthritis," *Gait and Posture*, vol. 32, no. 2, pp. 155–160, 2010.

[24] F. O. Black, "Present vestibular status of subjects implanted with auditory prostheses," *The Annals of Otology, Rhinology & Laryngology. Supplement*, vol. 86, no. 3, part 3, supplement 38, pp. 49–56, 1977.

[25] M. Magnusson, H. Petersen, S. Harris, and R. Johansson, "Postural control and vestibular function in patients selected for cochlear implantation," *Acta Oto-Laryngologica. Supplementum*, vol. 520, part 2, pp. 277–278, 1995.

[26] C. Parietti-Winkler, G. C. Gauchard, C. Simon, and P. P. Perrin, "Pre-operative vestibular pattern and balance compensation after vestibular schwannoma surgery," *Neuroscience*, vol. 172, pp. 285–292, 2011.

[27] A. Berthoz, *Le Sens du Mouvement*, Odile Jacob, Paris, France, 1997.

[28] J. B. Hinderink, P. F. M. Krabbe, and P. van den Broek, "Development and application of a health-related quality-of-life instrument for adults with cochlear implants: the Nijmegen cochlear implant questionnaire," *Otolaryngology—Head and Neck Surgery*, vol. 123, no. 6, pp. 756–765, 2000.

[29] C. Parietti-Winkler, G. C. Gauchard, C. Simon, and P. P. Perrin, "Sensorimotor postural rearrangement after unilateral vestibular deafferentation in patients with acoustic neuroma," *Neuroscience Research*, vol. 55, no. 2, pp. 171–181, 2006.

[30] N. Uehara, H. Tanimoto, T. Nishikawa et al., "Vestibular dysfunction and compensation after removal of acoustic neuroma," *Journal of Vestibular Research: Equilibrium and Orientation*, vol. 21, no. 5, pp. 289–295, 2011.

[31] G. Dumas, A. Lion, G. C. Gauchard, G. Herpin, M. Magnusson, and P. P. Perrin, "Clinical interest of postural and vestibulo-ocular reflex changes induced by cervical muscles and skull vibration in compensated unilateral vestibular lesion patients," *Journal of Vestibular Research: Equilibrium & Orientation*, vol. 23, no. 1, pp. 41–49, 2013.

[32] H. Ceyte, C. Cian, R. Zory, P.-A. Barraud, A. Roux, and M. Guerraz, "Effect of Achilles tendon vibration on postural orientation," *Neuroscience Letters*, vol. 416, no. 1, pp. 71–75, 2007.

[33] J. X. Maier and J. M. Groh, "Multisensory guidance of orienting behavior," *Hearing Research*, vol. 258, no. 1-2, pp. 106–112, 2009.

[34] C. L. Folk, E. F. Ester, and K. Troemel, "How to keep attention from straying: Get engaged!," *Psychonomic Bulletin and Review*, vol. 16, no. 1, pp. 127–132, 2009.

[35] C. Ho, V. Santangelo, and C. Spence, "Multisensory warning signals: when spatial correspondence matters," *Experimental Brain Research*, vol. 195, no. 2, pp. 261–272, 2009.

[36] H. Colonius and P. Arndt, "A two-stage model for visual-auditory interaction in saccadic latencies," *Perception and Psychophysics*, vol. 63, no. 1, pp. 126–147, 2001.

[37] B. D. Corneil, M. Van Wanrooij, D. P. Munoz, and A. J. Van Opstal, "Auditory-visual interactions subserving goal-directed saccades in a complex scene," *Journal of Neurophysiology*, vol. 88, no. 1, pp. 438–454, 2002.

[38] S. M. Cohen, R. F. Labadie, M. S. Dietrich, and D. S. Haynes, "Quality of life in hearing-impaired adults: the role of cochlear implants and hearing aids," *Otolaryngology—Head and Neck Surgery*, vol. 131, no. 4, pp. 413–422, 2004.

[39] F. Zhao, Z. Bai, and D. Stephens, "The relationship between changes in self-rated quality of life after cochlear implantation and changes in individual complaints," *Clinical Otolaryngology*, vol. 33, no. 5, pp. 427–434, 2008.

[40] F. Tjernström, A. Bagher, P.-A. Fransson, and M. Magnusson, "Short and long-term postural learning to withstand galvanic vestibular perturbations," *Journal of Vestibular Research: Equilibrium and Orientation*, vol. 20, no. 6, pp. 407–417, 2010.

[41] F. Tjernström, P.-A. Fransson, and M. Magnusson, "Improved postural control through repetition and consolidation," *Journal of Vestibular Research: Equilibrium and Orientation*, vol. 15, no. 1, pp. 31–39, 2005.

[42] D. M. Wrisley, M. J. Stephens, S. Mosley, A. Wojnowski, J. Duffy, and R. Burkard, "Learning effects of repetitive administrations of the sensory organization test in healthy young adults," *Archives of Physical Medicine and Rehabilitation*, vol. 88, no. 8, pp. 1049–1054, 2007.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 463792, 10 pages
http://dx.doi.org/10.1155/2015/463792

Clinical Study

Recovery of Posture Stability at Different Foot Placements in Patients Who Underwent Minimally Invasive Total Hip Arthroplasty: A One-Year Follow-Up Study

Chun-Ju Chang,[1,2] Na-Ling Lin,[1] Mel S. Lee,[3,4] and Jen-Suh Chern[1]

[1] *Department of Occupational Therapy and Graduate Institute of Clinical Behavioral Science, Chang Gung University, Taoyuan 333, Taiwan*
[2] *Department of Biomedical Engineering, National Yang Ming University, Taipei 112, Taiwan*
[3] *Department of Orthopaedic Surgery, Kaohsiung Chang Gung Memorial Hospital, Kaohsiung 833, Taiwan*
[4] *Department of Medicine, Chang Gung University, Taoyuan 333, Taiwan*

Correspondence should be addressed to Jen-Suh Chern; jschern@mail.cgu.edu.tw

Received 23 April 2015; Revised 30 June 2015; Accepted 5 August 2015

Academic Editor: Luis-Millán González

To understand the progression of recovery in postural stability and physical functioning after patients received the minimally invasive total hip arthroplasty (MTHA), we monitor the pain level, functional capacity, and postural stability before and after operation within one year. In total of 23 subjects in our study, we found out that MTHA was effective in relieving pain in first 2 weeks and restoring the hip joint integrity, but the postural stability was influenced especially in tandem stand in both anterior-posterior and medial-lateral directions. The recovery of postural stability and functional capacity in one year duration fluctuated and no consistent improvement tendency was found. We suggested clinicians designing postsurgery rehabilitation program for consistent and progressive long-term recovery of postural stability and fall prevention to optimize surgical results and prevent undesired postoperative consequences.

1. Introduction

Hip joint is one of the two most mobile joints in the human body. Due to its location in the body and upright dominated human posture, hip joint not only has to fulfill mobility demands but also has to be able to bear multidirectional loads constantly. Therefore, it is one of the most vulnerable joints developing degenerative osteoarthritis (OA) [1]. The destruction of hip joint due to OA is irreversible and total hip arthroplasty (THA) has been developed several decades ago to reconstruct the joint structure [2, 3]. Advancement of surgical technique to minimize physiological impact during surgical process has been the focus for the recent decade. Minimally invasive THA (MTHA) was, thus, invented and gradually replaces traditional THA [4].

Studies have shown that both THAs are effective in reducing joint pain and MTHA was superior to traditional THA in terms of decreasing blood loss during surgery, decreasing length of hospital stay, fastening walking speed, and daily functional independence [2–4]. However, cross-sectional studies reported subnormal hip joint function even at 24 months after MTHA. However, the impact of MTHA on relieving joint pain, improving functional capacity, and improving postural stability across a one-year period has not been studied [4–6]. It is reasonable to hypothesize that insufficient recovery and lacking of appropriate postsurgery intervention [6, 7] might contribute to the reported incidence of dislocation of the newly replaced join, development of ipsilateral hip OA, and falls after MTHA. Understanding of the progression of recovery of functional capacity and postural stability after MTHA is critical in lengthening the durability of the prostheses, preventing ipsilateral hip joint pathology, and increasing daily living safety [3].

Frequently used outcome measure for MTHA includes self-report visual analogue pain scale (VAS), functional reach tests, Berg Balance Scale, Activities-Specific Balance

Confidence Scale [8–11], and self-administered hip-rating questionnaire. Postural stability is unique outcome measures and was used often to quantify the risk of fall in daily living context [12].

Quiet stance is an easy and safe task to perform, compared to level walking for patients with acute and severe unilateral neuromuscular and/or musculoskeletal impairment in lower extremity. Postural stability during quiet stance has been well investigated and is recognized as a dynamic motor control process involving active sensory processing with a constant mapping of perception to action [13]. Somatosensory apparatus located in the neck, trunk, and lower extremities is inevitable in sensing stance perturbation information through upward neural pathway, while trunk and lower extremity muscles are modulated by downward neural pathway formulating matched muscular synergy as postural actions to control stance posture [14]. Efficient postural stability maintains the vertical projection of the center of mass (COM), which is center of gravity (COG), within the configuration of base of support (BOS) to prevent falls [15, 16].

Stance postural stability is frequently quantified by amount of postural sway and patterns of weight distribution over two lower limbs (symmetry versus asymmetry) [13]. The parameters derived from postural sway and weight distribution are the results of ankle and hip movement strategies which actuated through bilateral lower limb loading and unloading mechanism (LULM). It is well accepted that increment of postural sway and asymmetry weight distribution reflect LULM impairments, which in turn indicated insufficient somatosensory and neuromuscular functioning in the lower extremity [13, 15, 16]. However, Chern et al. had suggested that compensatory weight bearing strategy might be developed along with the somatosensory and neuromuscular recovery process to fulfill the priority of daily living safety in patients with unilateral lower limb pathology, such as hemiplegia after stroke, and this compensatory strategy might decrease postural instability [17].

Stance with different foot placements is not uncommon in our daily situation. Foot placement configures the dimension of BOS in both anterior-posterior (AP) and medial-lateral (ML) planes [12]. Changing foot placement perturbs stability through redistributing body mass and deviating COG locations in the dimensions. Therefore, quiet stance postural control features in different foot placement are suitable for and valid in documenting the efficiency of somatosensory and neuromuscular functions of the lower extremity [17], like tandem stand or foot together stand.

With a little knowledge of the progression of functional recovery after receiving the MTHA, it is necessary to monitor the pain level, activity of daily life, and posture stability after the surgery for prevention of falls in daily context and decreasing the need of medical care following falls. The purpose of this study is to investigate pain level, functional capacity, and postural stability in patients with MTHA at pre- and postoperative stages. The postural stability was measured while the subjects stood with four different foot placement conditions. Our hypothesis is that the patients' level of pain, functional capacity, and postural stability would recover progressively at the time immediately after surgery through one year after surgery. The second hypothesis was that the subjects' postural stability would vary as a function of the foot placements which post graded demands on control of postural stability. Therefore, multiple functional and balance measurements were taken before and after the surgery. The results of the study would advise clinicians in designing presurgery education and postsurgery intervention programs for this group of patients to optimize surgical results and prevent undesired postoperative consequences, such as falls.

2. Materials and Methods

2.1. Participants. Convenient and intentional sampling method was used to recruit participants in this study. A total of 23 participants (10 male and 13 female, mean age of 60 ± 9.9 years, mean height of 165.6 ± 9.7 cm, mean weight of 58.2 ± 11.9 kg, and mean BMI of 23.55 ± 3.50 kg/m^2) agreed to participate in this study. The inclusion criteria were (1) being diagnosed as unilateral hip arthritis, (2) planning to accept total hip arthroplasty (THA) procedure, (3) being able to stand independently for 3 minutes before surgery, and (4) being able to follow verbal instructions. Comorbidities of the lower extremity, such as osteoarthritis or misalignment at other joints, and other systematic diseases, which might affect stance postural control capability, were the exclusion criteria. A sample size of 23 was justified by statistical power of 0.96 calculated based on an effect size of 0.5. The experimental procedure was approved by Chang Gung Medical Foundation Institutional Review Board (number 97-1389B) and all participants signed informed consent form prior to entering into this study.

2.2. Total Hip Arthroplasty. All THAs were performed by an experienced orthopaedic surgeon (the third author) at a medical center, Department of Orthopedic Surgery. All the THAs are minimally invasive but with slight variation in the number and location of incisions. The surgeon chose one of the three procedures based on his professional judgments and patient's choice of hip joint prostheses: (1) modified Watson-Jones minimally invasive approach (MIS-WJ), (2) minimally invasive anterolateral approach procedure (MIS-AL), and (3) two-incision minimally invasive procedure (MIS-2). Incisions were made at either anterior, anterolateral, or posterior aspect of the hip joint. MIS-WJ requires exposure of the hip joint through the muscular intervals [18, 19]; MIS AL involves dissecting of 20–25% of gluteus medius between anterior border of gluteus medius and tensor fasciae latae. All procedures involved dissection across and blunt dissection of the gluteus maximus to create space for orthotic stem implantation [20–22]. All the incisions were within 10 centimeters long with the shortest incision of 2–6 cm in MIS-2 approach [23–26].

2.3. Experiment Procedures and Instrumentations. After signed informed consent form, participants were arranged for measurement of pain level, functional capacity of the hip joint, and postural stability as the dependent variables. All measurements took place at 1 day before surgery and 2 weeks, 6 weeks, 3 months, 6 months, and 1 year after surgery. The time point and the foot placement described as follows would be the independent variables in our study. Pain level was measured using visual analogue pain scale (VAS, 1: no pain at all, 10: extreme pain). Functional capacity was measured by functional reach tests (FRT), Berg Balance Scale (BBS), self-administered hip-rating questionnaire (hip score), functional independence measure (FIM), and Activities-Specific Balance Confidence Scale (ABC). FRT is a quick screen tool for risk of falls in adults with or without balance deficits. The individual is asked to firstly stand erect with their feet shoulder width apart, secondly elevate one arm at 90° shoulder flexion with hand closed as a fist, and finally slide a yard (which is fixed on the wall at acromion height) as far as possible without moving their feet or losing balance [27]. BBS is a 14-item test with a 4-point rating scale. It is a performance-based test with standardized rating criteria. The total score ranges from 0 to 56 [12]. Hip score is both a self-report questionnaire and subjective measurement tool. It is administered routinely in orthopaedic clinics to monitor progression of hip function for patients with hip problems. The total score ranges from 0 to 100. A score less than 70 indicates poor hip capacity and a score over 90 indicates excellent hip function [28]. FIM is a measure of level of disability in daily living context with a 7-point rating scale. The total score ranges from 18 to 126, and it was used to justify the need of assistance of patients [29]. The ABC is a 16-item test with an 11-point scale and ratings consist of whole numbers (0–100) for each item. The possible range of the item is 0–1600. The total rating is divided by 16 to get each subject's ABC score. ABC score less than 67 indicates a risk of falls and a score more than 80 indicates high level of physical functioning [10]. The psychometric characteristics of all functional tests used in this study were established and testing and rating process were standardized.

To measure postural stability, participants were instructed to stand barefoot with 4 foot placements as shoulder width stance (SWS), feet side-by-side stance (SSS), tandem stance with affected limb in the front (AFS), and tandem stance with nonaffected limb in the front (SFS) on a 0.5 m long pressure measurement mat (RSscan International Co., Belgium) and were instructed to stand as still as possible with both arms at their sides and eyes staring at a target 5 m away in front of them. The pressure mat is 488 mm × 325 mm containing 4096 sensors and the maximum sampling rate is 500 Hz. In each foot placement condition, three trials lasting 30 s (sample frequency 30 Hz) were recorded with 15 s resting interval between trials. To eliminate the unstable data before an individual stands still, the instrument started to collect data at 5 s after the subject stood on the mat. To avoid the order effects, the sequence of four foot placements was randomized.

All patients administered for MTHA received limited postsurgery rehabilitation during either their hospital stay (which is 3 to 5 days) for surgical procedure or outpatient follow-up after surgery because the financial reimbursement of the National Health Insurance System is very limited. Bedside education before and after surgery regarding the hip precautions, such as prohibited motion type and range [30] after MTHA, is given by either nurses or physical therapists in a short counseling session. No regular rehabilitation program was prescribed by surgeons and rehabilitation specialists were not regular members of the team providing medical care for MTHA patients. There is no active training effort initiated or imported by any individual in the team. No more regular rehabilitation was provided after the surgery and the progression of functional recovery in those patients was not monitored.

2.4. Data Processing. Average value and standard deviation of all measures were used for statistical analysis. Stance postural control actions were measured by a pressure measurement mat which was connected with a 3D data processing unit and 2-dimensional coordinates of center of pressure (CoP) in anterior-posterior (AP) and medial-lateral (ML) axis and the amount of weight loaded under each limb was output to a customized written program for calculation of parameters. The parameters frequently used for representing stance postural stability include resultant CoP sway path length (CoPR), maximum CoP sway in ML direction (CoPML), maximum CoP sway in AP direction (CoPAP), and body weight distribution symmetry index (BWDSI). The amplitude of CoP parameters is determined by the movement strategy which is actuated by LULM. Normalization of CoP parameters is not necessary because of repeated measure design in this study. Calculation of BWSDI is shown in formula (1). The closer the BWDSI is to the value of zero, the more symmetrical the bilateral weight distribution over two lower limbs is, indicating that both lower limbs are able to bear body weight equally:

$$\mathrm{BWDSI} = \frac{(\mathrm{Load}_{\mathrm{UO}} - \mathrm{Load}_{\mathrm{O}})}{(\mathrm{Load}_{\mathrm{UO}} + \mathrm{Load}_{\mathrm{O}})} * 100. \tag{1}$$

$\mathrm{Load}_{\mathrm{UO}}$ is body weight loaded over unoperated limb. $\mathrm{Load}_{\mathrm{O}}$ is body weight loaded under operated limb.

2.5. Statistical Analysis. Firstly, the function of boxplot in the MATLAB 2012a (Mathworks, Natick, MA) was used for excluding the outlier values of measured variable. The Shapiro-Wilk test for small sample size was used to confirm the normal distribution. And then, the effects of time points on pain level and functional capacity were examined by one-way repeated measure ANOVA. Interaction effects of foot placements and time points on measures of postural stability were examined by two-way repeated measure ANOVA, and degree of freedom was adjusted based on Mauchly's test of sphericity assumption. When the interaction effects were significant, simple main effects of both time points and foot placement conditions were examined by one-way repeated measure ANOVA. Finally, Tukey's HSD test was used as post hoc analysis to show the difference between time points or the foot placements. Statistical significance was set at the level of $p < 0.05$. Commercialized statistical software SPSS 20.0 for windows package (IBM Corp., Armonk, NY) was used for all statistical analysis.

3. Results

3.1. Primary Outcomes. There was significant difference in BBS, FIM, VAS, and hip scores among time points (Table 1,

Table 1: One-way repeated measure ANOVA summary showing the effects of time points on functional assessment.

Functional assessment	Time points before and after surgery						$p^{\dagger}$
	Presurgery	2 w	6 w	3 M	6 M	1 yr	
Berg balance test (56)	44.6 ± 7.3^{ac}	36.4 ± 12.9^{b}	41.4 ± 7.9^{bc}	47.7 ± 4.7^{ac}	53.2 ± 1.9^{a}	52.5 ± 3.3^{a}	$<0.0001^{**}$
Functional reach test (cm)							
Forward	22.2 ± 6.7	22.6 ± 4.2	24.6 ± 5.3	20.4 ± 5.1	28.7 ± 7.6	27.4 ± 7.5	0.0854
Toward affected side	19.8 ± 4.1	20.9 ± 9.4	19.4 ± 3.6	18.9 ± 3.6	25.7 ± 4.7	22.2 ± 5.2	0.0828
Toward unaffected side	18.8 ± 5.2	18.5 ± 3.2	18.6 ± 4.4	18.3 ± 8.7	25.3 ± 2.8	23.7 ± 4.5	0.0712
ADL independence/pain							
VAS* (10)	4.8 ± 1.4^{a}	2.3 ± 1.6^{b}	1.1 ± 1.2^{b}	1.1 ± 1.0^{b}	1.4 ± 1.0^{b}	2.1 ± 2.3^{b}	$<0.0001^{**}$
Hip score* (100)	63.2 ± 9.3^{a}	65.0 ± 9.8^{a}	74.6 ± 9.8^{b}	83.6 ± 10.3^{bc}	82.7 ± 9.5^{b}	85.0 ± 11.6^{c}	$<0.0001^{**}$
FIM* (126)	119.7 ± 4.6^{a}	110.2 ± 19.6^{b}	118.0 ± 5.8^{ab}	122.3 ± 4.0^{a}	125.7 ± 0.7^{a}	125.8 ± 0.6^{a}	$<0.0001^{**}$

*Visual analogue pain scale (VAS); self-administered hip-rating questionnaire (hip score); functional independence measure (FIM).
†One-way repeated measure ANOVA. Tukey's HSD test for comparison of the period and the same letter means no significant difference.
$^{**}p < 0.01$.

$p < 0.01$). Subsequent analysis showed that level of pain decreased significantly ($p < 0.05$) at 2 weeks postoperatively but increased slightly starting at 6 months postoperatively. The hip function as measured by hip score tended to improve gradually and the highest score of 85.0 ± 11.6 was reached at 1 year after surgery. The daily living function as measured by FIM decreased significantly ($p < 0.05$) at 2 weeks and reached plateau score of 125.8 ± 0.6 at 6 months postoperatively. Berg balance test decreased significantly ($p < 0.05$) at 2 weeks and improved gradually, reaching the highest score of 53.2 ± 1.9 at 6 months. The difference in functional reach test in all three directions was approaching significant level (Table 1, $p = 0.07$–0.08). The descriptive data showed that functional reach distance tended to improve gradually after surgery, reached the maximum distance, and decreased at 1 year postoperatively.

3.2. Effects of Foot Placement on Stance Postural Stability. The interaction effects between foot placement and time point relative to surgery date were significant in all measures (Table 2, $p = 0.000$–0.018), indicating that the effects of foot placement on standing stability depended on the time point when the sway was measured. Subsequent main effect analysis found that foot placement affects CoPR, CoPAP, and CoPML significantly (Table 2, main effect of foot placement factor $p < 0.01$) at both presurgery and postsurgery time points. When the foot placement changes from shoulder width stance (SWS), side-by-side stance (SSS), and nonaffected limb in the front stance (SFS) to affected limb in the front stance (AFS), the subjects tended to increase their postural sway after surgery and the progressive increment of CoPR (Figures 1–4). Descriptive data further showed that the amplitude of CoPR is the least when standing in the position of SSS and the greatest during AFS and SFS position (Table 2). The CoP sway in both the CoPAL and CoPML direction was affected significantly by foot placement especially when the width of BOS was narrowed down significantly by foot placement in tandem stance (Figures 2 and 3).

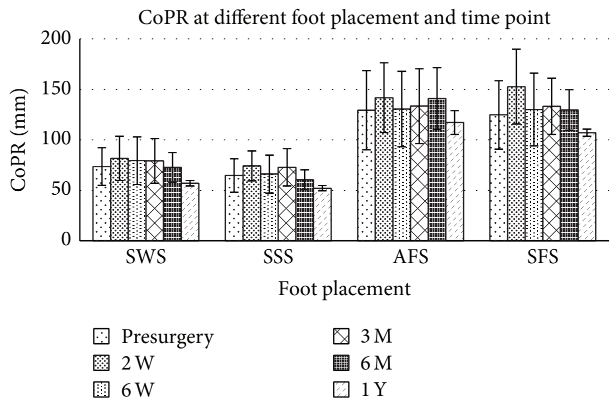


Figure 1: CoP sway path length (CoPR) at different foot placement and time points before and after surgery. Shoulder width stance (SWS), feet side-by-side stance (SSS), tandem stance with affected limb in the front (AFS), and tandem stance with nonaffected limb in the front (SFS).

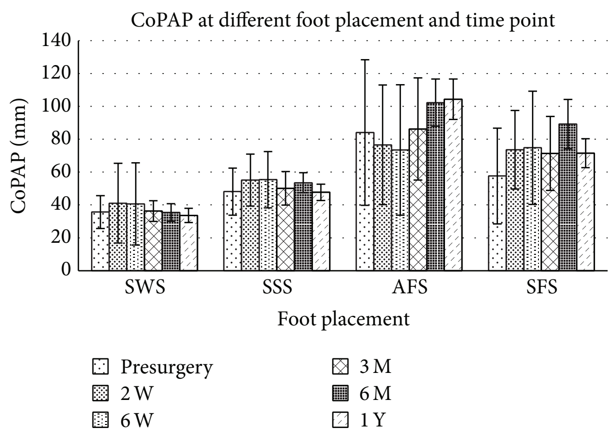


Figure 2: Maximum CoP sway in AP direction (CoPAP) at different foot placement and time points before and after surgery. For abbreviation of foot placement, refer to Figure 1.

Table 2: Summary of two-way repeated measure ANOVA examining the interaction of time point and foot placement and subsequent post hoc analysis of parameters representing postural stability.

Time points	Foot placement SWS	SSS	AFS	SFS	p^{i}	p^{t}	p^{f}
CoPR (unit: mm)							
Presurgery	73.5 ± 18.6^{a1}	64.7 ± 16.6^{a1}	129.3 ± 39.2^{a2}	124.7 ± 33.8^{a2}			
Postsurgery, 2 wk	81.8 ± 21.9^{a1}	74.1 ± 14.8^{b1}	141.8 ± 34.6^{ab2}	152.7 ± 37.1^{b2}			
Postsurgery, 6 wk	79.3 ± 23.7^{a1}	66.1 ± 18.8^{a1}	130.6 ± 37.4^{ad2}	130.1 ± 36.0^{a2}	0.003^{**}	0.000^{**}	0.000^{**}
Postsurgery, 3 mo	79.2 ± 22.1^{a1}	72.8 ± 18.5^{a1}	133.4 ± 37.1^{a2}	133.2 ± 27.8^{a2}			
Postsurgery, 6 mo	72.8 ± 14.5^{ab1}	60.5 ± 9.9^{ac1}	141.0 ± 30.5^{a2}	129.6 ± 20.0^{ac2}			
Postsurgery, 1 yr	57.0 ± 2.7^{b1}	52.1 ± 2.6^{c2}	117.2 ± 11.7^{acd3}	107.1 ± 3.5^{acd4}			
CoPAP (unit: mm)							
Presurgery	35.7 ± 10.0^{a}	48.1 ± 14.3	84.1 ± 44.3^{a}	57.6 ± 29.1^{a}			
Postsurgery, 2 wk	41.1 ± 24.3^{b}	55.1 ± 15.8	76.6 ± 36.4^{a}	73.6 ± 23.9^{b}			
Postsurgery, 6 wk	40.6 ± 25.0^{bc}	55.4 ± 17.1	73.5 ± 39.7^{a}	74.8 ± 34.5^{bc}	0.018^{*}	0.185	0.000^{**}
Postsurgery, 3 mo	36.3 ± 6.3^{ab}	50.1 ± 10.2	86.2 ± 31.1^{a}	71.3 ± 22.5^{bd}			
Postsurgery, 6 mo	35.5 ± 5.3^{ab}	53.5 ± 6.1	102.3 ± 14.3^{ab}	89.2 ± 15.0^{be}			
Postsurgery, 1 yr	33.6 ± 4.3^{ab}	47.7 ± 5.0	104.3 ± 12.3^{b}	71.5 ± 8.8^{ab}			
CoPML (unit: mm)							
Presurgery	22.9 ± 8.3^{a1}	44.8 ± 10.4^{2}	65.3 ± 17.5^{a3}	61.8 ± 19.0^{a34}			
Postsurgery, 2 wk	26.8 ± 10.3^{ac1}	41.9 ± 16.2^{2}	84.7 ± 31.5^{ac3}	67.3 ± 36.6^{ab34}			
Postsurgery, 6 wk	22.7 ± 8.4^{a1}	58.2 ± 25.0^{2}	55.9 ± 24.6^{a3}	65.5 ± 35.3^{ab34}	0.003^{**}	0.000^{**}	0.005^{**}
Postsurgery, 3 mo	22.6 ± 10.5^{ab1}	52.0 ± 23.2^{2}	59.8 ± 12.0^{ad3}	58.1 ± 12.5^{a34}			
Postsurgery, 6 mo	21.9 ± 4.1^{ab1}	50.6 ± 8.4^{2}	88.7 ± 22.1^{b3}	89.8 ± 14.8^{bc34}			
Postsurgery, 1 yr	25.8 ± 3.0^{a1}	50.6 ± 3.1^{2}	91.1 ± 14.4^{b3}	78.9 ± 3.2^{ac34}			
BWDSI (unit: % of body weight)							
Presurgery	2.0 ± 6.7	6.4 ± 9.1	10.4 ± 14.7	4.7 ± 21.5			
Postsurgery, 2 wk	10.5 ± 9.7^{1}	8.5 ± 9.3^{13}	5.2 ± 19.4^{2}	5.3 ± 12.4^{23}			
Postsurgery, 6 wk	2.2 ± 10.8	2.4 ± 10.0	8.5 ± 15.6	6.5 ± 14.5	0.000^{**}	0.000^{**}	0.129
Postsurgery, 3 mo	6.3 ± 8.5^{1}	5.7 ± 4.9^{1}	1.8 ± 11.9^{23}	3.0 ± 9.9^{13}			
Postsurgery, 6 mo	7.3 ± 3.8^{1}	-0.9 ± 3.7^{2}	8.5 ± 7.3^{1}	12.8 ± 10.7^{1}			
Postsurgery, 1 yr	-9.3 ± 1.4^{1}	-9.5 ± 0.8^{1}	10.9 ± 2.2^{2}	-3.5 ± 3.6^{3}			

Shoulder width stance (SWS), feet side-by-side stance (SSS), tandem stance with affected limb in the front (AFS), and tandem stance with nonaffected limb in the front (SFS).
p^{i}: interaction between factors; p^{t}: main effect of time; p^{f}: main effect of foot placement; $^{**}p < 0.01$; $^{*}p < 0.05$.
Superscript of alphabet: Tukey's HSD post hoc analysis of simple main effect of foot placement factor.
Superscript of number: Tukey's HSD post hoc analysis of simple main effect of time factor.

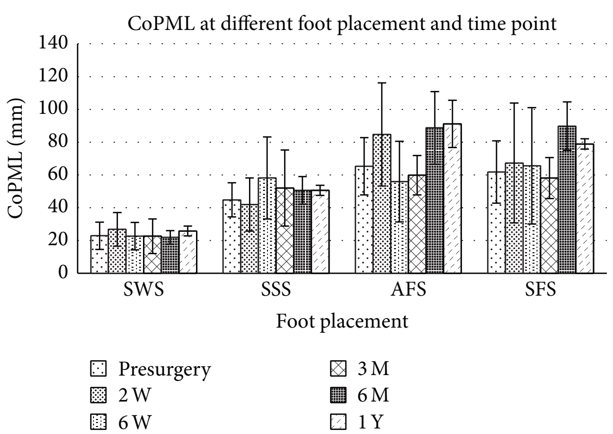


Figure 3: Maximum CoP sway in ML direction (CoPML) at different foot placement and time points before and after surgery. For abbreviation of foot placement, refer to Figure 1.

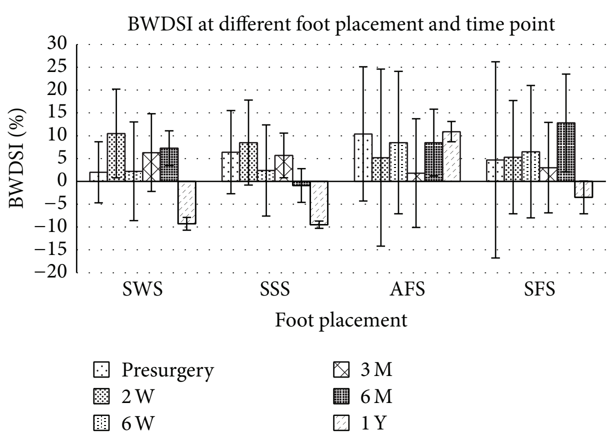


Figure 4: Body weight distribution symmetry index (BWDSI) at different foot placement and time points before and after surgery. For abbreviation of foot placement, refer to Figure 1.

The average bilateral weight bearing symmetrical index was not affected by foot placement (Table 2, $p = 0.129$) but by time point (Table 2, $p < 0.01$). Before and after surgery, the THA patients tended to load their nonaffected limb slightly more than the affected limb when the subjects stood in SWS, SSS, and SFS posture until 1 year after the surgery. When they stood in AFS posture, the subjects tended to increase the weight loaded over the affected limb (Figure 4). At 6 months after surgery, the subjects started to load their weight over the affected limb more than the nonaffected limb only when they adopted SSS posture (Table 2). At 1 year after surgery, the subjects loaded their weight over the affected limb more than the nonaffected limb with almost all four stance postures except when they stood with the affected foot in the back tandem stance (AFS).

3.3. Effects of Time Points on Stance Postural Control Features. The effects of time points on stance postural stability depended on which stance posture the subject adopted. The CoPR, CoPML, and BWDSI showed different across time point in all four foot placements (Table 2), but CoPAP had no main effect on the factor of time point ($p = 0.185$). Subsequent analysis showed that the effects of time point on CoPR were not consistent across foot placement. The CoPR had the greatest postural sway at the time of 2 weeks after surgery, but the CoPML at the time of 6 months after surgery. The descriptive data showed that the CoPR tended to decrease progressively after surgery (Figure 1). CoPAP peaked between 6 months and 1 year after surgery (Figure 2), especially at the AFS and SFS position.

4. Discussion

This is the first and the only study investigating the pain level, functional capacity, and postural stability in patient who underwent MTHA with 1 year follow-up. Quiet stance postural stability during four foot placement conditions was measured to represent the ability of hip joint to manage graded demands for control of postural stability. Though quiet stance has been called a “static” postural control task because the base of support (BOS) is not changing, several studies [31, 32] have suggested that quiet stance is dynamic sensorimotor tasks requiring a very precise control of fine movements of the lower extremities by unceasingly displacing the location of CoP and modulating interlimb loading ratio. Studies also showed that both musculoskeletal and neuromuscular impairments can change the amount of CoP movement and the ratio of interlimb weight bearing [13]. Therefore, postural control actions while quiet stance was measured in this study at six time points before and after MTHA to reveal their sensorimotor recovery process around hip joint. In general, our results showed that MTHA in this study did relieve hip pain and improve function capacity effectively (Table 1). The sensorimotor function around the hip during control of postural stability tended to recover irregularly, and the hypotheses in this study were therefore partially supported.

Our results first showed that the functional capacity of the hip was affected by MTHA but the trend of progressive improvement was not observed through 1 year after surgery. As previously reported [33], the most prominent effects of MTHA were on decreasing the hip pain and on increasing independence of daily functions. However, the MTHA tended to affect postural stability positively within 6 months but negatively within 6 months after surgery as shown by measures of Berg Balance Scale [34, 35] and functional reach test [17], indicating that the somatosensory and neuromuscular function around hip joint did not recover progressively or completely within 1 year. The results of muscle strength further showed that a single joint surgery, even with minimally invasive approach, affects not only the hip muscle strength but also muscle strength around the remote joint in the lower extremity, indicating that the muscular synergy for control of stance postural stability might be also affected [1, 36]. Moreover, our results showed that the stance postural stability during different foot placement conditions was influenced by time points at which postural stability was measured, indicating that the performance of neuromuscular apparatus around hip joint in the process of stance postural control depends on both the configuration of BOS and level of recovery after surgery [13, 31, 32]. The following discussion was based on simple main effects of foot placement and time points on stance postural control features to elucidate the impact of MTHA on hip joint sensorimotor functions.

4.1. Simple Main Effects of Time Points on Stance Postural Control Features. Our results showed significant alteration of postural actions between pre- and postoperative stages and between 6 months and 1 year after surgery. The measured postural stability seemed to alter in accordance with subjective feeling of pain or discomfort [2, 3]. As shown in Tables 1 and 2 the joint pain was relieved significantly and immediately after surgery. However, the patients started to report feelings of discomfort and joint tightness at six months or one year after surgery. This result suggested that level of pain associated with hip joint structural integrity but not with hip joint functional integrity. The progression of stance stability was consistent with Berg balance score and functional reach tests. The results of FIM and hip scores [37–39] suggested that the patients with MTHA tended to be more confident than they are before surgery in daily functioning. The good side of these results is that the autonomy of the patients increased which might accompany increase of quality of life. On the other hand, the patient might put themselves in danger of falling because deficits in stance stability (as shown in Table 2) might emerge and/or last at least until one year after surgery. Clinicians and patients should both be advised and the patients should be instructed to follow postsurgery precautions even though the pain decreased significantly within three months after surgery [2, 3].

Interestingly, the variability of postural stability, as shown by standard deviation measured, was quite large at each time point postoperatively, although the average postural actions remained constant across time point between 2 weeks and 6 months after surgery. The explanation of the large variability might be due to inconsistent postsurgery rehabilitation regimen. Several studies have shown that after 4 weeks of postoperative rehabilitation programs, THA patients were able to achieve an almost complete restoration of the lower limb function and hip joint range of motion and independence in the activities of daily living [8, 40]. In general, the rehabilitation program was implemented by health professionals beginning on the first postoperative day and continued in rehabilitation unit on inpatient basis at 5 days postoperatively [32, 41]. The main goal of rehabilitation was to improve range of motion, muscle strength, aerobic capacity and activities of daily living, and functional abilities such as standing and walking. However, the patients in our study did not have access to standard rehabilitation resources. They are usually educated pre- and postoperatively regarding the wound care and movement precautions after surgery during their short hospital stay (usually 3 to 5 days). The inpatient and postsurgery rehabilitation includes teaching of use of a walker and/or crutches for indoor ambulation and isometric exercises for the operated limb [36]. Finally, the surgeon usually advises the patients to move around as much as possible after they were discharged to home at 3 to 5 days after surgery. Our data with fluctuated postural stability after surgery indicated that the patients might not train themselves enough and self-training might vary across each individual. Clinical interview of our patients showed that some of them tended to decrease their activity level after THA because they were afraid of dislocation of the newly operated joint or OA over the nonoperated joint; others would participate in aerobic activity such as swimming to facilitate functional ability. No therapist was involved in the rehabilitation treatment; either specific ROM or muscle strength training instruction was provided. This might cause the feeling of joint tightness or discomfort that developed at six months after surgery and, thus, change the postural stability while standing. Whether activity levels and self-training intensity facilitate neuromuscular outcomes in patients who underwent MTHA should be further investigated [42–44].

Another noteworthy factor contributing to the fluctuated stance postural stability might be different incision approaches used by the same surgeon across patients. Although the THAs in this study fulfilled the requirement of minimally invasive technique, the level of soft tissue damaged and the muscle groups injured varied across approaches [45–48]. Our results suggested that the neuromuscular outcomes among four MTHAs might be different. Previous studies reported on asymmetric limb kinetic performances during walking in patients who underwent two approaches of MTHA [9, 49]. Further studies should compare the outcomes among three different approaches used.

4.2. Simple Main Effects of Foot Placements on Stance Postural Control Features. Our results have highlighted the tendency of increased postural sway as the foot placement changes from bilateral feet shoulder width stance, bilateral feet side-by-side stance to tandem stance, as reported by previous studies investigating upright standing control features in stroke or normal subjects [17, 50]. These results indicated that the upward neural pathway responsible for perceiving changes in dimensions of BOS in patients with MTHA might be active and efficient for modulation of postural actions, as reported by other studies. On the other hand, the significant increment of postural sway during tandem stance comparing with other stance postures might indicate that the MTHA induced limited neuromuscular capacity around hip joint [9] to constrain postural sway when the demands of postural control increased due to change of BOS dimension. Therefore, it is reasonable to hypothesize that the risks of fall in patients who underwent MTHA might be higher when they stood with tandem stance posture than with the other two stance postures. We, therefore, suggest that pre- and postoperative education programs, in addition to prohibiting certain hip joint movements, should advise the patients to avoid adopting tandem stance posture for at least three months to prevent falls [51].

Another striking result is that patients with MTHA tended to load the newly operated hip more than the non-operated limb during tandem stance without cautions (Figure 3). Inappropriate weight loading over the prosthetic limb was reported to not only increase the incidence of dislocations but also decrease the durability of the prostheses. Therefore, clinicians should advise the patients to avoid standing with heel-toe posture to decrease the incidence of dislocation over the newly operated hip caused by inappropriate weight acceptance [41, 51]. On the other hand, the asymmetric limb loading pattern shed light on the significant contribution of impairment of interlimb LULM on increment of CoP sway magnitude increment. Significant decrement of pain over the OA hip makes the patients subconsciously load the operated limb intermittently but the insufficient hip muscular strength, as proposed by several previous studies [17, 52–54], might cause the frequent interlimb loading and unloading. The weighting of painful sensation and muscular capacity on changes of limb loading patterns and CoP sway magnitude should be further investigated.

Another noteworthy result in this study is that the influence of foot placements on CoP sway in AP and ML direction is different and this phenomenon is the biomechanical consequence of alteration of the dimension of BOS in AP and ML aspects [13]. Shoulder width stance (SWS) and bilateral feet side-by-side stance (FSS) configure the same dimension of BOS in AP aspect but the dimension of BOS in ML aspect is wider in SWS than in SFS. Both tandem stances (AFS and SFS) configure the same dimension of BOS in ML aspect, but the dimension of BOS in AP aspect in SFS is much longer than in SWS and in AFS. Our results found that the CoPAP tended to increase not only as the BOS lengthened in AP dimensions but also as the BOS remained constant in AP aspect. As reported by Rougier [31], the CoP sway in AP aspect is mainly modulated by ankle joint motion and was weakly linked with bilateral limb loading pattern which is primarily modulated by hip joint motion. Therefore, our finding, as those reported [25], suggested that though MTHA affects only the hip joint, it could cause impairments of interjoint coordination (between hip, knee, and ankle joints) necessary for bipedal stance postural control. In ML aspect, our data indicated that CoPML linked with asymmetric limb loading pattern and the COPML was affected by alteration of BOS dimension in ML aspect more than COPAP was. This result is consistent with Rougier's finding [31] and is not surprising since CoP sway in ML aspect is mainly achieved by the control of hip abductor and adductors [42], indicating that MTHA impaired significantly the sensorimotor function and thus limits patient's capacity to effectively constrain postural sway in ML aspects according to the dimension of BOS.

The associated asymmetrical limb loading further indicated that the ankle mechanism of the patients with MTHA is not able to compensate the impaired hip mechanism and, therefore, an adaptive strategy to constrain CoP sway after surgery was not seen. Further studies analyzing relationship between reaction forces, COP sway under each foot, and muscular activation pattern are necessary to understand the adaptive strategies at different foot placement in patients with MTHA.

The major limitation of this study was that the researchers responsible for data collection were not blinded to the purpose of this study. This might impede the objectivity of the data collected when testing the functional capacity. The reason that had caused this limitation is the limited personnel resource allocated. The second limitation was that MTHA with four different approaches was used. Different approaches used might cause the prominent variability of postural stability observed in this study. Small sample size is another limitation which is not uncommon among clinical studies.

5. Conclusions

MTHA is a frequently used procedure to alleviate functional deficits caused by hip joint osteoarthritis. This study found that MTHA is effective in relieving pain in first 2 weeks and restoring the hip joint integrity. However, deterioration of joint pain and functional capacity starting at 6 months postoperatively was noted and these results were associated with sensorimotor recovery fluctuation. Lack of standard and regular postoperative rehabilitation might contribute to this fluctuation. The results of this study also indicated that stance postural stability in patients who underwent MTHA was influenced by foot placement and inappropriate postural stability in both AP and ML directions was noted, especially when the patients stood in tandem posture. Advisement of establishing standardized and regular postsurgical rehabilitation program, including advising the patients to avoid standing postures which expose the operative joint and/or the ipsilateral limb to danger, is recommended for preventions of falls, prolonging the usable duration of the prosthesis and reducing the needs of medical care. Effects of postsurgical rehabilitation program need to be clarified with more delicate research design (such as blindness of the researchers responsible for data collection and inclusion of larger sample size). Comparison among four different MTHA approaches for their effects on functional capacity and sensorimotor recovery is inevitable for clinical decision making.

Conflict of Interests

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

Acknowledgments

This research was funded by CMRPD1B0161, XMPRG360811-3 from Chang Gung Memorial Hospital, Linkou, and BMRP898 from Chang Gung University.

References

[1] M. P. M. Steultjens, J. Dekker, M. E. Van Baar, R. A. B. Oostendorp, and J. W. J. Bijlsma, "Muscle strength, pain and disability in patients with osteoarthritis," *Clinical Rehabilitation*, vol. 15, no. 3, pp. 331–341, 2001.

[2] Y. S. Chiu, W. M. Chen, C. K. Huang et al., "Long-term outcome of primary total hip arthroplasty after a minimum twenty-year follow-up," *Journal of Orthopedic Surgery Taiwan*, vol. 21, no. 2, pp. 55–64, 2004.

[3] O. Ethgen, O. Bruyerè, F. Richy, C. Dardennes, and J.-Y. Reginster, "Health-related quality of life in total hip and total knee arthroplasty: a qualitative and systematic review of the literature," *The Journal of Bone & Joint Surgery—American Volume*, vol. 86, no. 5, pp. 963–974, 2004.

[4] M. Nallegowda, U. Singh, S. Bhan, S. Wadhwa, G. Handa, and S. N. Dwivedi, "Balance and gait in total hip replacement: a pilot study," *American Journal of Physical Medicine & Rehabilitation*, vol. 82, no. 9, pp. 669–677, 2003.

[5] V. Lugade, V. Klausmeier, B. Jewett, D. Collis, and L.-S. Chou, "Short-term recovery of balance control after total hip arthroplasty," *Clinical Orthopaedics and Related Research*, vol. 466, no. 12, pp. 3051–3058, 2008.

[6] M. Majewski, H. A. Bischoff-Ferrari, C. Grüneberg, W. Dick, and J. H. J. Allum, "Improvements in balance after total hip replacement," *The Journal of Bone & Joint Surgery—British Volume*, vol. 87, no. 10, pp. 1337–1343, 2005.

[7] V. C. Mow and R. Huiskes, *Basic Orthopaedic Biomechanics and Mechano-Biology*, Lippincott Williams & Wilkins, Philadelphia, Pa, USA, 2005.

[8] H.-Y. Shi, H.-C. Chiu, J.-K. Chang, J.-W. Wang, R. Culbertson, and M. M. Khan, "Evaluation and prediction of health-related quality of life for total hip replacement among Chinese in Taiwan," *International Orthopaedics*, vol. 32, no. 1, pp. 27–32, 2008.

[9] C. Shih, Y. Du, Y. Lin, and C. Wu, "Muscular recovery around the hip joint after total hip arthroplasty," *Clinical Orthopaedics and Related Research*, vol. 302, pp. 115–120, 1994.

[10] L. E. Powell and A. M. Myers, "The activities-specific balance confidence (ABC) scale," *The Journals of Gerontology: Series A Biological Sciences and Medical Sciences*, vol. 50, no. 1, pp. M28–M34, 1995.

[11] A. M. Myers, P. C. Fletcher, A. H. Myers, and W. Sherk, "Discriminative and evaluative properties of the activities-specific balance confidence (ABC) scale," *Journals of Gerontology—Series A: Biological Sciences and Medical Sciences*, vol. 53, no. 4, pp. M287–M294, 1998.

[12] P. Levinger, D. T. H. Lai, H. B. Menz et al., "Swing limb mechanics and minimum toe clearance in people with knee osteoarthritis," *Gait & Posture*, vol. 35, no. 2, pp. 277–281, 2012.

[13] A. Shumway-Cook and M. H. Woollacott, *Motor Control: Translating Research into Clinical Practice*, Lippincott Williams & Wilkins, Philadelphia, Pa, USA, 3rd edition, 2006.

[14] F. B. Horak and F. Hlavacka, "Somatosensory loss increases vestibulospinal sensitivity," *Journal of Neurophysiology*, vol. 86, no. 2, pp. 575–585, 2001.

[15] S. Mitchell, A. McCaskie, R. Francis, R. Peaston, F. Birrell, and E. Lingard, "The need for a falls prevention programme for patients undergoing hip and knee replacement surgery," *Journal of Orthopaedic Nursing*, vol. 11, no. 2, pp. 98–103, 2007.

[16] D. L. Sturnieks, A. Tiedemann, K. Chapman, B. Munro, S. M. Murray, and S. R. Lord, "Physiological risk factors for falls in older people with lower limb arthritis," *The Journal of Rheumatology*, vol. 31, no. 11, pp. 2272–2279, 2004.

[17] J.-S. Chern, C.-Y. Lo, C.-Y. Wu, C.-L. Chen, S. Yang, and F.-T. Tang, "Dynamic postural control during trunk bending and reaching in healthy adults and stroke patients," *American Journal of Physical Medicine & Rehabilitation*, vol. 89, no. 3, pp. 186–197, 2010.

[18] K. C. Bertin and H. Röttinger, "Anterolateral mini-incision hip replacement surgery: a modified Watson-Jones approach," *Clinical Orthopaedics and Related Research*, no. 429, pp. 248–255, 2004.

[19] J.-M. Laffosse, P. Chiron, F. Accadbled, F. Molinier, J.-L. Tricoire, and J. Puget, "Learning curve for a modified Watson-Jones minimally invasive approach in primary total hip replacement: analysis of complications and early results versus the standard-incision posterior approach," *Acta Orthopaedica Belgica*, vol. 72, no. 6, pp. 693–701, 2006.

[20] R. A. Berger, "Mini-incision total hip replacement using an anterolateral approach: technique and results," *Orthopedic Clinics of North America*, vol. 35, no. 2, pp. 143–151, 2004.

[21] K. S. Park, T. R. Yoon, S. Y. Hwang, and K. B. Lee, "Modified minimally invasive two-incision total hip arthroplasty using large diameter femoral head," *Indian Journal of Orthopaedics*, vol. 46, no. 1, pp. 29–35, 2012.

[22] J. M. Matta and T. A. Ferguson, "The anterior approach for hip replacement," *Orthopedics*, vol. 28, no. 9, pp. 927–928, 2005.

[23] L. D. Dorr, A. V. Maheshwari, W. T. Long, Z. Wan, and L. E. Sirianni, "Early pain relief and function after posterior minimally invasive and conventional total hip arthroplasty. A prospective, randomized, blinded study," *The Journal of Bone & Joint Surgery—American Volume*, vol. 89, no. 6, pp. 1153–1160, 2007.

[24] L. Ogonda, R. Wilson, P. Archbold et al., "A minimal-incision technique in total hip arthroplasty does not improve early postoperative outcomes: a prospective, randomized, controlled trial," *The Journal of Bone & Joint Surgery Series A*, vol. 87, no. 4, pp. 701–710, 2005.

[25] T. P. Sculco, "Minimally invasive total hip arthroplasty: in the affirmative," *Journal of Arthroplasty*, vol. 19, no. 4, pp. 78–80, 2004.

[26] B. S. Bal, D. Haltom, T. Aleto, and M. Barrett, "Early complications of primary total hip replacement performed with a two-incision minimally invasive technique. Surgical technique," *The Journal of Bone & Joint Surgery*, vol. 88, supplement 2, no. 1, pp. 221–233, 2006.

[27] P. W. Duncan, D. K. Weiner, J. Chandler, and S. Studenski, "Functional reach: a new clinical measure of balance," *Journals of Gerontology*, vol. 45, no. 6, pp. M192–M197, 1990.

[28] W. H. Harris, "Traumatic arthritis of the hip after dislocation and acetabular fractures: treatment by mold arthroplasty. An end-result study using a new method of result evaluation," *The Journal of Bone & Joint Surgery—American volume*, vol. 51, no. 4, pp. 737–755, 1969.

[29] M. G. Stineman, A. Jette, R. Fiedler, and C. Granger, "Impairment-specific dimensions within the functional independence measure," *Archives of Physical Medicine and Rehabilitation*, vol. 78, no. 6, pp. 636–643, 1997.

[30] V. Wylde, E. Lenguerrand, L. Brunton et al., "Does measuring the range of motion of the hip and knee add to the assessment of disability in people undergoing joint replacement?" *Orthopaedics and Traumatology: Surgery and Research*, vol. 100, no. 2, pp. 183–186, 2014.

[31] P.-R. Rougier, "What insights can be gained when analysing the resultant centre of pressure trajectory?" *Neurophysiologie Clinique*, vol. 38, no. 6, pp. 363–373, 2008.

[32] E. Trudelle-Jackson and S. S. Smith, "Effects of a late-phase exercise program after total hip arthroplasty: a randomized controlled trial," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 7, pp. 1056–1062, 2004.

[33] I. Boutron, F. Rannou, M. Jardinaud-lopez, G. Meric, M. Revel, and S. Poiraudeau, "Disability and quality of life of patients with knee or hip osteoarthritis in the primary care setting and factors associated with general practitioners' indication for prosthetic replacement within 1 year," *Osteoarthritis and Cartilage*, vol. 16, no. 9, pp. 1024–1031, 2008.

[34] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: preliminary development of an instrument," *Physiotherapy Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[35] K. Berg, S. Wood-Dauphinee, and J. I. Williams, "The balance scale: reliability assessment with elderly residents and patients with an acute stroke," *Scandinavian Journal of Rehabilitation Medicine*, vol. 27, no. 1, pp. 27–36, 1995.

[36] H. Clarkson, *Musculoskeletal Assessment: Joint Range of Motion and Manual Muscle Strength*, Lippincott Williams & Wilkins, 2000.

[37] A. M. J. Chorus, H. S. Miedema, A. Boonen, and S. van der Linden, "Quality of life and work in patients with rheumatoid arthritis and ankylosing spondylitis of working age," *Annals of the Rheumatic Diseases*, vol. 62, no. 12, pp. 1178–1184, 2003.

[38] N. A. Johanson, M. E. Charlson, T. P. Szatrowski, and C. S. Ranawat, "A self-administered hip-rating questionnaire for the assessment of outcome after total hip replacement," *The Journal of Bone & Joint Surgery—American Volume*, vol. 74, no. 4, pp. 587–597, 1992.

[39] D. Whalley, S. P. McKenna, Z. De Jong, and D. Van Der Heijde, "Quality of life in rheumatoid arthritis," *British Journal of Rheumatology*, vol. 36, no. 8, pp. 884–888, 1997.

[40] H. Pendleton and W. Schultz-Krohn, *Pedretti's Occupational Therapy: Practice Skills for Physical Dysfunction*, Mosby Elsevier, St. Louis, Mo, USA, 2007.

[41] C. Kisner and L. A. Colby, *Therapeutic Exercise: Foundations and Technique*, F.A. Davis Company, Philadelphia, Pa, USA, 2007.

[42] R. Meneghini, M. Pagnano, R. Trousdale, and W. Hozack, "Muscle damage during MIS total hip arthroplasty: Smith-Petersen versus posterior approach," *Clinical Orthopaedics and Related Research*, vol. 453, pp. 293–298, 2006.

[43] K. Reardon, M. Galea, X. Dennett, P. Choong, and E. Byrne, "Quadriceps muscle wasting persists 5 months after total hip arthroplasty for osteoarthritis of the hip: a pilot study," *Internal Medicine Journal*, vol. 31, no. 1, pp. 7–14, 2001.

[44] M.-L. Lu, S.-W. Chou, W.-E. Yang et al., "Hospital course and early clinical outcomes of two-incision total hip arthroplasty," *Chang Gung Medical Journal*, vol. 30, no. 6, pp. 513–520, 2007.

[45] D. Glaser, D. A. Dennis, R. D. Komistek, and T. M. Miner, "In vivo comparison of hip mechanics for minimally invasive versus traditional total hip arthroplasty," *Clinical Biomechanics*, vol. 23, no. 2, pp. 127–134, 2008.

[46] V. Klausmeier, V. Lugade, B. A. Jewett, D. K. Collis, and L.-S. Chou, "Is there faster recovery with an anterior or antero-lateral THA? A Pilot Study," *Clinical Orthopaedics and Related Research*, vol. 468, no. 2, pp. 533–541, 2010.

[47] R. M. Meneghini, S. A. Smits, R. R. Swinford, and R. E. Bahamonde, "A randomized, prospective study of 3 minimally invasive surgical approaches in total hip arthroplasty. Comprehensive gait analysis," *The Journal of Arthroplasty*, vol. 23, no. 6, pp. 68–73, 2008.

[48] C. S. Mow, S. T. Woolson, S. G. Ngarmukos, E. H. Park, and H. P. Lorenz, "Comparison of scars from total hip replacements done with a standard or a mini-incision," *Clinical Orthopaedics and Related Research*, vol. 441, pp. 80–85, 2005.

[49] J. Perry and J. M. Burnfield, *Gait Analysis: Normal and Pathological Function*, SLACK Incorporated, Thorofare, NJ, USA, 2nd edition, 2010, SLACK incorporated.

[50] J.-S. Chern, S.-W. Yang, and C.-Y. Wu, "Whole-body reaching as a measure of dynamic balance in patients with stroke," *American Journal of Physical Medicine & Rehabilitation*, vol. 85, no. 3, pp. 201–208, 2006.

[51] C. Trombly, M. Radomski, Trombly, and Radomski, *Occupational Therapy for Physical Dysfunction*, Lippincott Williams & Wilkins, Baltimore, Md, USA, 1989.

[52] S. Yang, W.-H. Hwang, Y.-C. Tsai, F.-K. Liu, L.-F. Hsieh, and J.-S. Chern, "Improving balance skills in patients who had stroke through virtual reality treadmill training," *American Journal of Physical Medicine & Rehabilitation*, vol. 90, no. 12, pp. 969–978, 2011.

[53] P. A. Gribble and J. Hertel, "Effect of lower-extremity muscle fatigue on postural control," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 4, pp. 589–592, 2004.

[54] M. Salavati, M. Moghadam, I. Ebrahimi, and A. M. Arab, "Changes in postural stability with fatigue of lower extremity frontal and sagittal plane movers," *Gait & Posture*, vol. 26, no. 2, pp. 214–218, 2007.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 385269, 7 pages
http://dx.doi.org/10.1155/2015/385269

Research Article

Mobility and Balance and Their Correlation with Physiological Factors in Elderly with Different Foot Postures

Aisyah Mohd Said, Haidzir Manaf, Saiful Adli Bukry, and Maria Justine

Department of Physiotherapy, Faculty of Health Sciences, Universiti Teknologi MARA, Puncak Alam Campus, 42300 Puncak Alam, Selangor, Malaysia

Correspondence should be addressed to Maria Justine; maria205@salam.uitm.edu.my

Received 6 March 2015; Revised 15 June 2015; Accepted 1 July 2015

Academic Editor: Frédéric Noé

This study determines (1) the correlation between mobility and balance performances with physiological factors and (2) the relationship between foot postures with anthropometric characteristics and lower limb characteristics among elderly with neutral, pronated, and supinated foot. A cross-sectional observational study was conducted in community-dwelling elderly (age: 69.86 ± 5.62 years). Participants were grouped into neutral ($n = 16$), pronated ($n = 14$), and supinated ($n = 14$) foot based on the foot posture index classification. Anthropometric data (height, weight, and BMI), lower limb strength (5-STS) and endurance (30 s chair rise test), mobility (TUG), and balance (FSST) were determined. Data were analyzed using Spearman's correlation coefficient. Body weight was negatively and moderately correlated ($r_s = -0.552$, $P < 0.05$) with mobility in supinated foot; moderate-to-high positive linear rank correlation was found between lower limb strength and mobility ($r_s = 0.551$ to 0.804, $P < 0.05$) for pronated and neutral foot. Lower limb endurance was negatively and linearly correlated with mobility in pronated ($r_s = -0.699$) and neutral ($r_s = -0.573$) foot. No correlation was observed in balance performance with physiological factors in any of the foot postures. We can conclude that muscle function may be the most important feature to make movement possible in older persons regardless of the type of foot postures.

1. Introduction

The foot is an important body part because it supports body weight and organizes locomotion. However, this body part is vulnerable to daily strains when an individual walks [1]. Musculoskeletal disorders, such as foot malalignment, may be associated with functional restraint, even though a particular disorder is not painful [2]. Foot characteristics are related to mobility and functionality in elderly [3]. The present study focuses on the dynamic foot function and gait performances concerning individuals of advanced age rather than the type of foot. In theory, the structure of the lower limb and foot may be vulnerable to several factors, such as footwear [4–6], excessive body weight [7], job nature [8], and physical activity level [9]; these factors may affect foot structures as people age. These factors may also influence some lower limb functions, such as balance and gait performances.

Older persons with foot problems have reported multiple falls compared with those who do not have foot problems; this phenomenon may indicate a higher risk of fall in the future [10]. For instance, individuals with pronated feet are at a high risk of falls or loss of balance during unilateral stance in functional activities; individuals with supinated feet may present disturbed postural control [11]. The lower limb function is also affected by foot posture; for example, in contrast to individuals with normal-arched feet, individuals with flat-arched feet use their tibialis anterior muscle during the contact phase but use the tibialis posterior muscle during midstance or propulsion [12]. Differences in muscle activity may be a sign of a neuromuscular compensation to decrease the overwork of the medial longitudinal arch. These differences may lead to other problems, such as muscular fatigue, which eventually affects balance performance during dynamic activities [13–15].

This study aimed to address the problems regarding foot posture and balance performances in elderly. Some variations in foot posture are associated with changes in lower limb motion and muscle activity, which are strongly influenced by several systemic conditions, such as neurological or rheumatological diseases [16]. Foot posture also affects the mechanical alignment and dynamic function of the lower limb; therefore, foot posture may be related to the development of lower limb disorders [17]. Furthermore, our research scope is similar to those described in previous studies [11, 18–20]. Significant changes have been observed in functional performances among different foot postures; however, the tested population is limited to young adults. With ongoing interests, whether different types of foot postures affect mobility and balance of the elderly remains inconclusive; elderly possibly exhibit high severity partly because of the aging process.

This study aimed to (1) determine whether physiological factors, such as anthropometric data (height, body weight, and body mass index (BMI)) and lower limb characteristics (strength and endurance), are associated with balance and mobility in elderly with different types of foot postures, that is, neutral, pronated, and supinated feet; this study was also conducted to (2) determine the relationship of foot postures with anthropometric characteristics and lower limb characteristics. The result of this study may provide the basis of extensive studies on the assessments of foot posture in clinical settings. This study may also be applied to identify lower limb conditions in the early stages and predict the risk of falls. We hypothesized that the anthropometric data and lower limb characteristics were significantly correlated with balance and mobility regardless of the types of foot postures. We also hypothesized that foot posture may significantly differ in anthropometric and lower limb characteristics.

2. Methods

2.1. Participants and Study Design. This study applied a cross-sectional design. Power analysis [11, 18–20] was performed to compare foot postures and functional performances of individuals with different foot types; in this method, an estimated sample of 30 to 48 participants could provide significant results. Thus, 44 community-dwelling elderly females (age range = 60 years to 85 years) were recruited via a convenience sampling to yield a significant result. The participants were included if the following criteria were satisfied: (1) there are no chronic orthopedic conditions, such as rheumatoid arthritis, severe knee osteoarthritis, and acute fracture, injury, or pain in the lower limb area; (2) there are no vestibular or neurological impairments; (3) there is no peripheral neuropathy or sensory deficits caused by diabetes or any systemic conditions; (4) they can walk continuously for 10 m without walking aids; and (5) they are not involved in any structured exercise classes of more than three times a week (physically inactive). All participants included in the study signed an informed consent approved by the institutional ethics committee. All three subgroups (pronated, supinated, and neutral feet) were formed from the total eligible participants. On the basis of the assessment, we found that 16 patients (age 65 years to 81 years) exhibited neutral feet, 14 patients (age 60 years to 80 years) presented pronated feet, and 14 patients (age 61 years to 85 years) manifested supinated feet. All the participants were actively involved in religious classes 3 to 5 times per day, every day, which were held at the mosque located at least 20 to 500 metres from their home.

2.2. Study Procedures and Outcome Measures. The test procedure was performed indoors, in a controlled environment. The feet of the participants were examined by one assessor using the six-item foot posture index (FPI), a clinical diagnostic tool that can distinctively quantify and classify the particular foot as neutral, pronated, or supinated posture [21]. The FPI reliability coefficient of the application on elderly is 0.61 [22].

2.2.1. Anthropometric Data. Anthropometric factors, including height (m), weight (kg), and BMI (kg/m^2), were evaluated, in accordance with a standard procedure.

2.2.2. Lower Limb Characteristics (Lower Limb Strength and Endurance). The five-time sit-to-stand test (5-STS) was used to measure lower limb strength [23]. The participants were initially instructed to be in a sitting position on a chair with a standard height of 45 cm from the ground [24]; in a sitting position, the participants were also instructed to have both of their arms crossed at the wrists and placed on the chest. The test required the participants to stand and sit repeatedly five times as fast as possible; during this test, the researcher used a stopwatch and recorded the time (s) at which the task was completed. A short time to complete the test corresponds to a strong lower limb.

Lower limb endurance was measured using the 30 s chair rise test [25]. In this test, the participants were instructed to stand upright from a chair and to sit again with their arms folded across their chest. This task was performed repeatedly or as much as they can in their self-preferred speed for 30 s. Numerous repetitions from a sitting position to a standing position indicate excellent lower limb endurance.

2.2.3. Mobility. The Timed-Up and Go test (TUG) was used to measure the mobility of the participants. Several studies [1, 4, 26, 27] have applied this test, particularly in elderly populations and patients with neurological conditions. TUG has also demonstrated good interrater reliability (ICC = 0.99) [28] in elderly when this parameter is used to assess functional mobility. The testing procedure [29] required the participant to stand from a seated position on a standard chair with a seat height of approximately 40 cm to 50 cm, walk at a normal walking speed along a 3 m distance, and turn and walk back toward the chair to sit again. TUG was chosen to reflect mobility based on its characteristic that involves "transition" of multiple activities of sit-to-stand, walking at short distances, and changing direction [30]. A short time to complete the test indicates good functional mobility.

2.2.4. Balance. The Four-Square Step test (FSST) was used to measure balance performance. This test can be used as a reliable and valid tool to assess the dynamic standing balance

Table 1: Characteristics of the participants ($N = 44$).

Characteristics	Neutral ($n = 16$) Mean (SD)	Pronated ($n = 14$) Mean (SD)	Supinated ($n = 14$) Mean (SD)	P value
Age (years)	71.13 (4.674)	67.79 (5.780)	70.50 (6.223)	0.196
Height (m)	1.49 (0.065)	1.50 (0.054)	1.52 (0.056)	0.516
Weight (kg)	54.44 (14.289)	60.24 (14.322)	59.91 (11.99)	0.422
Body mass index (BMI) (kg/m^2)	24.09 (5.238)	26.41 (5.837)	25.94 (4.584)	0.474
Five-time sit-to-stand (sec.)	12.93 (2.608)	11.93 (3.050)	11.67 (2.786)	0.521
30-second chair rise (rep.)	12.63 (2.446)	12.00 (3.762)	13.36 (3.342)	0.230
Timed-Up and Go (sec)	10.73 (2.566)	10.38 (2.166)	9.85 (2.638)	0.484
Four-Square Step test (sec)	14.33 (4.594)	16.75 (6.427)	13.40 (4.232)	0.291

Comparisons were tested using Kruskal-Wallis analysis (nonparametric).
P values were set at a significance level of $P < 0.05$.

of older people, including those with transtibial amputation or those with vestibular dysfunctions [31]. The stepping base was constructed using two canes crossed with each other; four squares numbered from 1 to 4 were formed on the floor. The participants performed this test in the following stepping sequence: clockwise starting at square 1, proceeding to squares 2, 3, and 4, and back to square 1; counterclockwise starting at square 4, proceeding to squares 3 and 2, and ending at square 1. Both feet must be placed in each square as the participants moved from one square to another. The score was recorded as the time spent to complete the sequence. The stopwatch started when the first foot contacted the floor in square 2 and ended when the last foot came back to touch the floor in square 1 [32]. The test was repeated if the participant failed to complete the sequence, lost balance, or came in contact with the cane. A good balance performance is indicated by a short time in seconds to complete the task.

2.3. Statistical Analysis. Descriptive statistics and correlation analysis were performed using SPSS 20.0 (IBM Corporation, Somers, NY). The mean and standard deviation were calculated for each variable. The significance level was set as a priori at $P < 0.05$. Comparisons between foot posture (neutral, pronated, and supinated) with anthropometric characteristics and lower limb characteristics were tested using Kruskal-Wallis analysis (nonparametric). Neutral, pronated, and supinated feet as subgroups were analyzed using Spearman's correlation coefficients to determine the associations of the physiological domains with balance and mobility. All analyses were done using nonparametric test. The correlation results were interpreted as poor ($r_s < 0.30$), low ($r_s = 0.30$ to 0.50), moderate ($r_s = 0.50$ to 0.70), high ($r_s = 0.70$ to 0.90), or very high ($r_s > 0.90$) [33].

3. Results

3.1. Demographic Data, Physiological Factors, and Mobility and Balance in Different Types of Foot Postures. Table 1 presents the demographic data, physiological factors, and mobility and balance among the participants. The results of mean comparisons revealed that none of the variables was significantly different from one another among all of the groups of elderly in terms of foot postures.

3.2. Correlation of Mobility and Balance. Spearman's correlation coefficients (Table 2) indicated that height and BMI were not correlated with balance or mobility in any types of foot. However, weight was moderately and negatively correlated with mobility in the supinated feet ($r_s = -0.552$, $P < 0.05$); by contrast, weight was not correlated with pronated and neutral foot group. Lower limb strength was significantly and moderately correlated with mobility in elderly with pronated feet ($r_s = 0.551$); lower limb strength also exhibited a significantly and highly positive correlation with mobility in elderly with neutral feet ($r_s = 0.804$, $P < 0.01$). Furthermore, lower limb endurance was highly and negatively correlated with mobility in the pronated feet ($r_s = -0.669$, $P < 0.01$), while lower limb endurance was negatively and moderately correlated with mobility in the neutral feet ($r_s = -0.573$, $P < 0.05$). Balance performance was not correlated with any of the anthropometric or physiological factors in all of the three types of foot.

4. Discussion

This study aimed to determine the associations of balance and mobility with physiological factors in elderly with neutral, pronated, and supinated foot postures. To our knowledge, studies have extensively investigated the relationship of physiological factors with balance and mobility; however, these studies have not implemented the commonly used senior fitness test to represent the physiological characteristics of participants [34, 35]. We believed that the implementation of the measurement tools in this study could represent the basic movement that is more functional for elderly rather than the use of advanced technologies. This study is the first to demonstrate the direct relationship of balance variables with physiological and anthropometric factors among elderly with different types of foot postures.

4.1. Anthropometric, Mobility, and Balance. In this study, height was not correlated with mobility in all types of foot. This result is not consistent with that described in a previous

Table 2: Spearman (ρ) correlation coefficients of physiological factors with mobility and balance performances in supinated, pronated, and neutral feet.

Correlates	Mobility (TUG)			Balance (FSST)		
	Neutral	Pronated	Supinated	Neutral	Pronated	Supinated
	r_s (P value)	r_s (P value)	r_s (P value)	r_s (P value)	r_s (P value)	r_s (P value)
Height (m)	−0.395 (0.130)	0.378 (0.182)	−0.402 (0.154)	−0.093 (0.733)	−0.013 (0.964)	−0.365 (0.200)
Weight (kg)	−0.241 (0.368)	−0.024 (0.9.34)	**−0.552*** **(0.041)**	−0.006 (0.983)	0.130 (0.658)	−0.187 (0.523)
BMI	−0.144 (0.594)	−0.122 (0.679)	−0.310 (0.281)	−0.109 (0.688)	0.103 (0.725)	−0.020 (0.946)
LL strength	**0.804**** **(0.041)**	**0.551*** **(0.041)**	0.484 (0.079)	0.368 (0.161)	0.163 (0.578)	0.491 (0.075)
LL endurance	**−0.573*** **(0.020)**	**−0.669**** **(0.009)**	−0.242 (0.405)	−0.243 (0.365)	−0.356 (0.212)	−0.266 (0.357)

Comparisons were tested using Spearman correlation coefficient analysis (nonparametric).
*Correlation is significant at the 0.05 level (2-tailed).
**Correlation is significant at the 0.01 level (2-tailed).
LL: lower limb.

Table 3: Comparisons of mean age, BMI, and number of participants in the present study and previous studies.

Author, year	Mean age (years)	Mean BMI (kg/m^2)	Total participants
Present study	69.08	25.44	50 (44 female; 6 male)
[40]	40.5	35.2	59 (male only)
[39]	22.1	17.4–33.8*	80 (40 female; 40 male)
[34]	22.8	24.6	108 (68 female; 40 male)

*Comparisons were made between groups of underweight, normal weight, overweight, and obese subjects.

study [36], which revealed that height is correlated with balance variables, such as static standing; increased height corresponds to poor balance. This finding is attributed to the center of mass [36, 37] and the increase in the response of ankle and gastrocnemius as height increases [38]. Thus, muscle activation may explain the findings of the current study. In neutral and supinated foot groups, the intrinsic factors that influenced foot arch might also affect the ankle range of movement and agility, especially during turning in the TUG. However, this theory should be verified through a biomechanical analysis or electromyography (EMG) to accurately determine the involvement of intrinsic musculatures.

Our results also demonstrated that weight was associated with the mobility of individuals with supinated feet; by contrast, weight was not associated with the balance and mobility of individuals with other types of foot. However, this finding is inconsistent with that in previous studies, which demonstrated a decrease in balance in individuals with a heavier body weight compared to those with a lighter body weight [34, 39, 40]. Indeed, body weight is a reliable predictor of mobility and balance [40]. The inconsistency between the present study and a previous study [34] can be due to the differences in the mean age and BMI. In a previous study [39], underweight, normal weight, overweight, and obese individuals are compared in terms of stability; the results show that body weight is correlated with stability. Weight gain induced changes in stability regardless of gender; in particular, individuals in the underweight group demonstrate a good balance performance; postural activity is inversely proportional to increased BMI. These inconsistencies could also be attributed to the different physiological factors of the participants in the studies; for instance, the mean age, BMI, and number of participants in each study differed (Table 3). In our study, our unexpected results could be attributed to the small sample size.

A previous study [39] explained that balance deterioration in the obese group is due to an individual's inability to generate sufficient muscle force to control the displacement mass trajectory during balance performance. However, the results of the previous study [39] demonstrated contrasting findings; in particular, most participants are overweight but characterized with considerable lower limb strength. Individuals who spend more than 13.6 s to rise from a chair for five repetitions likely exhibit increased disability and morbidity [41]. In contrast to previous findings, the present results showed that approximately 74% of the participants completed the 5-STS in less than 13.6 s. The result explains that most of the participants have considerably good lower limb strength regardless of their age and BMI. This result can be attributed to the nature of the physical activities of the participants.

In terms of mobility, which was measured using TUG, the participants in all of the three groups scored an average of

9.85 s to 10.73 s, which was less than the cut-off time of 13 s. For balance performance as tested by the FSST, only the participants with pronated feet scored an average of 16.75 s, with the cut-off time of 15 s [32]. Thus, most of the participants presented good lower limb strength regardless of the body weight and types of feet; therefore, these participants should be able to theoretically generate sufficient force to maintain balance. We related these findings to the nature of the daily activities of the participants. The participants are community-dwellers, independent in their basic activities of daily living, and actively involved in religious activities. The participants also live near a religious center, where they attend daily classes at least three times to five times a day; these participants also walk from their houses to the center daily. These independent folks may be used to walking at such distances. Thus, these participants presented a good function of the lower limb regardless of the types of feet.

4.2. Lower Limb Functions, Mobility, and Balance. On the basis of the results of the correlation analysis, we found that a linear rank correlation existed between lower limb strength and balance performances of the pronated and neutral foot groups. Hence, individuals with better lower limb strength may exhibit a higher ability to perform activities in standing or dynamic standing. This result is consistent with that in a previous study [42], which revealed that foot disorders are unlikely associated with functional outcomes. Thus, the result of our study may be best explained by the lifestyles of the participants; this way of life could be the confounding factor of balance and mobility rather than the type of foot, especially in women [1].

The lower limb endurance demonstrated a good linear rank correlation with balance and mobility in the neutral and pronated foot groups. Previous studies [24, 43, 44] showed that functional balance and mobility are directly improved as lower limb endurance increased. However, the current study revealed contradicting findings through comparisons among types of foot. In theory, individuals with supinated feet may be less stable because of the intrinsic muscle tightness and reduced medial longitudinal arch that causes less flexibility and less stability than those with neutral feet [45]. TUG test encompasses the elements of standing from sitting and walking at a distance as well as turning [30], where agility may play a role more than strength to complete the test. Thus, the TUG performance cannot be determined regardless of muscle strength. The pronated and neutral foot groups showed a greater foot contact area and better intrinsic stability than the supinated foot group; thus, the administration of TUG and the relationships with lower limb endurance yielded better results with less confounding factors.

In the population considered for this study, the confounding factor may be caused by the administration and the psychometric properties of the assessment tool. For instance, TUG is known as an excellent outcome measure to identify the risk of fall; however, TUG is unable to identify the existing impairments in static or dynamic balance skills of an individual [41]. The type of foot malalignment has been determined and grouped accordingly to establish the effects on the participants' performances. This process also applies to the results of balance performance. An ability to maintain balance should be associated with various factors, such as coordination, vestibular system, motor response, sensorimotor, and musculoskeletal characteristics of a particular individual. In the administration of the FSST, only the ability of the participants to plan, step, and change directions across obstacles is evaluated [31]; any change in the foot is not considered. Thus, this test might not be sensitive enough to detect any changes in the foot structure during the administration of the test. Furthermore, this type of outcome measure might not be the best to identify the differences in a performance-based perspective. For this reason, the dynamic balance should be evaluated using technical assessments, such as EMG modalities with a sensor plate. Therefore, foot malalignment of the lower limbs does not have a major role in balance and mobility, as measured by TUG and FSST in relation to some physiological characteristics.

4.3. Study Limitations. We noted several limitations of this study. First, the participants in this study were community-dwellers who are actively involved in religious classes three times to five times per day; these participants are used to walking independently around their community. This condition may have led to some effect towards their test result. The small sample size also contributed largely to the lack of strength of this finding. Thus, this study could not be used for a higher level of analysis to evaluate the physiological factors as predictors of balance and mobility due to the nonparametric approach in the analysis. With this limitation, the findings may not be generalized to a larger population. Further studies should be conducted using a larger population with a significant disability, including individuals at risk of falls. The researchers of this study also relied solely on the reliability of the FPI based on a previous study and retained only one researcher to evaluate this measure. We wish to extend our study to explore the muscle activity among different types of foot. We believe that further studies, especially those applying EMG, may enhance the objectivity and accuracy of findings to determine the potential differences among these foot postures.

5. Conclusion

Muscle properties such as strength and endurance of the lower limb may be the main factors that can affect mobility performance in elderly, regardless of their types of foot postures and thus may be an important feature to make movement possible in older persons. Thus, activities with the element of strength training should be encouraged among older persons; with this method, these individuals can preserve their basic functions for a prolonged period.

Conflict of Interests

The authors declare that they have no potential conflict of interests with respect to the authorship and/or publication of this paper.

Acknowledgments

The authors wish to thank the Ministry of Education, Malaysia, for funding the research project through the Research Acculturation Grant Scheme (Ref. no. 600-RMI/RAGS 5/3 (66/2014)) and the Research Management Institute (RMI), Universiti Teknologi MARA (UiTM), for the administrative support.

References

[1] D. Chaiwanichsiri, S. Janchai, and N. Tantisiriwat, “Foot disorders and falls in older persons,” *Gerontology*, vol. 55, no. 3, pp. 296–302, 2009.

[2] F. Badlissi, J. E. Dunn, C. L. Link, J. J. Keysor, J. B. McKinlay, and D. T. Felson, “Foot musculoskeletal disorders, pain, and foot-related functional limitation in older persons,” *Journal of the American Geriatrics Society*, vol. 53, no. 6, pp. 1029–1033, 2005.

[3] M.-C. Chiu, H.-C. Wu, L.-Y. Chang, and M.-H. Wu, “Center of pressure progression characteristics under the plantar region for elderly adults,” *Gait & Posture*, vol. 37, no. 3, pp. 408–412, 2013.

[4] A. Bhatia and S. Kalra, “Footwear effects on balance and gait in elderly women of Indian population between the ages 55 and 75 years,” *Indian Journal of Physiotherapy and Occupational Therapy*, vol. 5, no. 1, 2011.

[5] P. A. O. Pezzan, I. C. N. Sacco, and S. M. A. João, “Foot posture and classification of the plantar arch among adolescent wearers and non-wearers of high-heeled shoes,” *Revista Brasileira de Fisioterapia*, vol. 13, no. 5, pp. 399–404, 2009.

[6] K. D’Août, T. Pataky, D. de Clercq, and P. Aerts, “The effects of habitual footwear use: foot shape and function in native barefoot walkers,” *Footwear Science*, vol. 1, no. 2, pp. 81–94, 2009.

[7] T. R. Aurichio, J. R. Rebelatto, and A. P. de Castro, “The relationship between the body mass index (BMI) and foot posture in elderly people,” *Archives of Gerontology and Geriatrics*, vol. 52, no. 2, pp. e89–e92, 2011.

[8] N. Shibuya, R. T. Kitterman, J. LaFontaine, and D. C. Jupiter, “Demographic, physical, and radiographic factors associated with functional flatfoot deformity,” *The Journal of Foot and Ankle Surgery*, vol. 53, no. 2, pp. 168–172, 2014.

[9] L. Stathokostas, M. W. McDonald, R. M. D. Little, and D. H. Paterson, “Flexibility of older adults aged 55–86 years and the influence of physical activity,” *Journal of Aging Research*, vol. 2013, Article ID 743843, 8 pages, 2013.

[10] H. B. Menz and S. R. Lord, “The contribution of foot problems to mobility impairment and falls in community-dwelling older people,” *Journal of the American Geriatrics Society*, vol. 49, no. 12, pp. 1651–1656, 2001.

[11] L.-C. Tsai, B. Yu, V. S. Mercer, and M. T. Gross, “Comparison of different structural foot types for measures of standing postural control,” *Journal of Orthopaedic and Sports Physical Therapy*, vol. 36, no. 12, pp. 942–953, 2006.

[12] G. S. Murley, H. B. Menz, and K. B. Landorf, “Foot posture influences the electromyographic activity of selected lower limb muscles during gait,” *Journal of Foot and Ankle Research*, vol. 2, article 35, 25 pages, 2009.

[13] I. R. Lanza, D. W. Russ, and J. A. Kent-Braun, “Age-related enhancement of fatigue resistance is evident in men during both isometric and dynamic tasks,” *Journal of Applied Physiology*, vol. 97, no. 3, pp. 967–975, 2004.

[14] Z. Abdolvahabi, S. S. Bonab, H. Rahmati, and S. S. Naini, “The effects of ankle plantar flexor and knee extensor muscles fatigue on dynamic balance of the female elderly,” *World Applied Sciences Journal*, vol. 15, no. 9, pp. 1239–1245, 2011.

[15] F. Islami, Z. Fallah, S. Mahdavi, A. F. Hesari, and H. Janani, “Effect of quadriceps and ankle plantar flexor muscle fatigue on balance of elderly women,” *HealthMED*, vol. 6, no. 3, pp. 875–878, 2012.

[16] G. S. Murley, H. B. Menz, and K. B. Landorf, “A protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements,” *Journal of Foot and Ankle Research*, vol. 2, no. 1, article no. 22, 2009.

[17] P. Levinger, H. B. Menz, M. R. Fotoohabadi, J. A. Feller, J. R. Bartlett, and N. R. Bergman, “Foot posture in people with medial compartment knee osteoarthritis,” *Journal of Foot and Ankle Research*, vol. 3, article 29, 8 pages, 2010.

[18] S. C. Cobb, L. L. Tis, B. F. Johnson, and E. J. Higbie, “The effect of forefoot varus on postural stability,” *Journal of Orthopaedic and Sports Physical Therapy*, vol. 34, no. 2, pp. 79–85, 2004.

[19] K. P. Cote, M. E. Brunet, B. M. Gansneder, and S. J. Shultz, “Effects of pronated and supinated foot postures on static and dynamic postural stability,” *Journal of Athletic Training*, vol. 40, no. 1, pp. 41–46, 2005.

[20] J. Hertel, M. R. Gay, and C. R. Denegar, “Differences in postural control during single-leg stance among healthy individuals with different foot types,” *Journal of Athletic Training*, vol. 37, no. 2, pp. 129–132, 2002.

[21] A. C. Redmond, Y. Z. Crane, and H. B. Menz, “Normative values for the Foot Posture Index,” *Journal of Foot and Ankle Research*, vol. 1, no. 1, article 6, 9 pages, 2008.

[22] H. B. Menz, M. E. Morris, and S. R. Lord, “Foot and ankle risk factors for falls in older people: a prospective study,” *Journals of Gerontology, Series A: Biological Sciences and Medical Sciences*, vol. 61, no. 8, pp. 866–870, 2006.

[23] S. S. M. Ng, S. Y. Cheung, L. S. W. Lai, A. S. L. Liu, S. H. I. Ieong, and S. S. M. Fong, “Association of seat height and arm position on the five times sit-to-stand test times of stroke survivors,” *BioMed Research International*, vol. 2013, Article ID 642362, 6 pages, 2013.

[24] M. L. Gault, R. E. Clements, and M. E. T. Willems, “Functional mobility of older adults after concentric and eccentric endurance exercise,” *European Journal of Applied Physiology*, vol. 112, no. 11, pp. 3699–3707, 2012.

[25] M. G. Jørgensen, “Assessment of postural balance in community-dwelling older adults—methodological aspects and effects of biofeedback-based Nintendo Wii training,” *Danish Medical Journal*, vol. 61, no. 1, Article ID B4775, 2014.

[26] M.-R. Lin, H.-F. Hwang, M.-H. Hu, H.-D. I. Wu, Y.-W. Wang, and F.-C. Huang, “Psychometric comparisons of the Timed Up and Go, one-leg stand, functional reach, and tinetti balance measures in community-dwelling older people,” *Journal of the American Geriatrics Society*, vol. 52, no. 8, pp. 1343–1348, 2004.

[27] M.-L. Bird, K. D. Hill, and J. W. Fell, “A randomized controlled study investigating static and dynamic balance in older adults after training with pilates,” *Archives of Physical Medicine and Rehabilitation*, vol. 93, no. 1, pp. 43–49, 2012.

[28] G. Thrane, R. M. Joakimsen, and E. Thornquist, “The association between timed up and go test and history of falls: the Tromsø study,” *BMC Geriatrics*, vol. 7, article 1, 2007.

[29] M.-R. Lin, H.-F. Hwang, M.-H. Hu, H.-D. I. Wu, Y.-W. Wang, and F.-C. Huang, “Psychometric comparisons of the timed up

and go, one-leg stand, functional reach, and tinetti balance measures in community-dwelling older people," *Journal of the American Geriatrics Society*, vol. 52, no. 8, pp. 1343–1348, 2004.

[30] F. Dobson, R. S. Hinman, E. M. Roos et al., *Recommended Performance-Based Tests to Assess Physical Function in People Diagnosed with Hip or Knee Osteoarthritis*, University of Melbourne, Melbourne, Australia, 2011.

[31] J. M. Blennerhassett and V. M. Jayalath, "The Four Square Step Test is a feasible and valid clinical test of dynamic standing balance for use in ambulant people poststroke," *Archives of Physical Medicine and Rehabilitation*, vol. 89, no. 11, pp. 2156–2161, 2008.

[32] W. Dite and V. A. Temple, "A clinical test of stepping and change of direction to identify multiple falling older adults," *Archives of Physical Medicine and Rehabilitation*, vol. 83, no. 11, pp. 1566–1571, 2002.

[33] T. Broekmans, D. Gijbels, B. O. Eijnde et al., "The relationship between upper leg muscle strength and walking capacity in persons with multiple sclerosis," *Multiple Sclerosis Journal*, vol. 19, no. 1, pp. 112–119, 2013.

[34] S. C. Cobb, D. M. Bazett-Jones, M. N. Joshi, J. E. Earl-Boehm, and C. R. James, "The relationship among foot posture, core and lower extremity muscle function, and postural stability," *Journal of Athletic Training*, vol. 49, no. 2, pp. 173–180, 2014.

[35] L. Hasselgren, L. L. Olsson, and L. Nyberg, "Is leg muscle strength correlated with functional balance and mobility among inpatients in geriatric rehabilitation?" *Archives of Gerontology and Geriatrics*, vol. 52, no. 3, pp. e220–e225, 2011.

[36] A. C. Alonso, N. M. S. Luna, L. Mochizuki, F. Barbieri, S. Santos, and J. M. D. GreveI, "The influence of anthropometric factors on postural balance: the relationship between body composition and posturographic measurements in young adults," *Clinics*, vol. 67, no. 12, pp. 1433–1441, 2012.

[37] P. Allard, M.-L. Nault, S. Hinse, R. LeBlanc, and H. Labelle, "Relationship between morphologic somatotypes and standing posture equilibrium," *Annals of Human Biology*, vol. 28, no. 6, pp. 624–633, 2001.

[38] W. Berger, M. Trippel, M. Discher, and V. Dietz, "Influence of subjects' height on the stabilization of posture," *Acta Oto-Laryngologica*, vol. 112, no. 1, pp. 22–30, 1992.

[39] P. X. Ku, N. A. Abu Osman, A. Yusof, and W. A. B. Wan Abas, "Biomechanical evaluation of the relationship between postural control and body mass index," *Journal of Biomechanics*, vol. 45, no. 9, pp. 1638–1642, 2012.

[40] O. Hue, M. Simoneau, J. Marcotte et al., "Body weight is a strong predictor of postural stability," *Gait & Posture*, vol. 26, no. 1, pp. 32–38, 2007.

[41] T. E. Shubert, L. A. Schrodt, V. S. Mercer, J. Busby-Whitehead, and C. A. Giuliani, "Are scores on balance screening tests associated with mobility in older adults?" *Journal of Geriatric Physical Therapy*, vol. 29, no. 1, pp. 33–39, 2006.

[42] J. J. Keysor, J. E. Dunn, C. L. Link, F. Badlissi, and D. T. Felson, "Are foot disorders associated with functional limitation and disability among community-dwelling older adults?" *Journal of Aging and Health*, vol. 17, no. 6, pp. 734–752, 2005.

[43] D. S. F. Yu, D. T. F. Lee, and N. W. Man, "Fatigue among older people: a review of the research literature," *International Journal of Nursing Studies*, vol. 47, no. 2, pp. 216–228, 2010.

[44] N. C. P. Avelar, A. C. Bastone, M. A. Alcântara, and W. F. Gomes, "Effectiveness of aquatic and non-aquatic lower limb muscles endurance training in the static and dynamic balance of elderly people," *Revista Brasileira de Fisioterapia*, vol. 14, no. 3, pp. 229–236, 2010.

[45] A. Aminian and B. J. Sangeorzan, "The anatomy of cavus foot deformity," *Foot and Ankle Clinics*, vol. 13, no. 2, pp. 191–198, 2008.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 328142, 15 pages
http://dx.doi.org/10.1155/2015/328142

Review Article

Mobile Phone-Based Joint Angle Measurement for Functional Assessment and Rehabilitation of Proprioception

Quentin Mourcou,[1,2,3] Anthony Fleury,[2,3] Bruno Diot,[1,4] Céline Franco,[1] and Nicolas Vuillerme[1,5,6]

[1]*Univ. Grenoble-Alpes, AGIM, La Tronche, France*
[2]*Univ. Lille, 59000 Lille, France*
[3]*Mines Douai, URIA, 59508 Douai, France*
[4]*IDS, Montceau-les-Mines, France*
[5]*Institut Universitaire de France, Paris, France*
[6]*Laboratory for Ergonomics and Work-related Disorders, Center for Sensory-Motor Interaction (SMI), Department of Health Science and Technology, Univ. Aalborg, Denmark*

Correspondence should be addressed to Quentin Mourcou; qmourcou@gmail.com and Anthony Fleury; anthony.fleury@mines-douai.fr

Received 24 April 2015; Revised 22 June 2015; Accepted 2 July 2015

Academic Editor: Massimiliano Pau

Assessment of joint functional and proprioceptive abilities is essential for balance, posture, and motor control rehabilitation. Joint functional ability refers to the capacity of movement of the joint. It may be evaluated thereby measuring the joint range of motion (ROM). Proprioception can be defined as the perception of the position and of the movement of various body parts in space. Its role is essential in sensorimotor control for movement acuity, joint stability, coordination, and balance. Its clinical evaluation is commonly based on the assessment of the joint position sense (JPS). Both ROM and JPS measurements require estimating angles through goniometer, scoliometer, laser-pointer, and bubble or digital inclinometer. With the arrival of Smartphones, these costly clinical tools tend to be replaced. Beyond evaluation, maintaining and/or improving joint functional and proprioceptive abilities by training with physical therapy is important for long-term management. This review aims to report Smartphone applications used for measuring and improving functional and proprioceptive abilities. It identifies that Smartphone applications are reliable for clinical measurements and are mainly used to assess ROM and JPS. However, there is lack of studies on Smartphone applications which can be used in an autonomous way to provide physical therapy exercises at home.

1. Introduction

Joint movement and sensorimotor control can actually be assessed with range of motion and proprioception measurements. Range of motion (ROM), which is the measurement of the extent of a movement of a joint, is used to evaluate and classify joints impairments in patients or the efficacy of certain rehabilitation program. It could be performed in various ways such as, for example, a simple visual estimation or high speed cinematography, in passive or active condition. It is well recognized that proprioceptive function is crucially important for balance, posture, and motor control. Proprioception, which can be defined as the perception of the position and of the movement of various body parts in space, is generally composed by the two following modalities: joint position sense and the sensation of limb movement. On the one hand, "joint position sense" (JPS) is relative to the awareness of the position of the members or segments against each other [1]. On the other hand, "kinaesthesia" is defined as the sensation of the motion to locate the different parts of the body and to evaluate their movement (velocity and direction) and the static part is named statesthesia. Proprioceptive alterations resulting either from diseases, accidents, trauma, surgery, or normal ageing may lead to necessitating specific rehabilitation to prevent injuries and reduce balance deficits. Indeed, it has been shown that proprioception is more important than vision to maintain balance in elderly people population and a decrease in proprioception increases the risk of falling [2, 3].

Moreover, proprioceptive physical activities have previously been shown to improve balance control in elderly [4].

Clinicians and clinical researchers usually employ specific and dedicated methods and tools to measure and improve proprioceptive function. Effects of therapies and robot-based rehabilitation therapies on proprioceptive function are well explored [5], as the assessment of proprioception with all its testing methods [6]. To measure JPS, clinicians mainly use apparatus such as goniometers, inclinometers, or video in a controlled environment under the direction of a medical staff member. Recently, new measurement tools appear with the noteworthy particularity to be based on an unavoidable object of daily living: the Smartphone. Smartphones have led to significant improvements in healthcare systems [7, 8]. They have the advantage of being small, easy to use, affordable, and connected and of including an inertial motion unit (IMU) composed of 3D accelerometer, magnetometer, and gyroscope almost as standard (only one of gyroscopes or magnetometers can be absent). Interestingly, these built-in inertial sensors allow detecting and monitoring both linear and angular movements of the phone. To provide this range of motion, fusion algorithms can be used and some of them have already been validated in several studies with dedicated IMU systems [9]. Smartphones can then be used as goniometric tools [10]. However, Smartphones also contain additional standard technologies such as a screen display, an audio system, or a tactile feedback system for the interaction with the user of the device. In this context, it could be used independently to perform self-measurements during home-based training. At this point, however, to the best of our knowledge, no review has been published on validated Smartphone tool allowing self-measurement and/or providing physical therapy exercises which may be performed in an autonomous way. Along these lines, the present paper was designed to report Smartphone applications that are currently used for measuring and improving proprioceptive abilities.

The remaining of this paper is organized as follow. Section 2 describes related works for the assessment of functional and proprioceptive function abilities using Smartphone-based system. Then, advantages of autonomous self-measurement and training are discussed in Section 3. Current and future works are presented in Section 4. Conclusions are finally drawn in Section 5.

2. Related Works on Smartphone-Based Systems

2.1. Clinical Assessment. Assessment and training of functional and proprioceptive abilities are based on a variety of tests for ROM, JPS, kinaesthesia, force sense, and balance [39]. Passive and active conditions can be used to, respectively, bias joint mechanoreceptors or stimulating joint and muscle-tendon mechanoreceptors [40].

Assessing ROM is used to quantify baseline limitations of motion. It has been demonstrated that ROM measures depend on the number of degrees of freedom of the joint, the initial position, the direction of the movement, and diurnal variation [41].

Assessing JPS consists in different exercises of joint position matching during which the patient is asked to get back to a specified angular position from a neutral one without using visual information. Different factors are likely to influence performances of such a test (e.g., time spent in the expected position [42]) and it has been shown that passive matching, from such range of motion measurement, is more difficult to measure reliably than active one [43]. Such a test is usually performed by a physiotherapist using goniometer to measure joint angle.

Studies about clinical assessment of proprioceptive function, mainly JPS, are briefly presented in Table 1 and described according to the following plan: (Section 2.2) spine proprioception assessment, (Section 2.3) upper extremity proprioception assessment, and (Section 2.4) lower extremity proprioception assessment.

2.2. Spine Proprioception Assessment

2.2.1. Cervical Spine Proprioception Assessment. For the cervical range of motion (CROM), JPS or ROM testing is used to measure head repositioning accuracy which leads to great errors for people with neck disorders [44, 45]. A laser-pointer is mounted at the top of individual's head and a handled button can switch it on to mark head position before and after head rotation. Differences in the placement of the marks are measured in millimeters and can be calculated in degrees or directly recorded in degrees with certain devices [46].

Active and passive CROM can be measured with Smartphone. Two recent studies have assessed validity and reliability of Smartphone applications to measure CROM [11, 12].

The first one used two commercial applications on an iPhone 4 and an iPhone 3GS: Clinometer (Plaincode Software Solutions, Stephanskirchen, Germany) and Compass (Apple, Cupertino, USA) [11]. These apps were compared to a specific CROM gold standard device compound of eyeglasses with three inclinometers placed at three different positions: one near the left ear for flexion/extension (sagittal plan) and another for the lateral flexions on the forehead (frontal plane). Both are gravity dependent. The last positioning was the top of the head. For this one, the magnetic dependence was compensated by placing an adapted brace. Measures were taken with the Smartphone placed on left and right side of the head aligned with the ear or with the eyes depending of the observed motion. Head rotations were measured with the iPhone placed on individual's head with the arrow aligned with the nose. Each participant performed maximal neck movement for each rotation, flexion, and extension. Twenty-eight healthy volunteers (23 ± 6 years) were observed by two different groups of two students in physical therapy. Results were analyzed with intraclass correlation coefficient (ICC) for validity and reliability. For concurrent validity, ICC values were between 0.50 and 0.65 but <0.50 for rotation. For intraobserver reliability, ICC values were between 0.65 and 0.85. For interobserver reliability, ICC values were under 0.60. Authors concluded that Smartphones have good intrarater reliability but lower interrater reliability. Validity is good for movements in sagittal and frontal planes but poor for rotation. This scientific work validates the use of

Table 1: Characteristics of the different studies that have been examined.

Study	Population (Sample size and age)	App	Reference	Body segment	Movement	Type	Results: Validity	Results: Reliability: Intraobserver	Results: Reliability: Interobserver
Tousignant-Laflamme et al., 2013 [11]	Healthy volunteers (28, 23 ± 6 years)	Clinometer and Compass on iPhone	Eyeglasses with three inclinometers	Head	Flexion/extension/rotation	Active CROM	ICC between 0.50 and 0.65 but <0.50 for rotation	ICC = 0.65–0.85	ICC < 0.60
Quek et al., 2014 [12]	Healthy volunteers (21, 31 ± 9 years)	Customized Android	3D motion analyses system	Head	Flexion/extension/rotation	Active CROM	ICC = 0.53–0.98, Spearman's ρ = 0.52–0.98	ICC = 0.82–0.90 but poor in rotation: ICC = 0.05–0.33	N/A
Shin et al., 2012 [13]	Healthy volunteers (41, 52.7 ± 17.5 years)	Clinometer on Android	Double armed goniometer	Shoulder	Forward flexion/abduction/external and internal rotation and abduction	Passive and active ROM	LOA = 10–40° ICC(3, 1) > 0.72 Pearson's correlation coefficient (PCC) = 0.79–0.97	ICC(2, 1) > 0.70 except for ICC(2, 1) = 0.63–0.68 (internal rotation at 90° abduction)	ICC(3, 1) > 0.90
Werner et al., 2014 [14]	Healthy volunteers (24) and symptomatic patients (15)	Clinometer on iPhone	Visual estimation and standard goniometer	Shoulder	Forward flexion/abduction/external and internal rotation	Passive ROM	For healthy volunteers, ICC > 0.60 and SEM < 4.3° For symptomatic patients, ICC > 0.80 and SEM < 0.1°	N/A	For healthy volunteers, ICC > 0.60 and SEM < 10.1° For symptomatic patients, ICC > 0.60 and SEM < 5.8°
Mitchell et al., 2014 [15]	Healthy volunteers (94)	GetMyRom and DrGoniometer on iPhone	Standard goniometer	Shoulder	External rotation	Active ROM	ICC = 0.94	ICC = 0.94	ICC = 0.79
Oïhénart et al., 2012 [16]	Healthy volunteers (5)	iShould on iPhone or iPod touch	3D kinematic sensors	Shoulder	Anterior elevation and extension/abduction/adduction/internal and external rotation	Active ROM	Mean difference (%): 1.09 for RAV and 0.60 for *P* score	N/A	N/A
Johnson et al., 2015 [17]	Healthy volunteers (1)	Customized Android	Double-armed goniometer and full-scale motion capture	Shoulder	Abduction	Passive ROM	Mean difference in B&A plots = −1.7° (seated position) and 1.4° (supine position)	CCC > 0.992	CCC > 0.989
Ferriero et al., 2011 [18]	Healthy volunteers (28 pictures!)	DrGoniometer on iPhone	Plastic universal goniometer	Elbow	Elbow placed at different angles	Passive ROM	LOA = +4.51°, −5.75°	ICC = 0.998	ICC = 0.998
Ockendon and Gilbert, 2012 [19]	Healthy volunteers (5, 30–40 years)	Knee goniometer on iPhone	Telescopic-armed goniometer	Knee	Flexion	Passive ROM	LOA = 15.2° PCC = 0.947	PCC = 0.982	PCC = 0.994
Hambly et al., 2012 [20]	Healthy volunteers (96, 31 ± 11 years)	Knee goniometer on iPhone	Telescopic-armed goniometer	Knee	Flexion	Active ROM	PCC = 0.932 paired t-test indicates significant difference but not clinically significant	ICC = 0.894 PCC = 0.795	N/A
Jones et al., 2014 [21]	Healthy volunteers (36, 60.6 ± 6.2 years)	Simple goniometer on iPhone	Universal goniometer	Knee	Lunge forward	Active ROM	LOA = 13.1° r = 0.96–0.98	ICC = 0.97–0.99	N/A
Milanese et al., 2014 [22]	Healthy volunteers (6)	Knee goniometer on iPhone	Universal goniometer	Knee	Flexion	Passive ROM	CCCexpert = 0.982 CCCnovice = 0.983 CCCensemble = 0.991 CPP = −0.51 SEM < 2.7° (gonio) SEM < 1.4° (apps)	CCCexpert > 0.998 CCCnovice > 0.997	CCCexpert > 0.996 CCCnovice > 0.998 CCCtogether = 0.997

Table 1: Continued.

Study	Population (Sample size and age)	App	Reference	Body segment	Movement	Type	Results: Validity	Results: Reliability: Intraobserver	Results: Reliability: Interobserver
Rwakabayiza et al., 2013 [23]	Healthy volunteers (20) and symptomatic patients (20)	Knee goniometer on iPhone	Universal goniometer	Knee	Flexion/extension	Active and passive ROM	On 6 patients ICC = 0.54 (Smartphone) in active extension ICC = 0.92 (Smartphone) in active flexion (better than goniometer)	ICChealthy = 0.85 (0.75–0.94) ICCsymptomatic = 0.98 (0.96–0.99)	ICChealthy = 0.12 (0.00–0.25) ICCsymptomatic = 0.24 (0.00–0.45)
Bruyneel and Bridon, 2015 [24]	Healthy volunteers (41, 18–26 years)	Clinometer on iPhone	Hand-held bilevel inclinometer	Knee	Extension	Passive ROM	N/A	ICC > 0.76 (inclinometer) ICC > 0.72 (Smartphone) MDC < 5°	ICC > 0.64 (inclinometer) ICC > 0.64 (Smartphone) MDC < 8°
Jenny 2013 [25]	Symptomatic patients (10, 69 years)	Angle on iPhone	Navigation system	Knee	Flexion/extension	Passive ROM	LOA = 27.4° t-test not significant Spearman's = 0.99 B&A: good coherence	ICC = 0.81	ICC = 0.79
Ferriero et al., 2013 [26]	Healthy volunteers (1)	DrGoniometer on iPhone	Universal goniometer	Knee	Different knee angles measurement	Passive ROM	LOA = −7.5°/+10.71° B&A	ICC = 0.958	ICC = 0.994
Jenny et al., 2015 [27]	Symptomatic patients (10, 69 ± 10.8)	Goniometer Pro and DrGoniomter on iPhone	Navigation system	Knee	Flexion/extension	Passive ROM	Strong correlation and a good coherence (Levene's test, ANOVA test, Wilcoxon's test, Kendall's test, Spearman's test, and B&A). Inclinometer results differed	N/A	N/A
Andrea et al., 2014 [28]	Symptomatic patients (35)	"SmartJoint" Android and iOS application	KT 100	Knee	Lachman's test	NC	Mean ICC = 0.797 (uninvolved knee) ICC = 0.987 (involved knee)	ICC = 0.973 (uninvolved knee) ICC = 0.989 (involved knee)	ICC = 0.957 (uninvolved knee) ICC = 0.992 (involved knee)
Yoon et al., 2014 [29]	Healthy volunteers (10, 22.2 ± 1.69 years)	TiltMeter on iPhone	Digital inclinometer	Hip	Measure of femoral neck anteversion	Passive ROM	Similarly ICC	ICC(2, 3) = 0.95 SEM = 1.9°–2.2°	ICC(2, 3) = 0.85 SEM = 4.1°
Peters et al., 2012 [30]	Symptomatic patients (50, 67 years from 31 to 84)	Angle and Camera Protractor on iPhone	X-Rays	Hip	Measurements for total hip arthroplasty	NC	Differences are under 5% between pre- and postop	N/A	N/A

Table 1: Continued.

Study	Population (Sample size and age)	App	Reference	Body segment	Movement	Type	Results: Validity	Results: Reliability: Intraobserver	Results: Reliability: Interobserver
Charlton et al., 2015 [31]	Healthy volunteers (20, 23.8 ±4.6 years)	Hip ROM tester	Camera marker based 3DMA	Hip	Flexion/abduction/adduction/supine internal and external rotation	Passive ROM	ICC(2, 3) > 0.88 (for 6 movements) ICC(2, 3) = 0.71 (for supine external rotation)	ICC(2, 3) > 0.84 (for 4 movements) ICC(2, 3) = 0.63–0.68 (for 3 movements)	N/A
Vohralik et al., 2014 [32]	Healthy volunteers (20, 21–28 years)	iHandy Level on iPhone	Digital inclinometer	Ankle	Ankle dorsiflexion range	Active ROM	CPP = 0.99	ICC(2, 1) = 0.97	ICC(2, 1) = 0.76
Williams et al., 2013 [33]	Healthy volunteers (20, 40 ± 12 years)	TiltMeter on iPhone	Digital inclinometer	Ankle	Weight bearing lunge test	Active ROM	ICC = 0.83	ICC(2, 1) = 0.81–0.85	ICC(2, 2) = 0.80–0.96
Kolber et al., 2013 [34]	Healthy volunteers (30, 25.6 ± 2.1 years)	iHandy Level on iPhone	Bubble inclinometer	Spinal	Thoracolumbopelvic flexion, isolated lumbar flexion, thoracolumbopelvic extension right lateral flexion, and left lateral flexion	Active ROM	ICC(3, k) > 0.86 LOA = 18–30°	ICC(3, k) > 0.80	ICC(2, k) > 0.81
Izatt et al., 2012 [35]	NC (8 torso)	Scoligauge on iPhone	Scoliometer	Spinal	NC	Passive	LOA = 6.2°	ICC 95% = ±3.2°	ICC = 0.92 (absolute agreement definition) IC 95% = ±4.9°
Franko et al., 2012 [36]	NC (60 angles)	Scoligauge on iPhone	Scoliometer	Spinal	Sixty angles randomly selected	Passive	CPP = 0.99	N/A	N/A
Balg et al., 2014 [37]	Symptomatic patients (34)	Scoligauge on iPhone	Scoliometer	Spinal	Scoliosis angles	Active	ICC = 0.947 Mean difference = 0.4°	ICC = 0.961 Mean difference = 0.0°	ICC = 0.901 Mean difference = 0.1°
Qiao et al., 2014 [38]	Symptomatic patients (64, 15.7 year)	Scoligauge on iPhone	Scoliometer	Spinal	Scoliosis angles	Active	Mean angles are similar	ICC (scoliometer) = 0.954 ICC (apps) = 0.965	ICC (scoliometer) = 0.943 ICC (apps) = 0.964

a Smartphone application for the active angle measurement of neck in the sagittal and frontal planes.

The second study used a customized Android application compared with a validated gold standard three-dimensional motion analysis system (VICON, UK) [12]. It recorded motion with three reflective markers tracked by VICON Nexus V1.7.1 and a 9-camera VICON MX motion analysis system. The phone was mounted on a helmet to capture head flexion, extension, and rotation. A magnetic yoke was placed around participant neck to compensate magnetic dependence. Twenty-one healthy participants were recruited and sixteen of them come for a second session. Participants were instructed to perform each movement with a manual guidance provided by the single examiner who determined the end of CROM. Results were analyzed with Spearman's correlation, ICC, and Bland and Altman plots (B&A) for validity. Reliability was assessed with $ICC_{(3,3)}$, ordinary least products (OLP) regression, Standard Error of Measurement (SEM), Limit of Agreement (LOA), and Minimal Detectable Change (MDC). Validity results showed ICC values between 0.53 and 0.98, with a Spearman correlation coefficient ranged between 0.52 and 0.98. Intraobserver reliability was revealed by ICC values between 0.82 and 0.90 but under 0.33 for rotation. Authors established the validity and intrarater reliability for movements in sagittal and frontal planes but not rotation "likely due to magnetic field interference" [12]. This scientific work validates the use of a Smartphone application for the active angle measurement of neck in the sagittal and frontal planes.

2.2.2. Lumbar Spine Proprioception Assessment. Investigation of the reliability and validity of Smartphone application for measuring spinal range of motion was explored in five studies [34–38].

Kolber et al. compared the use of the Smartphone application iHandy Level (iHandy Inc.) on an iPhone 4 (Apple, Cupertino, USA) and a gravity-based bubble inclinometer (model 12-1056, Fabrication Enterprises, White Plains, New York, USA) [34]. Thirty healthy participants were recruited and observed by two examiners. Five active types of spinal range of motion measurements were taken: thoracolumbopelvic flexion, isolated lumbar flexion, thoracolumbopelvic extension right lateral flexion, and left lateral flexion. Reliability was assessed using $ICC_{(3,k)}$ for intrarater and ICC model 2 for interrater. Mean, SEM, and MDC were also calculated. Main results for validity presented $ICC_{(3,k)}$ values > 0.86. ICC values for intra- and interobserver reliability are, respectively, greater than 0.80 and 0.81. Authors concluded that "the iHandy Level application on the iPhone is both reliable and comparable to bubble inclinometry" [34]. This scientific work validates the use of a Smartphone application for the active angle measurement of lumbar spinal range of motion.

Another routine clinical angle measurement is the angle of thorax rotation or rib hump, which is important for patients with scoliosis. Izatt et al. [35] and Franko et al. [36] evaluated a Smartphone application, Scoligauge (Ockendon Partners Ltd, UK), on an iPhone (Apple, Cupertino, USA) compared to the standardize scoliometer. For the first study [35], eight plaster torsos were used for measurements performed by nine examiners (four experienced spinal orthopedic surgeons, a specialist physiotherapist, an experienced spinal orthotist, two training grade registrars, and an inexperienced physiotherapist). Plaster torsos were placed on a standard bench during passive measurements. Intra- and interobserver variability were assessed by using mean absolute difference and 95% Confident Interval (CI) and ICC. Limit of Agreement (LOA) value was 6.2°. For intraobserver reliability, CI value was ±3.2° and ±4,9° for interobserver with an ICC value equal to 0.92. Authors concluded that "clinical judgements as a result of iPhone rib hump measurements can be made with confidence based on readings taken from the iPhone when combined with the acrylic sleeve" [35]. In the second study, sixty angles were randomly selected and measured by four orthopaedic medical providers [36]. Validity was confirmed using Pearson's correlation coefficient (CPP), whose result was equal to 0.99. Authors concluded that "the Scoligauge app is a convenient novel tool that replicates the function of a standard clinical scoliometer but with a potentially decreased financial cost and greater convenience for providers" [36]. These scientific works validate the use of a Smartphone application for the active angle measurement of thorax rotation.

Two more recent studies evaluated Scoligauge application (Ockendon Partners Ltd., UK) but in clinical case with patients [37, 38]. Balg et al. [37] recruited thirty-four patients with adolescent idiopathic scoliosis and measurements were made by two examiners, one spinal orthopedic surgeon and one physical therapy student. Statistical analysis uses ICC to assess inter- and intraobserver reliability and validity. Bland and Altman plots were also used. For validity, the ICC value was equal to 0.947 and mean difference is 0.4 degrees. Intra- and interobserver reliability were assessed with ICC values of 0.961 and 0.901 and mean differences of 0.0 and 0.1 degrees. Authors concluded that "this study proved that even without an adapter the Scoligauge iPhone application is valid and can be used in the clinical setting for scoliosis evaluation" [37]. Qiao et al. recruited sixty-four patients with adolescent idiopathic scoliosis, in which thirty-two patients had main thoracic scoliosis while the rest had main thoracolumbar/lumbar scoliosis [38]. Measurements were made by two spine surgeons. Cobb angles were measured from posteroanterior radiographs. Patients performed Adam's forward bend test. Each examiner performed two evaluations; retest was done after twenty minutes of interval. Statistical analysis used ICC. Intraobserver ICC values were 0.954 for scoliometer and 0.965 for Scoligauge. Interobserver ICC values were 0.943 for scoliometer and 0.964 for Scoligauge. Authors conclude that "Smartphone-aided measurement for ATR showed excellent reliability, and the reliability of measurement by either scoliometer or Scoliogauge was influenced by Cobb angle where reliability was better for curves with larger Cobb angles" [38]. These scientific works validates the clinical use of a Smartphone application for the active angle measurement of spinal range of motion with patient.

2.3. Upper Extremity Proprioception Assessment

2.3.1. Shoulder Proprioception Assessment. Functional and proprioceptive abilities on extremities such as shoulder usually use universal goniometer to measure active or passive range of motion. However, laser-pointer devices can also be a solution for this specific joint [47]. Laser pointer can be used to measure, in millimeter, differences between joint movements from a position to another and then calculate joint position in degree like universal goniometer. JPS tests can evaluate the ability of the individuals to reproduce a specific movement and the precision to access a specific angle target. These tests must be conducted with full knowledge of their limitations. It can involve some cognitive component, and the size and speed of movement should be standardized [39]. Active and passive shoulder ROM can therefore be measured with Smartphone. Six studies can be listed with the aim of studying validity and reliability of Smartphone applications for shoulder ROM [13–17, 48]. These studies will be described in the next paragraphs.

Shin et al. used a previously cited commercial application, on a Samsung Galaxy S: Clinometer [13]. The application was compared with a standard double-arm goniometer. The Smartphone was attached on the wrist (ventral side of the forearm) with the help of an armband. Observed shoulder passive and active movements were forward flexion, abduction, external rotation with the arms at the sides, external rotation at 90 degrees abduction, and internal rotation at 90 degrees abduction. Forty-one volunteers were observed by two orthopedic resident doctors and one orthopedic surgeon. Reliability was evaluated with $ICC_{(2,1)}$ for interobserver and $ICC_{(3,1)}$ for intraobserver. Results were greater than 0.70 for intraobserver except for internal rotation at 90° abduction (0.63–0.68). Interobserver's results were greater than 0.90. Validity was evaluated with SEM, MDC, B&A, and PCC. LOA was between 10 and 40°, ICC > 0.72, and PCC between 0.79 and 0.97. Authors concluded that "Smartphone application is reliable compared to the double-arm goniometer, although the between-day reliability remains to be established" [13]. This validates the use of a standard commercial Smartphone application for the active and passive angle measurement of shoulder range of motion.

Werner et al. used the same commercial application, Clinometer, but with another device that is an iPhone (Apple, Cupertino, USA) [14]. Clinometer was compared with a visual estimation and a standard goniometer. Twenty-four healthy adults and fifteen symptomatic patients were recruited. Measurements were performed by 5 examiners (one sport fellowship-trained orthopedic surgeon, one orthopedic sports medicine fellow, one orthopedic resident physician, one orthopedic physician assistant, and one medical student). Passive abduction and forward flexion were measured in standing position. They also measured external rotation with the arm at the patient's side, external rotation with the arm abducted at 90 degrees, and internal rotation with the arm abducted at 90 degrees; all were measured with the patient supine on an examination table. Reliability was evaluated with an $ICC_{(2,1)}$ and this ICC for each measurement modality was compared by use of ANOVA with a Tukey post hoc test. Validity was evaluated using $ICC_{(2,1)}$, B&A, and SEM. For this validity, with healthy participants, ICC > 0.60 and SEM < 4.3°. For symptomatic patients, ICC > 0.80 and SEM < 0.1°. For interobserver reliability, with healthy participants, ICC > 0.60 and SEM < 10.1°. For symptomatic patients, ICC > 0.60 and SEM < 5.8°. Authors concluded that "Smartphone Clinometer has excellent agreement with a goniometer-based gold standard for measurement of shoulder ROM in both healthy subjects and symptomatic patients" [14]. This result validates the use of a standard commercial Smartphone application for a clinical active and passive angle measurement of shoulder range of motion with patients.

Mitchell et al. used two commercial applications, GetMyRom (Interactive Medical Productions, LLC, USA) and DrGoniometer (CDM S.r.L.), with an iPhone device [15]. GetMyRom can measure JPS thanks to orientation sensors while DrGoniometer is a photo-based application that calculates the angles from markers positioned after the measurement. Both applications were compared with a standard goniometer. Ninety-four healthy women were recruited and measurements were made by one novice examiner and one expert at two different moments. Participants were instructed to perform active shoulder external rotation. Reliability, for inter- and intrarater, and validity were evaluated using ICC. Main results were as follows: for validity, ICC is equal to 0.94, for intraobserver, ICC is equal to 0.94, and for interobserver ICC = 0.79. Authors concluded that "both applications were found to be reliable and comparable to SG" [15]. In addition, the author plebiscite the use of the application based on the camera because of its potential for saving images. This scientific work validates the use of two specific Smartphone applications, which used inertial sensors or camera, for the active angle measurement of shoulder range of motion.

Oïhénart et al. used a custom application called iShould (Instrumented Shoulder Test) with iPhone 4 or iPod Touch devices (Apple, Cupertino, USA) [16]. iShould computes kinematics range of angular velocity (RAV) score, which quantifies the shoulder movement based on angular velocities and *P* score, which is based on the power of shoulder movement, directly from inertial signal sensors. The application is compared with another 3D kinematics sensors composed by three miniature capacitive gyroscopes (Analog device, ADXRS 250, 400°/s) and three miniature accelerometers (Analog device, ADXL 210, 5 g). Smartphone and sensors are attached to the anterior part of the humerus, using an armband. Five participants were recruited and performed active anterior elevation and extension, abduction and adduction, and internal and external rotation of the shoulder. Validity is evaluated using mean difference of RAV score and *P* score. Mean difference was 1.09% for RAV and 0.60% for *P* score. Authors concluded that "the application offers then an interesting alternative to the existing system" [16]. This work validates the use of a specific Smartphone application for the active angle measurement of shoulder range of motion.

Johnson et al. used a custom Android mobile application that mimics goniometer with the help of magnetometer on the first generation of Motorola Droid. Their application was compared with a universal standard double-arm goniometer [17] or with a full-scale motion capture system [48]. The Smartphone was used to collect angle in the same manner as the standard goniometer. Four passive shoulder's abductions were simulated in both seated and supine orientation. Only one participant was recruited for this pilot study and three therapists managed the measurement. Statistical analysis was performed with ANOVA, PCC, concordance correlation coefficient (CCC), and scatter plot were used to assess agreements, and B&A plots were used to compare differences. For validity, mean differences in B&A plots were −1.7° (in seated position) and 1.4° (in supine position). Intra- and interobserver reliability were presented with CCC values of 0.992 and 0.989. Authors concluded that "this study demonstrates the validity of the Smartphone goniometer application utilizing a built-in 3-axis magnetometer sensor when compared with a previously proven and universal goniometer. The Smartphone magnetometer-based goniometer also demonstrates comparably high reliability in measuring passive shoulder abduction ROM in both the seated and supine positions" [17]. These scientific works allow validating the use of a specific Smartphone application for the passive angle measurement of shoulder range of motion.

2.3.2. Elbow Proprioception Assessment. Ferriero et al. also used the "DrGoniometer" application for elbow angle measurement [18]. This application works on iPhone and it was compared to a small plastic universal goniometer. In this proof-of-concept study, one participant was recruited and seven examiners assess measurements. Twenty-eight pictures of elbows of healthy subjects were taken at different angles, and this complete protocol was repeated a second time after one week. ICC was used for intrarater and interrater reliability. Validity was interpreted with a LOA between +4.51° and −5.75°. Intra- and interobserver ICC values were 0.998 for both. Authors concluded that the application "is reliable for elbow joint goniometry" [18]. This scientific work validates the use of a specific Smartphone application, which used camera, for the passive angle measurement of elbow range of motion.

2.4. Lower Extremity Proprioception Assessment

2.4.1. Knee Proprioception Assessment. Functional and proprioceptive abilities tests for knee extremity usually use universal goniometers, in clinical practice, to measure active or passive ROM and evaluate the ability of individuals to reproduce a specific movement and the precision to reach a specific angle. Active and passive knee range of motion can therefore be measured using Smartphones. Ten researches can be listed with the aim of studying validity and reliability of Smartphone applications for knee ROM [19–28]. The next paragraph will briefly describes these studies.

Ockendon and Gilbert created their own Smartphone application called "Knee Goniometer" which is now published on the Apple Inc. App Store. The application is installed on an iPhone 3GS (Apple, Cupertino, USA) and compared with a telescopic-armed goniometer (Lafayette Instrument, Lafayette, IN) [19]. Five healthy participants were recruited and measurement was performed by two experienced and independent examiners. Each participant executed three different passive knee flexions, which were measured twice, separated by a time interval, on both right and left legs. Statistical analysis included B&A plot, Scatter plots, standard deviation (SD) of the difference, and PCC. For validity, LOA was equal to 15.2° and PCC was equal to 0.947. Intra- and interobserver main results for PCC were, respectively, 0.982 and 0.994. Authors concluded that "the iPhone goniometer [is] a reliable tool for the measurement of subtle knee flexion in the clinic setting when compared with the current standard bedside technique" [19]. This validates the use of a specific Smartphone application for the passive angle measurement of knee range of motion.

The "Knee Goniometer" application was also used to measure level of agreement with a goniometer for the assessment of maximum active knee flexion by an inexperienced tester [20]. An iPhone 3GS (Apple, Cupertino, USA) was also compared to a telescopic-armed goniometer (Lafayette Instrument, Lafayette, IN). Ninety-six healthy participants were recruited and measurements were performed by a graduate sports therapist inexperienced examiners. Participants were asked to perform three full active knee flexion movements from full knee extension to maximum knee flexion. Statistical analysis was conducted with a PCC test, a two-tailed paired t-test, an ICC to evaluate intratester reliability, and finally B&A. The paired t-test indicated a significant difference in results but it was not considered clinically significant. Intraobserver reliability is interpreted by an ICC value of 0.894 and a PCC value of 0.795. Authors concluded that "The iGoniometer demonstrated acceptable test-retest reliability and criterion validity for an inexperienced tester with healthy participants. There was a statistically significant difference between the iGoniometer and long arm goniometer measurements but this was not considered to be a level of difference that would have a clinical impact" [20]. This work validates the use of a specific Smartphone application with inexperienced testers for the active angle measurement of knee range of motion.

Matthew Ockendon has published another Smartphone application, called "Simple goniometer," which also mimics standard two-arm goniometer. However, this application has no specific interface for knee JPS unlike "Knee Goniometer." This application was used on an iPhone 3GS (Apple, Cupertino, USA) to assess its validity and reproducibility for JPS knee test compared with a universal goniometer [21]. Thirty-six healthy participants were recruited and measurements were made by two registered physiotherapists experienced in using the universal goniometer. Participants were instructed to actively and gently lunge forward with their dominant leg and remember the angle. They are then asked to return to the original position and to start again to reproduce the target angle. This was performed three times. Statistical analysis for reliability was determined using confidence intervals and for validity using PCC, $ICC_{(3,k)}$, and B&A plots. Validity is interpreted with LOA results of 13,1° and PCC ranged from 0.96 to 0.98. Intraobserver ICC was between 0.97 and 0.99. Authors concluded that "the scores obtained from the simple Goniometer app for iPhone showed that there were concurrent validity and reliability for knee joint angle as compared with the universal goniometer" [21]. This scientific work allows validating the use of a specific Smartphone application for the active angle measurement of knee range of motion.

The "Knee Goniometer" application was also used to assess its validity and reliability beside experienced and novice clinicians [22]. It was installed on an iPhone 4 (Apple, Cupertino, USA) and compared to the universal goniometer as a gold standard. Six healthy students were recruited and all goniometric measurements were performed by six independent examiners. Participants were placed into passive knee flexion, always with the right leg. Each examiner made one measurement with both universal goniometer and Smartphone to each participant during one session. Three sessions were made with a fifteen-minute break between each one. Statistical analysis included the calculation of concordance correlation coefficient (CCC), SEM, Scatter plots, and B&A plots. For validity, CCC for expert was 0.982, CCC for novice was 0.983, CCC for all was 0.991, and CPP was −0.51. SEM was under 2.7° for goniometer and under 1.4° for the Smartphone application. In intraobserver reliability, CCC values varied between 0.998 and 0.999 for expert (clinicians) and varied between 0.997 and 0.999 for novice (students). In interobserver reliability, CCC was greater than 0.996 for expert, 0.998 for novice, and 0.997 for both. Authors concluded that "this study established that both the universal goniometer and the Knee Goniomter application were reliable for measurement of knee flexion angles by experienced clinicians and final year physiotherapy students using standardized protocols" [22]. This scientific work encourages the use of a specific Smartphone application for the passive angle measurement of knee range of motion.

Rwakabayiza et al. also used the "Knee Goniometer" application made by Ockendon for iPhone (Apple, Cupertino, USA) [23]. Universal goniometer is still used as a gold standard. Twenty healthy participants were recruited and twenty patients in acute postoperative knee prosthesis phase also participate. Measurements were realized by one specialist in orthopedic surgery, one physiotherapist, and one assistant doctor. Active and passive flexion-extension knee amplitudes were measured with the Norkin and White technique three times by each examiner. ICC for intra- and interobserver reliability was calculated. Validity results were produced on six patients, for an ICC value of 0.54 on Smartphone application in active extension and an ICC value of 0.92 for Smartphone application in active flexion. These results were better than goniometer's. Intraobserver reliability revealed a mean in ICC of 0.85 for healthy participants and 0.98 for patients. Interobserver reliability revealed a mean in ICC of 0.12 for healthy participants and 0.24 for patients. These last two results were considered "bad" by authors and with the Smartphone application with the goniometer as well. This prompted them to change their protocol to avoid fatigue-related bias. Authors concluded that "this study shows that the "Knee Goniometer" Smartphone application can be used in clinical practice as well as the universal standard goniometer to measure the range of motion of the knee" [23]. It provides, in addition to others, validation on patients in postoperative phase. This scientific work allows validating the clinical use of a specific Smartphone application for the active and passive angle measurement of knee range of motion with patients.

"Clinometer" application was used not only in shoulder but also in knee ROM [24]. It was set up on an iPhone 4 (Apple, Cupertino, USA) and used in comparison with a hand-held bilevel inclinometer as a gold standard. Forty-one healthy students were recruited along with two examiners. JPS was measured using passive knee extension. Each examiner had measured this extension three times with both instruments twice, with a break of one day. Statistical analyses were performed using Mann-Whitney test, $ICC_{(3,1)}$, SEM, and MDC. Intraobserver results were ICC > 0.76 for the inclinometer, ICC > 0.72 for Smartphone application, and MDC < 5°. Interobserver results were ICC > 0.64 for the inclinometer, ICC > 0.64 for Smartphone application, and MDC < 8°. Authors concluded that "the results obtained at the level of the knee joint have similar characteristics between [the] two tools" [24]. This scientific work allows validating the use of a standard commercial Smartphone application for the passive angle measurement of knee range of motion.

Another clinical research was performed by Jenny with the "Angle" (Smudge App) Smartphone application [25]. A navigation system (OrthoPilot, Aesculap, Tuttlingen, FRG) was used as a gold standard. Ten patients, operated for end-stage osteoarthritis by navigation assisted for total knee arthroplasty, participated to the study. The knee was passively positioned at four full extensions and at maximal flexion angle. For each set of measurements, six navigated and six Smartphone data sets were obtained. Statistical analyses were performed with paired Student's t-test, and Spearman's coefficient of correlation, Bland and Altman plots, and intra- and interobserver reproducibility were assessed using ICC. Validity results were as follows: LOA was equal to 27.4°, t-test was not significant, Spearman's coefficient of correlation is 0.99, and B&A had good coherence. Intra- and interobserver ICC were, respectively, 0.81 and 0.79. Authors concluded that "the Smartphone application used may be considered as precise and accurate" [25]. This scientific work allows validating the clinical use of a standard commercial Smartphone application for the passive angle measurement of knee range of motion with patients.

Ferriero et al. used the "DrGoniometer" Smartphone photo-based application [26]. They studied its reliability in comparison to a universal goniometer. For the first experiment set, one healthy participant was recruited with four examiners, two experts (physiotherapists) and two novices (first-year physiotherapy students). Passive knee angle flexions were produced by an isokinetic device with the right leg fixed. Each examiner took twenty-five pictures at twenty-degree and eighty-degree knee flexion. The second set of experiments was made with ten healthy individuals assessed by ten examiners. Thirty-five pictures were taken at different knee angle measurements. This set was repeated one week later to evaluate inter- and intrarater correlation. Statistical analysis was carried out using $ICC_{(3,1)}$ for intra- and interrater correlation, and Bland and Altman plot was used to evaluate differences. Resulting LOA ranged from −7.5° to +10.71°. Intra- and interobserver ICC were, respectively, 0.958 and 0.994. Authors concluded that "DrG is a reliable method for knee joint angle measurement [...] the images of the measurement can be included in the patient's medical record as evidence of the quality of the care provided" [26]. This scientific work allows validating the use of a specific Smartphone application which used camera for the passive angle measurement of knee range of motion with patients.

For knee range of motion measurement, two types of Smartphone applications were used: sensor-based application and photo-based application [27]. Jenny et al. compared those two methods using Goniometer Pro (5fuf5) as the sensor-based application, DrGoniometer as the photo-based application, and a navigation system (OrthoPilot, Aesculap, Tuttlingen, FRG) as gold standard. Ten consecutive patients with end-stage osteoarthritis were selected and measurements were made by one examiner. Five measurements were obtained using each application. Statistical analysis was made using ANOVA test, paired difference, Level's test, Wilcoxon's test, Kendall's test, Spearman's test, and Bland and Altman plots. Results led to strong correlation and a good coherence. Authors conclude that "the camera smartphone application used in this study is fit for the purpose of measurement of the knee range of motion in a routine clinical setting and is substantially superior to inclinometer-based measurement" [27]. This scientific work validates the clinical use of a two specific Smartphone applications that used sensors or camera for the passive angle measurement of knee range of motion with patients. It aims to compare these applications and concluded that camera is superior to the sensors-based application. However, it must be noted that photo-based applications are not suitable for self-measurement.

Smartphone application for measuring range of motion in clinical practice was recently extended to new type of measure such as anterior tibial translation in anterior-cruciate ligament (ACL) deficient knees [28]. A specific Smartphone application, running on both Android and iOS, called "SmartJoint" was developed. This study compared this application, installed on both systems, with the arthrometer KT 1000 (Med Demetric, Kentucky, USA). Thirty-five patients with chronic ACL-deficient knees scheduled for ACL reconstruction were selected. Measurements were performed by two independent examiners. The Lachman test was performed three times on each knee with all devices. Statistical analysis used ICC to compare intertest, intraobserver, and interobserver reliability. Results were a mean ICC of 0.797 for uninvolved knee and mean ICC of 0.987 for involved knee; mean ICC of 0.973 for uninvolved knee and mean ICC of 0.989 for involved knee; and mean ICC of 0.957 for uninvolved knee and mean ICC of 0.992 for involved knee, respectively. Authors conclude that "the performance of SmartJoint is comparable and highly correlated with measurements obtained from KT 1000" [28]. This validates the clinical use of a specific Smartphone application for the passive angle measurement in Lachman's test.

2.4.2. Hip Proprioception Assessment. Detecting abnormal Femoral Neck Anteversion (FNA) is important for physiotherapist to identify lower limb problems. Measuring FNA is possible with the angle formed by the vertical line and the tibial crest, when the greater trochanter is most prominent laterally. Yoon et al. compared the reliability of the method to measure FNA, including the comparison between an industrial digital inclinometer (GemRed DBB, Gain Express Holdings, Ltd., Hong Kong, China) as gold standard and an iPhone (Apple, Cupertino, USA) with Tilt-Meter (IntegraSoftHN) application [29]. Nineteen hips were examined in ten healthy subjects observed by two physical therapists. Three sessions of each method were repeated with one hour between sessions. Statistical analysis used ICC, SEM, PCC, and Kolmogorov-Smirnov Z test. Intraobserver $ICC_{(2,3)}$ is 0.95, and SEM ranged from 1.9° to 2.2°. Interobserver $ICC_{(2,3)}$ is 0.85, and SEM is equal to 4.1°. Authors concluded that "using a Smartphone with an inclinometer application during the TCAT showed comparable reliability to a digital inclinometer" [29]. This scientific work validates the use of a standard commercial Smartphone application for the passive angle measurement of hip proprioception.

Smartphone applications could also be used in addition to or instead of conventional techniques and computer-assisted surgery. Peters et al. try to improve acetabular cup orientation in total hip arthroplasty by using Smartphone technology [30]. They used two applications, Angle (Smudge Apps) and Camera Protractor Lite (YJ Soft) on an iPhone (Apple, Cupertino, USA). Angle application directly measures angle with the help of accelerometer while Camera Protractor Lite displays a protractor through the phone camera. Standard postoperative pelvic X-rays are used a gold standard. Fifty patients who need primary total hip arthroplasty operations were selected. Measurement was realized by a surgeon and their first assistant. The Angle application was used for the inclination of the acetabular cup and the Camera Protractor application was used to determine anteversion. Statistical analysis compared differences between intraoperative and postoperative angles. Results showed that differences were less than 5% between before and after operation. Authors concluded that "the use of the iPhone for acetabular cup placements is quick and accurate" [30]. This encourages the clinical use of two standard commercial Smartphone applications for hip angle measurement in surgery.

Reliability and concurrent validity of a Smartphone application to measure hip joint range of motion were assessed by Charlton et al. [31]. Measurements obtained with a custom Smartphone application, called "Hip ROM Tester," were compared with those obtained with a camera marker-based 3DMA system (Vicon, Oxford, UK) and with a bubble inclinometer. Twenty healthy participants were recruited and all tests were conducted by one physiotherapist. These tests were passively performed by movements of flexion, abduction, adduction, supine internal and external rotation, and sitting internal and external rotation. Intratester reliability was performed using ICC, CV, and SEM values. Validity is performed using means, standard deviation, and ICC. Validity tests resulted in $ICC_{(2,3)} > 0.88$ for 6 movements and $ICC_{(2,3)}$ was equal to 0.71 for supine external rotation. Intraobserver tests resulted in $ICC_{(2,3)} > 0.84$ for 4 movements and $ICC_{(2,3)}$ was between 0.63 and 0.68 for 3 movements. Authors concluded that "a Smartphone application provides a reliable and valid method of assessing passive hip joint ROM in young active males" [31]. This scientific work allows validating the use of a specific Smartphone application for the passive angle measurement of hip range of motion.

2.4.3. Ankle Proprioception Assessment. Ankle ROM using Smartphone applications was studied in two studies [32, 33]. The first one compares the iHandy Level app (iHandy Inc.) on an iPhone (Apple, Cupertino, USA) with a digital, medically rated inclinometer (Baseline, Fabrication Enterprises Incorporated, USA) [32]. Twenty participants were recruited and measurements were made by two physiotherapy honor students in their final year of study. The test measures the ankle dorsiflexion range; participants were instructed to lunge forward, bringing their knee in contact with a vertical tape on the wall. Three measurements were performed and mean was used to perform analysis. ICC and CI were used for intra- and interrater reliability; SEM and Bland and Altman plots were also produced. Validity was evaluated using Pearson's product-moment correlation coefficients and resulted in 0.99. Intra- and interobserver ICC were, respectively, equal to 0.97 and 0.76. Authors concluded that "a smartphone with the iHandy Level app can measure ankle dorsiflexion with high reliability as well as construct and criterion validity" [32]. This work validates the use of a specific Smartphone application for the active angle measurement of ankle dorsiflexion.

The second study [33] evaluated a Smartphone application, Tiltmeter (IntegraSoftHN, Carlos E. Hernández Pérez) on iPhone 4 and 4S (Apple, Cupertino, USA) during the weight bearing lunge test. A digital inclinometer (Laser Depot, Adelaide, Australia) was used as a gold standard. Twenty healthy participants were recruited and measurements were performed by two podiatrists. Examiners helped participants to slowly move the right foot back until they were able to hold the lunge position with the heel on the floor and with the right foot straight and perpendicular to the wall. Intrarater reliability was determined using $ICC_{(2,1)}$ and 95% CI. Interrater reliability was determined using $ICC_{(2,2)}$ and 95% CI. Validity between both devices was explored using Bland and Altman plots and ICC and resulted in a mean value of 0.83. Intra- and interobserver ICC were ranged between 0.81 to 0.85 and 0.80 to 0.96. Authors concluded that "the use of the TiltMeter app on the iPhone is a reliable measure of ankle range of motion in healthy adults" [33]. This scientific work validates the use of a commercial standard Smartphone application for the active angle measurement for weight bearing lunge test.

3. Using Smartphone for Proprioception Rehabilitation in Autonomous Way

All the previously presented studies assess the use of the Smartphone for functional and proprioceptive abilities assessment. Most of them only focus on joint angle measurements through ROM. They all conclude that Smartphone applications, which are sensor-based or camera-based, are reliable and valid for measuring angle compared to some gold standard as goniometer, bubble inclinometer, 3D navigation system, or even scoliometer for assessing JPS. However, some limitations are pointed out by authors. In cervical range of motion, both studies concluded that rotation evaluations are not reliable due to magnetic field interference. Gimbal lock effect may also decrease reliability for JPS if its effect is not taken into account in the measurement protocol. ISB recommendation proposes, for each joint, a standard for the local axis system in each articulating segment or bone and thus can bring solutions to avoid Gimbal lock effect in protocols [49, 50]. In their recent review of Smartphone goniometric tools, Milani et al. [10] concluded that there are no validation studies focusing on Smartphone application in dynamic conditions. We fully agree with this conclusion. We further state that while the Smartphone is now validated as a reliable measurement tool and can be used in clinical practice, there are no studies which use the power of the Smartphone as both measurement tool and a standalone tool for autonomous rehabilitation at home. Ubiquitous, home health or telehealth and telecare services are well explored [51–55] but remain, for the moment, at the proof-of-concept state. Algar and Valdes evaluated in their study the use of Smartphone applications as hand therapy interventions [56]. They explained how Smartphone applications could bring solutions to clinician for rehabilitation at home and how it can improve patient compliance. A first example is given for treatment of trapeziometacarpal arthrosis with two Smartphone applications which require the use of both palmar abduction and the unconscious activation of thumb muscles. Exercises including these movements are essential to increase range of motion and grip strength and to decrease pain. The second example is for treatment following distal radius fracture. Smartphone applications can provide wrist proprioceptive and joint sense exercise, whose therapeutic roles are validated for rehabilitation after wrist injuries. It now remains to assess the benefits of these applications in clinical studies involving targeted populations on rehabilitations exercises at home. Using the Smartphone for home rehabilitation exercises just started since these tools are now available to the largest number in developed countries. It was firstly studied for cardiac disease [57, 58], pulmonary rehabilitation [59], or prevention of ankles sprains [60, 61].

In their study in cardiac rehabilitation, Varnfield et al. have compared the use of a Smartphone for cardiac rehabilitation against traditional home-based rehabilitation [57]. One hundred and twenty patients with postmyocardial infarction were recruited during six months and randomly separated into two distinct groups. Uptake, adherence, and completion were evaluated. Significance for relative risk was calculated using two-sided Fisher's exact test. *chi*-square test was used for categorical variables, two-sample t-test was used for continuous variables, and the Wilcoxon rank-sum test was used for skewed variables. A linear mixed model regression was used to compare longitudinal changes across baseline and a preliminary multivariate analysis was used to analyze the association between nine selected baseline characteristics and outcomes. The Smartphone-based program was used with the aim of delivering exercises monitoring, motivational and educational materials via Short Text Messages (SMS) and video, and a health diary. Authors concluded that "this smartphone-based home care CR program improved post-MI CR uptake, adherence and completion." [57], validating the use and clinical effectiveness of a Smartphone application for home care cardiac rehabilitation.

A second study, from Layton et al., aimed to determine the feasibility and the acceptability of a Smartphone-based application to monitor outpatient discharge instruction compliance in cardiac disease [58]. Sixteen patients were recruited. Smartphone was used to daily monitor medication compliance, physical activity, follow-up care, symptoms, and reading of education material. Findings suggest that stable patients used the application more than unstable patients. Acceptability was low and varied greatly but it is similar to other studies. Authors concluded that this study "demonstrated that usage alone may be a useful tool to highlight patients in need of closer monitoring" [58]. This scientific work allows validating the feasibility and acceptability of a Smartphone application for monitoring outpatient.

For prevention of ankles sprains, study from Vriend et al. led to the same acceptability results [61]. These authors have developed a Smartphone application providing an eight-week neuromuscular training program with a set of six different exercises. It was evaluated using the Reach Effectiveness Adoption Implementation Maintenance Framework. Results showed a low compliance but the app reached only 2.6% of the projected targeted population.

For their study in pulmonary rehabilitation, Marshall et al. described a model of Smartphone application which can support remote patients with chronic obstructive pulmonary disease and give them an automatic feedback during exercises [59]. This application was not yet evaluated in full patient trial.

Thus, some of these studies highlight the fact that the acceptability varied greatly [58] and, for prevention, targeted efforts have to be made to ensure that a specific population can and will be willing to use the application [61]. However, these studies confirm the feasibility and a certain acceptability to use Smartphone application for monitoring and rehabilitation at home. Following these observations, a need exists for validation studies focused on autonomous rehabilitation at home using the Smartphone as a personal physiotherapist that can bring measurement and feedback to patients to improve the follow-up between medical sessions.

4. Future Work

Autonomous rehabilitation could be provided by a Smartphone-based system. This system is composed of inertial sensors to measure orientations, calculation units to analyze motor control abilities, visual, auditory, and somatosensory systems to provide biofeedback to the user, screen display and headphones to provide test and/or training exercises instructions, and wireless connection to transmit data. With this system, physiotherapist could provide to patient specific and personalized exercises to optimally improve proprioceptive functions. Various proprioceptive exercises are possible: active joint repositioning training, path-of-motion training, and so forth. It was proved that, for proprioception assessment, active movements give more information from muscle and joints receptors while fatigue should be avoided [41]. Along these lines, to assess, monitor, improve, and train proprioceptive function, we have developed a specific Smartphone application called "iProprio." "iProprio" functioning is based on the use of inertial sensors to measure active range of motion from different body part such as shoulder, elbow, or knee. The innovative part of the application is based on the fact that it proposes different active joint repositioning training with the help of different sensory feedback. All these exercises could be performed in autonomous way at home thanks to the Smartphone. The instructions can be automatically vocally or visually supplied. We are currently evaluating "iProprio" with targeted population in terms of effectiveness, efficiency, satisfaction, usability, and acceptance with a specific design model called TEMSED for "Technology, Ergonomics, Medicine, Society, Economics, and Deontology" [62].

5. Conclusions

In this paper, we have reported related works on clinical assessment that uses Smartphone as a joint angle measurement tool to assess proprioceptive abilities. It is mainly used for assess joint position sense and range of motion. This state of the art highlights that Smartphone applications have proved their reliability and validity for clinical uses. At this point, although their usefulness is underlined in some studies conclusions, there are no studies that have evaluated the use of a Smartphone in autonomy during home rehabilitation through exercise therapy to enhance proprioception.

Disclosure

Nicolas Vuillerme and Anthony Fleury work at Jean-Raoul Scherrer International Associate Laboratory (LAI) which is shared by two universities, namely, University Geneva, Geneva, Switzetland, and University Grenoble-Alpes, France.

Conflict of Interests

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

Authors' Contribution

Anthony Fleury, Bruno Diot, Céline Franco, and Nicolas Vuillerme contributed equally to this work.

Acknowledgments

This work was supported in part by funding by IDS company, the French national program "Programme d'Investissements d'Avenir IRT Nanoelec" ANR-10-AIRT-05, Ph.D. scholarship of Mines Douai, and Institut Universitaire de France. The authors would like to thank anonymous reviewers for helpful comments and suggestions.

References

[1] K. R. Grob, M. S. Kuster, S. A. Higgins, D. G. Lloyd, and H. Yata, "Lack of correlation between different measurements of proprioception in the knee," *The Journal of Bone & Joint Surgery—British Volume*, vol. 84, no. 4, pp. 614–618, 2002.

[2] F. Ribeiro and J. Oliveira, "Factors influencing proprioception: what do they reveal?" in *Biomechanics in Applications*, V. Klika, Ed., chapter 14, InTech, Rijeka, Croatia, 2011.

[3] L. Hay, C. Bard, M. Fleury, and N. Teasdale, "Availability of visual and proprioceptive afferent messages and postural control in elderly adults," *Experimental Brain Research*, vol. 108, no. 1, pp. 129–139, 1996.

[4] G. C. Gauchard, C. Jeandel, A. Tessier, and P. P. Perrin, "Beneficial effect of proprioceptive physical activities on balance control in elderly human subjects," *Neuroscience Letters*, vol. 273, no. 2, pp. 81–84, 1999.

[5] C. M. L. Hughes, P. Tommasino, A. Budhota, and D. Campolo, "Upper extremity proprioception in healthy aging and stroke populations, and the effects of therapist- and robot-based rehabilitation therapies on proprioceptive function," *Frontiers in Human Neuroscience*, vol. 9, article 120, 2015.

[6] S. Hillier, M. Immink, and D. Thewlis, "Assessing proprioception—a systematic review of possibilities," *Neurorehabilitation and Neural Repair*, 2015.

[7] E. Ozdalga, A. Ozdalga, and N. Ahuja, "The smartphone in medicine: a review of current and potential use among physicians and students," *Journal of Medical Internet Research*, vol. 14, no. 5, article e128, 2012.

[8] M. N. K. Boulos, S. Wheeler, C. Tavares, and R. Jones, "How smartphones are changing the face of mobile and participatory healthcare: an overview, with example from eCAALYX," *BioMedical Engineering Online*, vol. 10, no. 1, article 24, 2011.

[9] R. Zhu and Z. Zhou, "A real-time articulated human motion tracking using tri-axis inertial/magnetic sensors package," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 12, no. 2, pp. 295–302, 2004.

[10] P. Milani, C. A. Coccetta, A. Rabini, T. Sciarra, G. Massazza, and G. Ferriero, "Mobile smartphone applications for body position measurement in rehabilitation: a review of goniometric tools," *PM&R*, vol. 6, no. 11, pp. 1038–1043, 2013.

[11] Y. Tousignant-Laflamme, N. Boutin, A. M. Dion, and C.-A. Vallée, "Reliability and criterion validity of two applications of the iPhone to measure cervical range of motion in healthy participants," *Journal of NeuroEngineering and Rehabilitation*, vol. 10, no. 1, article 69, 2013.

[12] J. Quek, S. G. Brauer, J. Treleaven, Y.-H. Pua, B. Mentiplay, and R. A. Clark, "Validity and intra-rater reliability of an Android phone application to measure cervical range-of-motion," *Measurements*, vol. 11, article 65, 2014.

[13] S. H. Shin, D. H. Ro, O.-S. Lee, J. H. Oh, and S. H. Kim, "Within-day reliability of shoulder range of motion measurement with a smartphone," *Manual Therapy*, vol. 17, no. 4, pp. 298–304, 2012.

[14] B. C. Werner, R. E. Holzgrefe, J. W. Griffin et al., "Validation of an innovative method of shoulder range-of-motion measurement using a smartphone clinometer application," *Journal of Shoulder and Elbow Surgery*, vol. 23, no. 11, pp. e275–e282, 2014.

[15] K. Mitchell, S. B. Gutierrez, S. Sutton, S. Morton, and A. Morgenthaler, "Reliability and validity of goniometric iPhone applications for the assessment of active shoulder external rotation," *Physiotherapy Theory and Practice*, vol. 30, no. 7, pp. 521–525, 2014.

[16] L. Oïhénart, C. Duc, and K. Aminian, "iShould: functional evaluation of the shoulder using a smartphone," *Gait & Posture*, vol. 36, supplement 1, pp. S61–S62, 2012.

[17] L. B. Johnson, S. Sumner, T. Duong et al., "Validity and reliability of smartphone magnetometer-based goniometer evaluation of shoulder abduction—a pilot study," *Manual Therapy*, 2015.

[18] G. Ferriero, F. Sartorio, C. Foti, D. Primavera, E. Brigatti, and S. Vercelli, "Reliability of a new application for smartphones (DrGoniometer) for elbow angle measurement," *PM&R*, vol. 3, no. 12, pp. 1153–1154, 2011.

[19] M. Ockendon and R. E. Gilbert, "Validation of a novel smartphone accelerometer-based knee goniometer," *The Journal of Knee Surgery*, vol. 25, no. 4, pp. 341–345, 2012.

[20] K. Hambly, R. Sibley, and M. Ockendon, "Level of agreement between a novel smartphone application and a long arm goniometer for the assessment of maximum active knee flexion by an inexperienced tester," *International Journal of Physiotherapy and Rehabilitation*, vol. 2, pp. 1–14, 2012.

[21] A. Jones, R. Sealey, M. Crowe, and S. Gordon, "Concurrent validity and reliability of the simple goniometer iPhone app compared with the Universal Goniometer," *Physiotherapy Theory and Practice*, vol. 30, no. 7, pp. 512–516, 2014.

[22] S. Milanese, S. Gordon, P. Buettner et al., "Reliability and concurrent validity of knee angle measurement: smart phone app versus universal goniometer used by experienced and novice clinicians," *Manual Therapy*, vol. 19, no. 6, pp. 569–574, 2014.

[23] S. Rwakabayiza, L. C. D. Pereira, E. Lécureux, and B. Jolles-Haeberli, "Mesurer l'amplitude articulaire du genou: goniomètre universel ou smartphone?" *Orthopédie*, vol. 411, no. 44, pp. 2372–2375, 2013.

[24] A.-V. Bruyneel and F. Bridon, "Inclinométrie du genou: comparaison de la reproductibilité d'un outil mécanique et d'une application sur smartphone," *Kinésithérapie la Revue*, vol. 15, no. 158, pp. 74–79, 2015.

[25] J.-Y. Jenny, "Measurement of the knee flexion angle with a smartphone-application is precise and accurate," *Journal of Arthroplasty*, vol. 28, no. 5, pp. 784–787, 2013.

[26] G. Ferriero, S. Vercelli, F. Sartorio et al., "Reliability of a smartphone-based goniometer for knee joint goniometry," *International Journal of Rehabilitation Research*, vol. 36, no. 2, pp. 146–151, 2013.

[27] J.-Y. Jenny, A. Bureggah, and Y. Diesinger, "Measurement of the knee flexion angle with smartphone applications: which technology is better?" *Knee Surgery, Sports Traumatology, Arthroscopy*, 2015.

[28] F. Andrea, V. Luigi, M. Daniele et al., "Smartphone versus knee ligament arthrometer when size does not matter," *International Orthopaedics*, vol. 38, no. 10, pp. 2197–2199, 2014.

[29] T.-L. Yoon, K.-M. Park, S.-A. Choi, J.-H. Lee, H.-J. Jeong, and H.-S. Cynn, "A comparison of the reliability of the trochanteric prominence angle test and the alternative method in healthy subjects," *Manual Therapy*, vol. 19, no. 2, pp. 97–101, 2014.

[30] F. M. Peters, R. Greeff, N. Goldstein, and C. T. Frey, "Improving acetabular cup orientation in total hip arthroplasty by using

smartphone technology," *Journal of Arthroplasty*, vol. 27, no. 7, pp. 1324–1330, 2012.

[31] P. C. Charlton, B. F. Mentiplay, Y.-H. Pua, and R. A. Clark, "Reliability and concurrent validity of a Smartphone, bubble inclinometer and motion analysis system for measurement of hip joint range of motion," *Journal of Science and Medicine in Sport*, vol. 18, no. 3, pp. 262–267, 2015.

[32] S. L. Vohralik, A. R. Bowen, J. Burns, C. E. Hiller, and E. J. Nightingale, "Reliability and validity of a smartphone app to measure joint range," *American Journal of Physical Medicine & Rehabilitation*, vol. 94, no. 4, pp. 325–330, 2015.

[33] C. M. Williams, A. J. Caserta, and T. P. Haines, "The TiltMeter app is a novel and accurate measurement tool for the weight bearing lunge test," *Journal of Science and Medicine in Sport*, vol. 16, no. 5, pp. 392–395, 2013.

[34] M. J. Kolber, M. Pizzini, A. Robinson, D. Yanez, and W. J. Hanney, "The reliability and concurrent validity of measurements used to quantify lumbar spine mobility: an analysis of an iPhone application and gravity based inclinometry," *International Journal of Sports Physical Therapy*, vol. 8, no. 2, p. 129, 2013.

[35] M. T. Izatt, G. R. Bateman, and C. J. Adam, "Evaluation of the iPhone with an acrylic sleeve versus the Scoliometer for rib hump measurement in scoliosis," *Scoliosis*, vol. 7, no. 1, article 14, 2012.

[36] O. I. Franko, C. Bray, and P. O. Newton, "Validation of a scoliometer smartphone app to assess scoliosis," *Journal of Pediatric Orthopaedics*, vol. 32, no. 8, pp. e72–e75, 2012.

[37] F. Balg, M. Juteau, C. Theoret, A. Svotelis, and G. Grenier, "Validity and reliability of the iPhone to measure rib hump in scoliosis," *Journal of Pediatric Orthopaedics*, vol. 34, no. 8, pp. 774–779, 2014.

[38] J. Qiao, L. Xu, Z. Zhu et al., "Inter- and intraobserver reliability assessment of the axial trunk rotation: manual versus smartphone-aided measurement tools," *BMC Musculoskeletal Disorders*, vol. 15, no. 1, article 343, 2014.

[39] U. Röijezon, N. C. Clark, and J. Treleaven, "Proprioception in musculoskeletal rehabilitation. Part 1: basic science and principles of assessment and clinical interventions," *Manual Therapy*, vol. 20, no. 3, pp. 368–377, 2015.

[40] B. L. Riemann, J. B. Myers, and S. M. Lephart, "Sensorimotor system measurement techniques," *Journal of Athletic Training*, vol. 37, no. 1, pp. 85–98, 2002.

[41] N. Strimpakos, "The assessment of the cervical spine. Part 1: range of motion and proprioception," *Journal of Bodywork and Movement Therapies*, vol. 15, no. 1, pp. 114–124, 2011.

[42] D. J. Goble, "Proprioceptive acuity assessment via joint position matching: from basic science to general practice," *Physical Therapy*, vol. 90, no. 8, pp. 1176–1184, 2010.

[43] R. L. Gajdosik and R. W. Bohannon, "Clinical measurement of range of motion. Review of goniometry emphasizing reliability and validity," *Physical Therapy*, vol. 67, no. 12, pp. 1867–1872, 1987.

[44] J. Treleaven, G. Jull, and M. Sterling, "Dizziness and unsteadiness following whiplash injury: characteristic features and relationship with cervical joint position error," *Journal of Rehabilitation Medicine*, vol. 35, no. 1, pp. 36–43, 2003.

[45] J. K. Loudon, M. Ruhl, and E. Field, "Ability to reproduce head position after whiplash injury," *Spine*, vol. 22, no. 8, pp. 865–868, 1997.

[46] M. Revel, C. Andre-Deshays, and M. Minguet, "Cervicocephalic kinesthetic sensibility in patients with cervical pain," *Archives of Physical Medicine and Rehabilitation*, vol. 72, no. 5, pp. 288–291, 1991.

[47] M. Balke, D. Liem, N. Dedy et al., "The laser-pointer assisted angle reproduction test for evaluation of proprioceptive shoulder function in patients with instability," *Archives of Orthopaedic and Trauma Surgery*, vol. 131, no. 8, pp. 1077–1084, 2011.

[48] P. Yan, G. Kurillo, R. Bajcsy et al., "mHealth application for upper extremity range of motion and reachable workspace," in *Medicine Meets Virtual Reality 20*, vol. 184 of *Studies in Health Technology and Informatics*, pp. 478–480, IOS Press, Amsterdam, The Netherlands, 2012.

[49] G. Wu, S. Siegler, P. Allard et al., "ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine," *Journal of Biomechanics*, vol. 35, no. 4, pp. 543–548, 2002.

[50] G. Wu, F. C. T. Van Der Helm, H. E. J. Veeger et al., "ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion. Part II. Shoulder, elbow, wrist and hand," *Journal of Biomechanics*, vol. 38, no. 5, pp. 981–992, 2005.

[51] M. Memon, S. R. Wagner, C. F. Pedersen, F. H. Aysha Beevi, and F. O. Hansen, "Ambient assisted living healthcare frameworks, platforms, standards, and quality attributes," *Sensors*, vol. 14, no. 3, pp. 4312–4341, 2014.

[52] E.-Y. Jung, J.-H. Kim, K.-Y. Chung, and D. K. Park, "Home health gateway based healthcare services through U-health platform," *Wireless Personal Communications*, vol. 73, no. 2, pp. 207–218, 2013.

[53] G. Lamprinakos, S. Asanin, T. Broden et al., "An integrated remote monitoring platform towards telehealth and telecare services interoperability," *Information Sciences*, vol. 308, pp. 23–37, 2015.

[54] W. Maass and U. Varshney, "Design and evaluation of Ubiquitous Information Systems and use in healthcare," *Decision Support Systems*, vol. 54, no. 1, pp. 597–609, 2012.

[55] D. Kotz, S. Avancha, and A. Baxi, "A privacy framework for mobile health and home-care systems," in *Proceedings of the 1st ACM Workshop on Security and Privacy in Medical and Home-Care Systems (SPIMACS '09)*, pp. 1–12, ACM, Chicago, Ill, USA, November 2009.

[56] L. Algar and K. Valdes, "Using smartphone applications as hand therapy interventions," *Journal of Hand Therapy*, vol. 27, no. 3, pp. 254–257, 2014.

[57] M. Varnfield, M. Karunanithi, C.-K. Lee et al., "Smartphone-based home care model improved use of cardiac rehabilitation in postmyocardial infarction patients: results from a randomised controlled trial," *Heart*, vol. 100, no. 22, pp. 1770–1779, 2014.

[58] A. M. Layton, J. Whitworth, J. Peacock, M. N. Bartels, P. A. Jellen, and B. M. Thomashow, "Feasibility and acceptability of utilizing a smartphone based application to monitor outpatient discharge instruction compliance in cardiac disease patients around discharge from hospitalization," *International Journal of Telemedicine and Applications*, vol. 2014, Article ID 415868, 10 pages, 2014.

[59] A. Marshall, O. Medvedev, and A. Antonov, "Use of a smartphone for improved self-management of pulmonary rehabilitation," *International Journal of Telemedicine and Applications*, vol. 2008, Article ID 753064, 5 pages, 2008.

[60] M. Van Reijen, I. I. Vriend, V. Zuidema, W. van Mechelen, and E. A. Verhagen, "The implementation effectiveness of the 'Strengthen your ankle' smartphone application for the prevention of ankle sprains: design of a randomized controlled trial," *BMC Musculoskeletal Disorders*, vol. 15, no. 1, article 2, 2014.

[61] I. Vriend, I. Coehoorn, and E. Verhagen, "Implementation of an app-based neuromuscular training programme to prevent ankle sprains: a process evaluation using the RE-AIM framework," *British Journal of Sports Medicine*, vol. 49, no. 7, pp. 484–488, 2015.

[62] V. Rialle, N. Vuillerme, and A. Franco, "Outline of a general framework for assessing e-health and gerontechnology applications: axiological and diachronic dimensions," *Gerontechnology*, vol. 9, no. 2, article 245, 2010.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 902896, 11 pages
http://dx.doi.org/10.1155/2015/902896

Clinical Study

Effect of Workplace- versus Home-Based Physical Exercise on Muscle Response to Sudden Trunk Perturbation among Healthcare Workers: A Cluster Randomized Controlled Trial

Markus D. Jakobsen,[1,2] Emil Sundstrup,[1,2] Mikkel Brandt,[1,3] Kenneth Jay,[1,2,4] Per Aagaard,[2] and Lars L. Andersen[1,3]

[1] *National Research Centre for the Working Environment, Lersø Parkallé 105, 2100 Copenhagen, Denmark*
[2] *Department of Sports Science and Clinical Biomechanics, SDU Muscle Research Cluster (SMRC), University of Southern Denmark, Campusvej 55, 5230 Odense, Denmark*
[3] *Physical Activity and Human Performance Group, SMI, Department of Health Science and Technology, Aalborg University, Fredrik Bajers Vej 7, 9220 Aalborg, Denmark*
[4] *Electronics and Computer Science, Faculty of Physical and Applied Sciences, University of Southampton, University Road, Southampton SO17 1BJ, UK*

Correspondence should be addressed to Markus D. Jakobsen; markusdue@gmail.com

Received 24 April 2015; Accepted 5 July 2015

Academic Editor: Luis-Millán González

Objectives. The present study investigates the effect of workplace- versus home-based physical exercise on muscle reflex response to sudden trunk perturbation among healthcare workers. *Methods*. Two hundred female healthcare workers (age: 42 [SD 11], BMI: 24 [SD 4], and pain intensity: 3.1 [SD 2.2] on a scale of 0–10) from 18 departments at three hospitals were randomized at the cluster level to 10 weeks of (1) workplace physical exercise (WORK) performed in groups during working hours for 5×10 minutes per week and up to 5 group-based coaching sessions on motivation for regular physical exercise, or (2) home-based physical exercise (HOME) performed during leisure time for 5×10 minutes per week. Mechanical and neuromuscular (EMG) response to randomly assigned unloading and loading trunk perturbations and questions of fear avoidance were assessed at baseline and 10-week follow-up. *Results*. No *group by time* interaction for the mechanical trunk response and EMG latency time was seen following the ten weeks ($P = 0.17$–0.75). However, both groups demonstrated within-group changes ($P < 0.05$) in stopping time during the loading and unloading perturbation and in stopping distance during the loading perturbation. Furthermore, EMG preactivation of the erector spinae and fear avoidance were reduced more following WORK than HOME (95% CI −2.7–−0.7 ($P < 0.05$) and −0.14 (−0.30 to 0.02) ($P = 0.09$)), respectively. WORK and HOME performed 2.2 (SD: 1.1) and 1.0 (SD: 1.2) training sessions per week, respectively. *Conclusions*. Although training adherence was higher following WORK compared to HOME this additional training volume did not lead to significant between-group differences in the responses to sudden trunk perturbations. However, WORK led to reduced fear avoidance and reduced muscle preactivity prior to the perturbation onset, compared with HOME. This trial is registered with Clinicaltrials.gov (NCT01921764).

1. Introduction

Low back pain (LBP) is one of the most prevalent and costly work related health problems which affects millions of workers and workplaces worldwide [1–4]. Healthcare work is associated with an elevated risk of back pain and musculoskeletal injuries among women [5, 6]. Particularly the frequent often nonanticipated and high loadings of the spine while twisting and bending the back during patient handling [7–10] increase the risk for experiencing acute injuries and/or developing LBP among healthcare workers [11, 12].

Previous literature suggests that LBP alters muscle recruitment patterns. For example, numerous studies have shown that LBP is associated with delayed muscle reflex responses to sudden trunk loadings compared with healthy controls [13–15]. Accordingly, people with current LBP respond differently to sudden trunk loading than people without a history of LBP. In addition, reports of elevated preactivation levels as an attempt to stabilize the trunk prior to perturbation—maybe as a result of fear avoidance of sudden movement—have been observed in subjects with LBP [16]. However, no previous studies have investigated whether reductions in LBP are accompanied by improved (faster) muscle response to sudden trunk loadings. Nevertheless, a faster muscle reflex response implies an earlier stabilization of the spine [17] which may protect against overload injury. In support of this, Cholewicki et al. demonstrated, in a prospective study, that healthy subjects with a delayed lower back reflex response during sudden trunk perturbation had an increased risk of future low back injury [18]. Thus, improving trunk reflex response through, that is, exercise intervention may protect against future injury to the spine and truncus region.

Only a few studies have investigated the effect of training on the neuromechanical response to sudden trunk loading. In a 9-week longitudinal study Pedersen et al. trained healthcare workers without a previous history of LBP to react to a variety of sudden trunk loadings [19]. The training resulted in reduced trunk displacement and stopping time during unexpected trunk perturbations but did not alter reflex latencies measured with surface electromyography (EMG) in the erector spinae muscles. However, the reduction in trunk stopping time was accompanied by an increase in the neuromuscular (EMG) activity just prior to the instant where the perturbation was effectively stopped. Similar observations of increased EMG amplitudes have been demonstrated in the erector spinae muscle after 10 weeks of stabilizing exercise programs in patients with subacute recurrent LBP without any changes in reflex latencies [20]. Pedersen and coworkers showed, in a recent study, that 16 weeks of recreational soccer training significantly reduced trunk stopping time and stopping distance in healthy women compared with subjects who underwent continuous running exercise [21]. These authors concluded that the high number of sudden loadings (tackles, accelerations, and decelerations) exerted on the trunk during soccer training was responsible for the observed changes in stopping time and distance [21]. Accordingly, it was suggested that trunk exercise programs should not only focus on training trunk muscle strength and flexibility but also incorporate exercises with unexpected sudden loadings. In these studies the average training exposure was 2 times 45–60 min per week performed either before or after working hours which may be difficult and expensive in terms of working hours spent to incorporate as a part of the daily routine of a healthcare worker. Therefore, it remains to be investigated whether short-term physical exercise performed without a specific focus on unexpected trunk reactions, either at the workplace during working hours or at home, can improve trunk muscle response to sudden unexpected perturbations.

The present study investigates the effect of workplace- versus home-based physical exercise on muscle response to sudden trunk perturbation among healthcare workers.

2. Methods and Analysis

2.1. Study Design. This two-armed parallel-group, single-blinded, cluster randomized controlled trial with allocation concealment recruited female healthcare workers from three hospitals (18 departments) situated in Copenhagen, Denmark, was conducted from August 2013 to January 2014. To increase adherence and avoid contamination between interventions we chose to cluster-randomize the participants at the department level. The participants were allocated to a 10-week intervention period and randomly assigned to receive either workplace or home-based physical exercise. To ensure that the study aim, hypothesis, and primary outcome parameters were predefined the study was approved by The Danish National Ethics Committee on Biomedical Research (Ethical committee of Frederiksberg and Copenhagen; H-3-2010-062) and registered in ClinicalTrials.gov (NCT01921764) prior to enrolment of participants. The present study followed the CONSORT checklist to ensure transparent and standardized reporting of the trial. All experimental conditions conformed to The Declaration of Helsinki. Details on the study protocol and primary outcome variables (change in average muscle pain intensity of the low back, neck, and shoulder) have been published elsewhere [22, 23].

2.2. Recruitment and Randomization. The recruitment of participants was two-phased and consisted of a short screening questionnaire conducted in June 2013, followed by a baseline clinical examination and questionnaire performed in Aug-Sept 2013.

Initially, a screening questionnaire was administered to 490 healthcare workers (aged 18–67 years) from three Danish hospitals situated in Copenhagen in June 2013. Subsequently, in August and September 2013, a total of 207 female healthcare workers participated in the baseline clinical examination. Exclusion criteria were pregnancy and cardiovascular and life-threatening disease. The overall flow of participant enrolment, test, and EMG measurement is depicted in Figure 1 and has been described in detail elsewhere [22].

On the basis of the questionnaire we randomly allocated the 18 departments (200 participants), using a computer-generated random numbers table, to receive either physical exercise at the workplace or at home. The participants at each department and their management were informed by e-mail about group allocation. All examiners were blinded to the group allocation at 10-week follow-up and participants were carefully instructed not to reveal their particular intervention group. Baseline characteristics of the two intervention groups are listed in Table 1.

2.3. Interventions. Participants in each cluster were allocated to a 10-week intervention period receiving either physical exercise at the hospital or physical exercise at home. Both groups were encouraged to perform physical exercises for

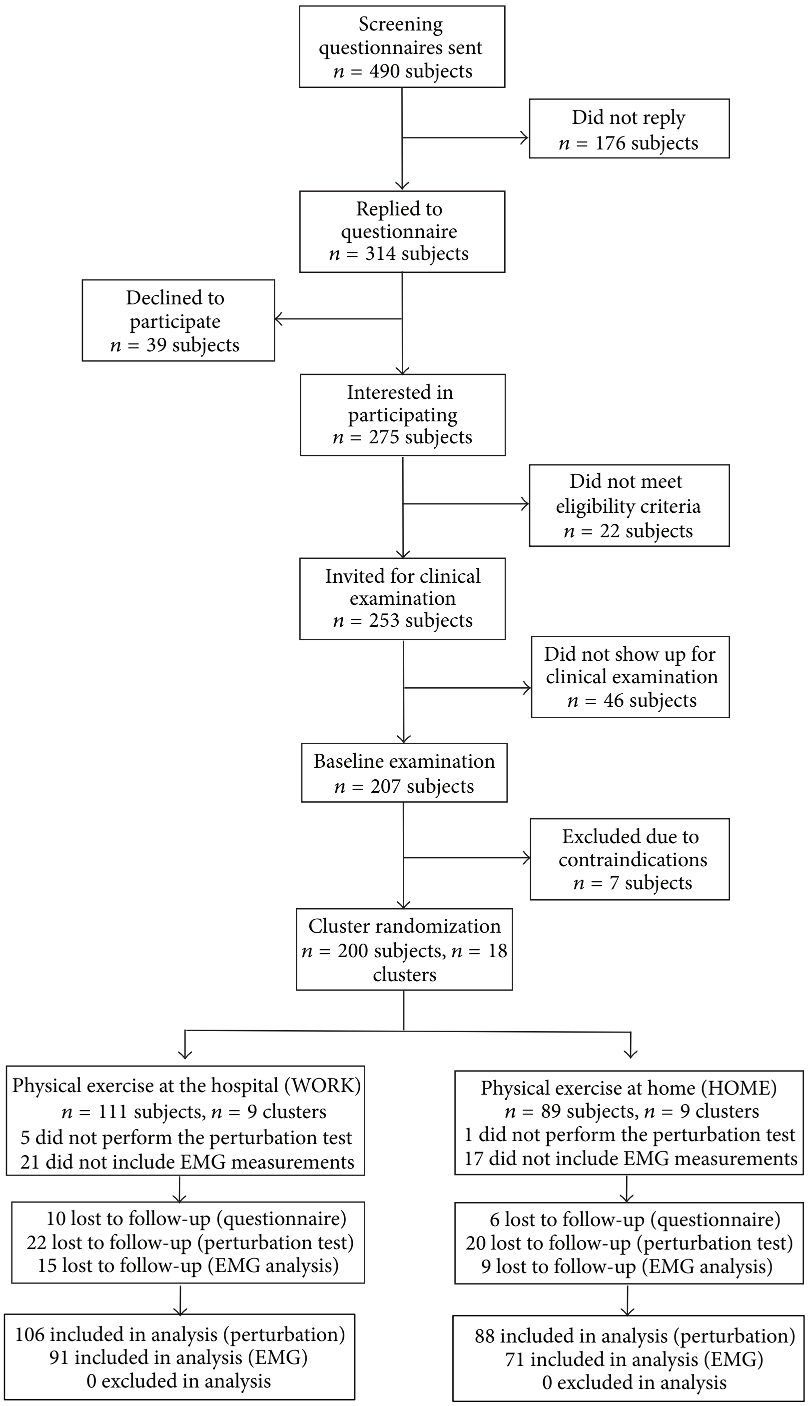


Figure 1: Participant recruitment flow-chart.

5×10 minutes a week. The specific intervention protocols are briefly summarized below, since they are described in detail elsewhere [23].

2.4. Workplace Physical Exercise (WORK). Subjects randomized to physical exercise at their workplace (WORK) (n = 111 subjects, n = 9 clusters) performed group-based and supervised high-intensity strength training using kettlebells, Swiss balls, and elastic bands (Thera-Band) exercises during working hours at the hospital (the exercises have been described in detail elsewhere [23]). All training sessions took place in designated rooms located at or close to the respective departments and all sessions were supervised by an experienced training instructor. The training program consisted of 10 separate exercises: kettlebell deadlifts, kettlebell swings, squeeze, lateral raises, golf swings, and woodchoppers using

Table 1: Baseline characteristics of the two intervention groups. Values are means (SD).

	WORK	HOME
N	111	89
Age (years)	40* (12)	44 (10)
Height (cm)	168.4 (6.2)	168.0 (7.2)
Weight (kg)	67.5 (12.1)	68.9 (12.2)
BMI (kg·m^{-2})	23.8 (3.8)	24.4 (4.0)
Average pain intensity in the low back, neck, and shoulders during the last week (scale 0–10)	3.0 (2.2)	3.1 (2.3)

* denotes difference between groups at baseline, $P < 0.05$. HOME: home-based physical exercise, WORK: work-based physical exercise.

elastic tubing, abdominal crunches, back extensions, and squats using a Swiss ball, and lunges using elastic tubing. For each training session the instructor chose 4–6 exercises that were performed as circuit training, that is, with quick transitions from one exercise to the next using no or minimal periods of rest. Training intensity (loads) progression was ensured by using progressively more resistant elastic bands and heavier kettlebells throughout the 10-week intervention period, as supervised by the instructors. WORK furthermore offered 5 group-based motivational coaching sessions (30–45 min with 5–12 participants in each session) during working hours.

2.5. Home-Based Physical Exercise (HOME). Participants randomized to home-based physical exercise (HOME) (n = 89 subjects, n = 9 clusters) performed physical exercises during leisure time at home. After the participants were informed about group allocation they received a bag with (1) training equipment (easy, medium, and hard elastic tubing) and (2) 3 posters that visually demonstrated the exercises that should be performed for the shoulder, abdominal, and back muscles and also contained recommendations for training progression [24–26].

2.6. Outcome Measures

2.6.1. Assessment of Sudden Perturbation. Measurements of the neuromechanical reaction to sudden unexpected trunk perturbations were performed by the same examiner before and after the intervention period. The method used for the perturbation has been described in detail previously [27]. In brief, perturbations were generated by means of a special loading/unloading device wired to a rigid bar attached to the upper part of the subject's trunk at level of insertion of the deltoid muscle (Figure 2). The subject was standing with the front facing the initial load (5.5 kg) and the pelvis fixated against a wooden plate to allow only movement of the trunk. Subsequently, an increase (load: 10.9 kg) or decrease (unload: 0.1 kg) in the load was randomly applied to the subject. The perturbation events were triggered by a computer with a random delay between 5 and 25 seconds unknown to the subject and the investigator. The test protocol consisted of 6 randomized unloaded or loaded perturbations (3 of each). To avoid anticipation of the direction of the 6th perturbation the subjects were instructed that a range of 6–8 perturbations would be performed. A minimum of 45 seconds of rest was ensured between each trial. The linear movement of the subject's trunk was recorded using a potentiometer attached to a reel that steered the wire. Prior to the perturbation event the subject was instructed to stand as relaxed as possible in the initial position and subsequently informed that within 25 seconds they would experience a moderate perturbation and instructed to immediately resist the perturbation and return to the initial position.

The analysis of the mechanical data obtained from the perturbations consisted of time elapsed from the onset of perturbation event until the movement of the trunk was reversed from perturbed direction (stopping time; time from event to maximum deviation of the movement curve from the initial position), distance moved from initial position to stop position (stopping distance) (Figure 3).

2.6.2. Electromyography (EMG) Recording and Analysis. EMG activity was recorded (1024 Hz) bilaterally from the left and right erector spinae (longissimus). A bipolar surface EMG configuration (Blue Sensor N-00-S/25, Ambu A/S, Ballerup, Denmark) and an interelectrode distance of 2 cm were used [28]. Before affixing the electrodes, the skin of the respective area was prepared with scrubbing gel (Acqua gel, Meditec, Parma, Italy) to effectively lower the impedance to less than 10 kΩ. The electrodes were placed bilaterally at 2-finger width lateral from the processus spinosi of L1 (http://www.seniam.org/). The electrodes were fixated with tape (Fixomull stretch) and connected through thin shielded cables to a datalogger (Nexus10, Mind Media, Netherlands) that was placed in a flexible belt to ensure unrestricted mobility during the test.

The EMG signals were digitally high-pass filtered using a 10 Hz cutoff frequency (4th order zero-lag Butterworth filter). To remove electrocardiographic (ECG) artefacts we band-pass filtered (10–25 Hz) the raw signal and subtracted this signal from the high-pass filtered signal. The filtered signal was subsequently rectified and smoothed using a moving root mean square (RMS; 10 ms time constant). The filtered and smoothed EMG signals were normalized with respect to maximal muscle activity obtained during maximal voluntary contractions of the back extensors (described later). EMG signals assessed during the MVCs were filtered with a Butterworth 4th order high-pass filter (10 Hz cutoff frequency) and smoothed by a moving root mean square (500 ms time constant) [29]. Data filtering and data analysis were performed using custom-made Matlab programs (MathWorks).

The following EMG parameters were calculated to determine onset latency and the relative muscular load: The EMG onset latency [EMG latency] was defined as the time between the loading of the trunk and the EMG onset. EMG onset was determined as the point where the filtered signal for more than 15 ms exceeded preactivation with 1.4 standard deviations of the preload EMG activity (measured

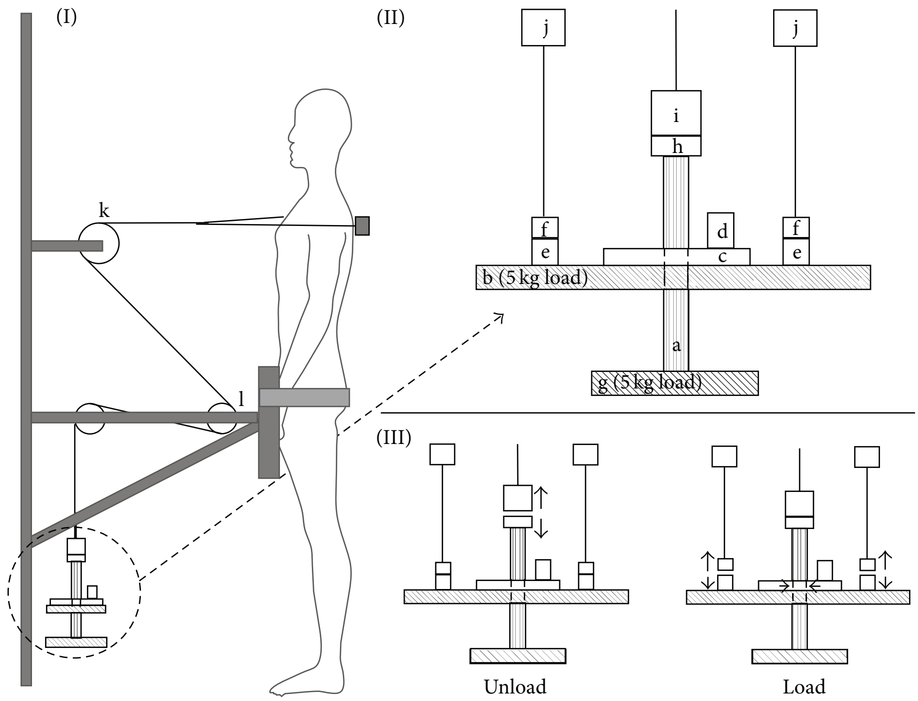


Figure 2: (I) Set-up for generating sudden perturbations to the upper part of the subject's trunk. The wire is fastened to a rigid bar fastened by a vest at the upper part of the trunk. The movement of the trunk is measured by a potentiometer mounted on a reel. (II) Details of the perturbation apparatus (right top: 90 degrees rotation) and standing position (left): (a) cylinder, (b) 5 kg load, (c) gripping device, (d) solenoid for activating gripping device, (e) holding magnets, (f) load-bearing construction, (g) 5 kg load, (h) anchor plate for magnet, (i) holding magnet, (j) bearing construction, (k) vertical adjustable reel with potentiometer, and (l) horizontal adjustable reel to adjust wire length to individual subject height. Generation of sudden unloading ((III) left): first the computer activates the magnet (i) and releases the load (b, h, and g) causing the weight of the load applied to the wire to suddenly decrease from 5.4 kg (a) to 0.1 kg (i). Generation of the sudden loading ((III) right): first the computer activates the solenoid (d) causing the gripping device (c) to fix the load (b) to the cylinder (a) and secondly deactivates the holding magnets (e). This releases the load (b, h, and g) causing the weight of the load applied to the wire to suddenly increase from 5.4 kg (a) to 10.9 kg (a–d).

1 s before loading) (this procedure was inspired by Radebold and coworkers [14]). Similarly, EMG shut-off latency [EMG shut-off] was defined as the point where the filtered signal was below 1.4 standard deviations of the preload activity (measured 1 second before loading) for more than 25 ms. Preactivation [EMG preactivation] was calculated as the mean EMG activity recorded within 1 second prior to the event (trunk loading/unloading). Postunload activation [EMG unload] was calculated as the mean EMG activity within the first 100 ms after the load was released. Finally, the maximal muscle activity obtained during the perturbation phase (from perturbation onset until perturbation was stopped) was identified as the peak-activation [EMG peak]. All EMG parameters where normalized to the maximal EMG activity measured during the maximal isometric back extension (MVC).

Synchronization of the EMG datalogger and the perturbation apparatus was ensured by a digital signal sent from the AD-converter (DAQCard-6036E, National Instruments) to the datalogger whenever a loading or unloading event occurred. A fixed 25 ms electromechanical time delay accounting for the triggering and loading/unloading was taken into account when calculating the EMG onset latencies. Figure 3 shows an example of the acquired data from a loading and unloading perturbation.

2.6.3. Back Extensor MVC Testing. Maximal voluntary isometric contraction strength (MVC) was obtained for the lower back extensor muscles using a custom-built dynamometer with a strain gauge load cell (KIS-2, 2 KN, Vishay Transducers Systems). During the MVC maneuver the subject was standing in an upright position wearing a vest with a steel rod horizontally placed at the upper part of the back, at the level of insertion of the deltoid muscle [22]. At the distal end of the rod a wire was horizontally connected to a strain-gauge dynamometer. The subject was facing the dynamometer with the pelvis positioned against a wooden plate (upper edge aligned with the subject's iliac crest) while performing a maximal back extensor contraction (3 s) on a cue given by the tester. The participants performed 3 MVC attempts, separated by a 30-second rest period, while instructed to apply force to the dynamometer as fast and forcefully as possible. The maximum EMG signal (peak filtered EMG amplitude) of the 3 MVCs was used for subsequent normalization.

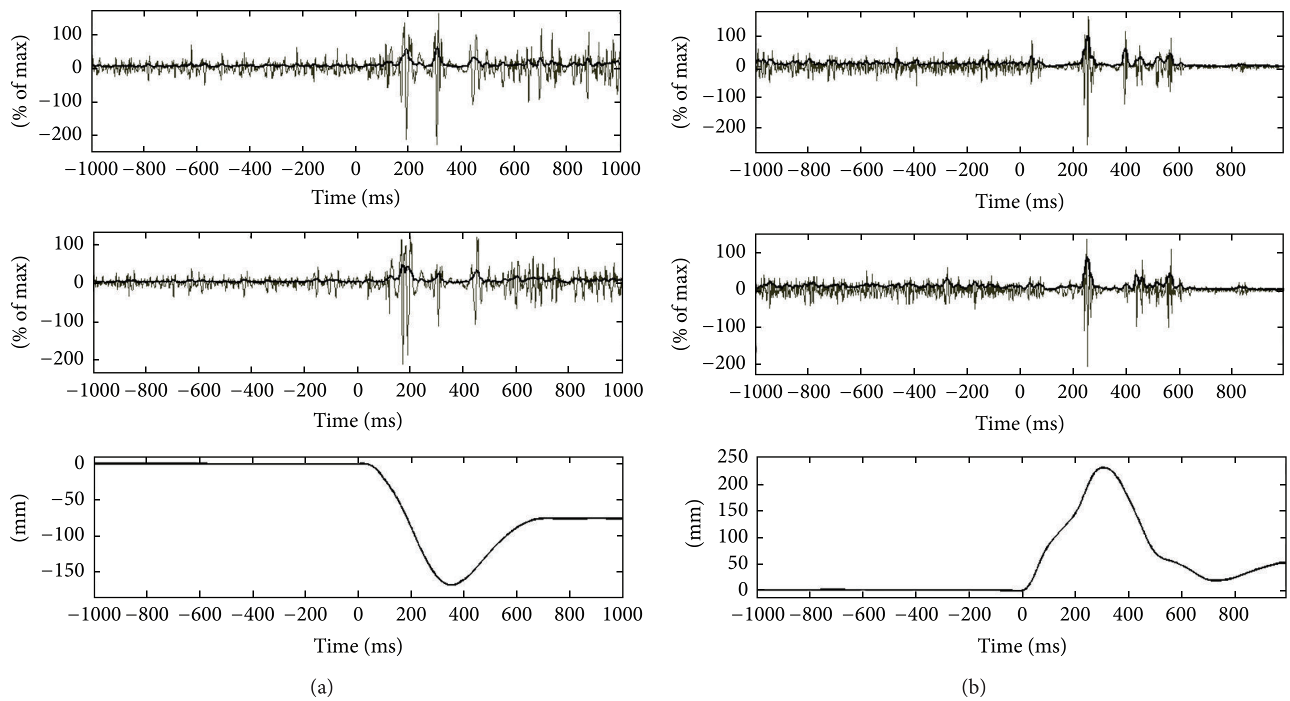


Figure 3: Potentionmeter and normalized back extensor EMG recordings of a (a) sudden trunk loading perturbation and (b) sudden unloading trunk perturbation. The perturbation was executed at time zero. The thin line is the raw signal and the thick line indicates the filtered EMG signal.

2.6.4. Fear Avoidance. Participants were asked to reply to the following question at baseline and at 10-week follow-up immediately before performing a maximal and rapid isometric back extension: "Do you think that this rapid and forceful back extension will induce back pain or increase your back pain?" Subjects replied on a scale using 4 levels of fear avoidance (FA): "Not at all" (FA = 0), "A little" (FA = 1), "Some" (FA = 2), and "A lot" (FA = 3).

2.7. Statistical Analysis. All statistical analyses were performed using the SAS statistical software for Windows (SAS Institute, Cary, NC). The change in mechanical parameters (stopping time and stopping distance) and EMG (onset latency, shut-off, preactivation, unload, and peak) was evaluated using a linear mixed model (Proc Mixed) with *group*, *time*, and *group by time* as independent variables. Participant nested within department was entered as random effect. Analyses were adjusted for age and baseline values. All statistical analyses were performed in accordance with the intention-to-treat principle, that is, using the mixed procedure which inherently accounts for missing values. An alpha level of 0.05 was accepted as statistically significant. Outcomes are reported as between-group least mean square differences and 95% confidence intervals at 10-week follow-up. For all EMG parameters (onset latency, shut-off, preactivation, unload, and peak) mean values of the left and right erector spinae were selected for the statistical analysis [19]. Effect sizes were calculated as Cohen's d [30] based on the observed within-group changes (within-group changes from baseline to follow-up divided by the pooled standard deviation at baseline). According to Cohen, effect sizes of 0.20 are considered small, 0.50 moderate, and 0.80 large [30].

3. Results

3.1. Study Population. Baseline characteristics of the study participants are shown in Table 1. Participant flow is shown in Figure 1 and further described in detail elsewhere [22].

As described previously, training adherence differed between intervention groups ($P < 0.001$). Out of the 5 offered training sessions per week subjects in WORK performed on average 2.2 (SD: 1.1) sessions per week whereas subjects in HOME performed 1.0 (SD: 1.2) session [22].

3.2. Trunk Perturbation and Fear Avoidance. A priori hypothesis testing showed no *group by time* interaction for stop time and stop distance during the loading and unloading perturbation ($P > 0.05$) (Table 2). However, significant group by time interaction was seen for muscle pre-activation and unload-activation during the unloading perturbation ($P < 0.05$). There were no differences in peak-activation and EMG onset latency during the loading perturbation ($P > 0.05$).

We observed a group by time interaction for fear avoidance ($P < 0.05$); that is, HOME and WORK changed differently over time. At 10-week follow-up, a tendency ($P = 0.09$) for a difference in fear avoidance beliefs (−0.14 [−0.30 to 0.02]) was seen between WORK and HOME. Fear avoidance decreased ($P < 0.001$) from 0.71 [0.61 to 0.81] to 0.49 [0.38 to 0.59] following WORK whereas fear avoidance was unaltered (0.62 [0.52 to 0.73] to 0.62 [0.50 to 0.74]) following HOME.

Table 2: Baseline, follow-up, and between-group differences at follow-up and within-group effect size for the mechanical and the EMG parameters of the loading and unloading perturbation and preactivation EMG measured immediately before each perturbation. Values are means (95% confidence interval). All values are adjusted for baseline value.

		WORK						HOME						Differences at follow-up		
		0 weeks		10 weeks				0 weeks		10 weeks						
		Mean	95% CI	Mean	95% CI	P	Effect size	Mean	95% CI	Mean	95% CI	P	Effect size	Mean	95% CI	P
Load	Stopping time (ms)	362	(358–366)	352	(347–357)	<0.01	−0.23	360	(355–365)	351	(346–356)	<0.01	−0.21	1.2	(−6–8.4)	0.84
	Stopping distance (mm)	178	(174–181)	167	(164–171)	<0.01	−0.32	178	(175–182)	166	(162–170)	<0.01	−0.38	1.5	(−0.7–3.9)	0.52
	EMG onset latency (ms)	94	(90–98)	88	(83–93)	0.09	−0.12	95	(90–99)	95	(89–101)	0.90	0.01	−6.8	(−14.6–0.9)	0.22
	EMG peak (% of max)	81	(75–86)	83	(77–89)	0.51	0.07	81	(75–87)	89	(82–97)	0.08	0.22	−5.9	(−15.7–3.8)	0.37
Unload	Stopping time (ms)	371	(363–379)	358	(350–367)	0.02	−0.12	374	(365–383)	351	(342–361)	<0.01	−0.23	7.0	(−6.1–20.2)	0.18
	Stopping distance (mm)	263	(256–270)	258	(250–266)	0.27	−0.06	265	(257–273)	258	(250–267)	0.16	−0.09	−0.1	(−1.2–1.1)	0.75
	EMG shut-off (ms)	97	(77–116.5)	72	(47.9–96.5)	0.13	−1.42	93	(70.6–115.7)	79	(51–107)	0.44	−0.83	−6.6	(−43.8–30.6)	0.68
	EMG peak (% of max)	50	(48–53)	46	(43–50)	0.04	−0.14	51	(47–54)	47	(43–52)	0.17	−0.11	−1.1	(−6.4–4.3)	0.79
	EMG unload (%)	6.6	(6-7)	6.1	(5–7)	0.25	−0.11	6.6	(6-7)	7.5	(7-8)	0.08	0.20	−1.4	(−2.6–−0.2)	0.04
Pre	EMG preactivation (% of max)	9.1	(8.5–9.6)	8.1	(7.5–8.8)	0.02	−0.16	9.0	(8.4–9.7)	9.8	(9.1–10.6)	0.07	0.13	−1.7	(−2.7–−0.7)	<0.01

HOME: home-based physical exercise, WORK: work-based physical exercise. Negative effect sizes denote a within-group decrease from baseline to follow-up.

4. Discussion

The present study demonstrated that ten weeks of low-frequency (2 sessions per week), short duration (10 min per session) physical exercise at the workplace is not superior to ten weeks of home-based exercise in reducing the muscle reflex response to sudden trunk perturbations among healthcare workers. However, compared to home-based exercise, greater reductions in muscle preactivation and fear avoidance were seen after physical exercise performed at the workplace.

Previous studies have demonstrated delayed muscle response to sudden trunk perturbations in patients with LBP compared with healthy controls [13–15]. Hence, the presence of pain may alter muscle recruitment. However, whether the impaired response to sudden perturbation is caused by the injury or pain itself or if the impaired neuromuscular function is a predisposing factor of pain remains unsolved. Damage to afferent receptors within the lumbar muscles and/or soft tissue of the spine could impair the magnitude and timing of somatosensory feedback from the trunk region to the CNS and in turn delay the reflex response. Pain may furthermore alter spinal neuronal excitability and thus negatively affect lumbar muscle activation [31, 32]. In addition, people with LBP may adopt an abnormal motor control strategy to avoid pain or to compensate for an injury or pain. Nevertheless, research is needed investigating whether reductions in musculoskeletal pain concurrently can reestablish concurrent impairments in motor control. In our study population, 10 weeks of workplace based exercise significantly reduced musculoskeletal back pain by 31% whereas smaller but nonsignificant changes (8%) were seen in response to 10 weeks of home-based exercise [22]. Despite these marked between-group differences in the effectiveness of pain reduction no between-group differences were observed for the mechanical (stopping time and distance) or neuromuscular (EMG onset latency) response to sudden trunk perturbation. Accordingly, it may be suggested that short-term reductions in perceived pain among subjects with mild to moderate pain intensity (average 3.1 SD 2.2 on a 0–10 scale) do not acutely alter motor control in response to sudden trunk perturbations, at least when achieved by means of low-frequency (twice per week) short-duration (10 min) exercise intervention.

Compared with ten weeks of home-based exercise, ten weeks of exercise at the workplace resulted in lowered preactivation of the erector spinae muscles immediately prior to trunk perturbation. There can be several explanations for this observation. Firstly, increased maximal muscle strength would result in a reduction in the relative magnitude of muscle loading for a given (fixed) perturbation load, hence potentially resulting in reduced levels of muscle preactivation. Accordingly, the 9% increase in maximal trunk extensor strength strength, shown in WORK [22], corresponds accurately to the 10% decrease in muscle preactivation observed in this group. Secondly, reports of increased muscle preactivation and elevated antagonist cocontraction levels as an attempt to stabilize the trunk and protect against injury and pain prior to perturbation have been observed in subjects with LBP [14, 16]. Thus, the reduction in pain in WORK may have decreased the subject's fear of injury and therefore contributed to the lower preactivation levels observed after the 10 weeks of workplace exercise. In support of this, a decrease in fear avoidance of rapid and forceful back movement was observed in WORK, suggesting that the subjects were less afraid of evoking lower pain or increasing their pain by performing fast and forceful muscle contractions. Whether this can also lead to more relaxed muscle activity patterns during the working day remains to be investigated. In female office workers with chronic neck and shoulder pain, when exposed to 10 weeks of strength training demonstrated a more relaxed muscle activity pattern throughout the working day [33].

Theoretically, lower levels of preactivation as seen following 10 weeks of workplace exercise may contribute to lower spinal stiffness thus potentially resulting in a larger perturbated trunk response. Hence, the lower levels of preactivation in WORK may have compromised the response to the sudden trunk perturbation. In addition, the HOME group demonstrated a tendency for an increase in preactivation which may have contributed to the faster stopping time and stopping distance during the loading and stopping time in the unloading perturbation observed in HOME. Consequently, these between-group differences in the change in preactivation should be taken into account when interpreting the present results on trunk perturbation.

Although neither of the present interventions were superior to the other in terms of stopping time and stopping distance several group by time differences emerged for the neuromuscular recruitment pattern during the perturbation task. During the unloading perturbation the WORK group demonstrated lower muscle activation in the first 100 ms after the instant of trunk unloading, compared with HOME. This suggests that the participants in WORK improved their ability to rapidly relax their trunk extensor muscles and/or to reduce the magnitude of cocontraction (the abdominal muscles are the primary muscles involved in stopping the trunk during the unloading) of the erector spinae muscles immediately after an unloading perturbation. Nevertheless, we did not see a group difference in how fast the muscle activity was reduced after the unloading as reflected by the nonsignificant change in EMG shut-off time. However, large negative within-group effect sizes for the EMG shut-off time indicate that a reduction was present in both groups, but large variations highlight the methodological challenges of evaluating this parameter using the present experimental setup. Nonetheless, delayed EMG shut-off time has been shown to increase the risk of future low back injury in healthy subjects [18]. Cholewicki et al. (2000), furthermore, proposed that measuring trunk stiffness and damping response to sudden loading of the trunk may provide an additional and more comprehensive understanding of the muscular patterns compared with the present calculations of mechanical displacement and timing [17]. Hodges and coworkers have moreover demonstrated that people with recurrent LBP have increased trunk stiffness and decreased damping [34]. It would therefore be interesting for future studies to investigate whether changes in low-back pain intensity, changes in recruitment pattern, and occurrence of low back injury are related with trunk stiffness and damping behavior.

Even though the present and previous studies report changes in EMG amplitude and/or stopping time and stopping distance in response to physical training [19–21, 35], training induced modifications in the EMG onset latency during sudden trunk loading remain to be demonstrated. Of notice, therefore, the participants in WORK demonstrated a tendency ($P = 0.09$) for a reduction in EMG onset latency which may have contributed to the faster stopping time that was observed during the sudden trunk loading perturbation following the ten weeks of workplace-based exercise.

We have previously shown, although using a slightly different test method, that 8 weeks of intensive kettlebell training (performed on average 2.1×20 min per week) significantly improved reaction to sudden unloading of the trunk, increased strength, and reduced musculoskeletal pain in laboratory technicians [36]. However, in the present study, only two out of the ten possible exercises in WORK involved the use of kettlebells. Consequently, the accumulated time the participants exercised with kettlebells might have been too little to induce similar improvements in trunk reaction. Moreover, besides increasing the volume of kettlebell training, incorporating exercises with unexpected trunk reactions, as suggested by Pedersen and coworkers [19], may have improved the perturbed response further.

Healthcare work involves high and unpredictable (nonanticipated) loadings of the spine [7–10] that may cause musculoskeletal pain and injury [5, 10, 37–39] in turn potentially leading to long-term sickness, improving the response ability to sudden trunk perturbations by means of exercise and physical training may be crucial for the working life of a healthcare worker. Accordingly, the within-group reductions in stopping time and stopping distance that were observed in WORK and HOME reflect a faster response capacity to counteract sudden unexpected trunk perturbations which may protect against future injury or pain.

As the within-group changes in trunk perturbation characteristics did not differ between groups a cost-effectiveness analysis would favor HOME as this represents a low cost intervention modality compared to the investment in working hours, instructors, coaches, and additional training equipment that is needed with WORK. However, if the aim of the intervention is not only to improve the response ability to trunk perturbation but also to reduce musculoskeletal pain and increase maximal muscle strength as seen in WORK [22], the workplace-based exercise intervention might be considered more favorable. Nevertheless, the overall summed effect of these different qualities needs to be evaluated in future cost-effectiveness studies.

4.1. Strength and Limitations. It may be considered a strength of present study that the test method, unlike those used in previous trunk perturbation studies, involved a random sequence of either loading or unloading perturbations. As the condition of the perturbation is unknown and therefore more difficult to foresee and thus to create a preprogrammed reflex pattern, this method may better resemble real life conditions such as unexpected trips or slips during patient handling.

A methodological limitation of the present study was that we only measured EMG on ~70% of all the participants (162 subjects). Nevertheless, the overall high number of participants in this study compared with previous studies investigating the effects of training on trunk stability with sudden perturbations definitely strengthens the validity of present observations. Yet, an even higher number of subjects would have increased the statistical power and may potentially have changed the tendencies to significant findings. Furthermore, the lack of an inactive control group may be viewed as a limitation of the study as it is difficult to say whether the within-group changes in both groups were caused by the physical exercise or alternatively caused by seasonal variations or reflecting a learning effect per se. However, disfavoring the possibility of learning effects, the detailed test-retest analysis of the present experimental methods did not reveal any significant learning effect within two weeks [22].

5. Conclusion

Although training adherence was higher when performed at the workplace (WORK) compared to exercising at home (HOME) this additional training volume did not appear to promote any between-group differences in the responses to sudden trunk perturbations. As the main findings of the study, however, significant within-group changes in both groups were seen for stopping time during both loading and unloading trunk perturbations and for stopping distance during unloading perturbations. Even though the relative perturbed load was reduced following for the intervention period as indicated by the lower preactivation levels of the erector spinae muscles in WORK, higher muscle strength does not necessarily result in faster reactions to sudden unknown trunk perturbations, at least when training is performed using low-frequency (2 session per week), short-duration (10 min) exercise sessions. Accordingly, exercise interventions aiming at improving neuromechanical trunk reaction ability should not only focus on increasing muscle strength but also contain elements that challenge coordination and trunk response.

Conflict of Interests

The authors of the paper declare that they have no conflict of interests whatsoever. Further, the research has not received any funding or grant from any commercial source.

Authors' Contribution

Markus D. Jakobsen developed the mechanical muscle tests for this study. Markus D. Jakobsen and Lars L. Andersen designed the study. Markus D. Jakobsen, Emil Sundstrup, and Mikkel Brandt collected the data. Kenneth Jay educated the training instructors. Markus D. Jakobsen performed the Matlab analyses, and Lars L. Andersen performed the statistical analyses. All authors (Markus D. Jakobsen, Emil Sundstrup, Mikkel Brandt, Kenneth Jay, Per Aagaard, and

Lars L. Andersen) were involved in the data interpretation. Markus D. Jakobsen drafted the paper and all coauthors (Markus D. Jakobsen, Emil Sundstrup, Mikkel Brandt, Kenneth Jay, Per Aagaard, and Lars L. Andersen) revised it critically for scientific and intellectual content. All authors have read and approved the final paper.

Acknowledgments

The authors thank Anne Zöega Kristensen, Jørgen Skotte, and Klaus Hansen from the National Research Centre for the Working Environment for valuable technical assistance and support.

References

[1] P. M. Brooks, "The burden of musculoskeletal disease—a global perspective," *Clinical Rheumatology*, vol. 25, no. 6, pp. 778–781, 2006.

[2] L. Manchikanti, V. Singh, S. Datta, S. P. Cohen, J. A. Hirsch, and American Society of Interventional Pain Physicians, "Comprehensive review of epidemiology, scope, and impact of spinal pain," *Pain Physician*, vol. 12, no. 4, pp. E35–E70, 2009.

[3] J. N. Katz, "Lumbar disc disorders and low-back pain: socioeconomic factors and consequences," *The Journal of Bone and Joint Surgery—American Volume*, vol. 88, supplement 2, pp. 21–24, 2006.

[4] S. Dagenais, J. Caro, and S. Haldeman, "A systematic review of low back pain cost of illness studies in the United States and internationally," *Spine Journal*, vol. 8, no. 1, pp. 8–20, 2008.

[5] L. L. Andersen, A. Burdorf, N. Fallentin et al., "Patient transfers and assistive devices: prospective cohort study on the risk for occupational back injury among healthcare workers," *Scandinavian Journal of Work, Environment & Health*, vol. 40, no. 1, pp. 74–81, 2014.

[6] L. L. Andersen, T. Clausen, R. Persson, and A. Holtermann, "Perceived physical exertion during healthcare work and risk of chronic pain in different body regions: prospective cohort study," *International Archives of Occupational and Environmental Health*, vol. 86, no. 6, pp. 681–687, 2013.

[7] J. Skotte and N. Fallentin, "Low back injury risk during repositioning of patients in bed: the influence of handling technique, patient weight and disability," *Ergonomics*, vol. 51, no. 7, pp. 1042–1052, 2008.

[8] J. H. Skotte, M. Essendrop, A. F. Hansen, and B. Schibye, "A dynamic 3D biomechanical evaluation of the load on the low back during different patient-handling tasks," *Journal of Biomechanics*, vol. 35, no. 10, pp. 1357–1366, 2002.

[9] A. Burdorf and G. Sorock, "Positive and negative evidence of risk factors for back disorders," *Scandinavian Journal of Work, Environment and Health*, vol. 23, no. 4, pp. 243–256, 1997.

[10] J. Smedley, P. Egger, C. Cooper, and D. Coggon, "Manual handling activities and risk of low back pain in nurses," *Occupational and Environmental Medicine*, vol. 52, no. 3, pp. 160–163, 1995.

[11] J. N. Jensen, A. Holtermann, T. Clausen, O. S. Mortensen, I. G. Carneiro, and L. L. Andersen, "The greatest risk for low-back pain among newly educated female health care workers; body weight or physical work load?" *BMC Musculoskeletal Disorders*, vol. 13, article 87, 2012.

[12] J. K. Sluiter, E. M. de Croon, T. F. Meijman, and M. H. W. Frings-Dresen, "Need for recovery from work related fatigue and its role in the development and prediction of subjective health complaints," *Occupational and Environmental Medicine*, vol. 60, supplement 1, pp. i62–i70, 2003.

[13] P. W. Hodges, "Changes in motor planning of feedforward postural responses of the trunk muscles in low back pain," *Experimental Brain Research*, vol. 141, no. 2, pp. 261–266, 2001.

[14] A. Radebold, J. Cholewicki, M. M. Panjabi, and T. C. Patel, "Muscle response pattern to sudden trunk loading in healthy individuals and in patients with chronic low back pain," *Spine*, vol. 25, no. 8, pp. 947–954, 2000.

[15] D. G. Wilder, A. R. Aleksiev, M. L. Magnusson, M. H. Pope, K. F. Spratt, and V. K. Goel, "Muscular response to sudden load: a tool to evaluate fatigue and rehabilitation," *Spine*, vol. 21, no. 22, pp. 2628–2639, 1996.

[16] I. A. F. Stokes, J. R. Fox, and S. M. Henry, "Trunk muscular activation patterns and responses to transient force perturbation in persons with self-reported low back pain," *European Spine Journal*, vol. 15, no. 5, pp. 658–667, 2006.

[17] J. Cholewicki, A. P. D. Simons, and A. Radebold, "Effects of external trunk loads on lumbar spine stability," *Journal of Biomechanics*, vol. 33, no. 11, pp. 1377–1385, 2000.

[18] J. Cholewicki, S. P. Silfies, R. A. Shah et al., "Delayed trunk muscle reflex responses increase the risk of low back injuries," *Spine*, vol. 30, no. 23, pp. 2614–2620, 2005.

[19] M. T. Pedersen, M. Essendrop, J. H. Skotte, K. Jørgensen, B. Schibye, and N. Fallentin, "Back muscle response to sudden trunk loading can be modified by training among healthcare workers," *Spine*, vol. 32, no. 13, pp. 1454–1460, 2007.

[20] A. Navalgund, J. A. Buford, M. S. Briggs, and D. L. Givens, "Trunk muscle reflex amplitudes increased in patients with subacute, recurrent LBP treated with a 10-week stabilization exercise program," *Motor Control*, vol. 17, no. 1, pp. 1–17, 2013.

[21] M. T. Pedersen, M. B. Randers, J. H. Skotte, and P. Krustrup, "Recreational soccer can improve the reflex response to sudden trunk loading among untrained women," *Journal of Strength and Conditioning Research*, vol. 23, no. 9, pp. 2621–2626, 2009.

[22] M. D. Jakobsen, E. Sundstrup, M. Brandt, K. Jay, P. Aagaard, and L. L. Andersen, "Effect of workplace- versus home-based physical exercise on musculoskeletal pain among healthcare workers: a cluster randomized controlled trial," *Scandinavian Journal of Work, Environment & Health*, vol. 41, no. 2, pp. 153–163, 2015.

[23] M. D. Jakobsen, E. Sundstrup, M. Brandt et al., "Effect of workplace- versus home-based physical exercise on pain in healthcare workers: study protocol for a single blinded cluster randomized controlled trial," *BMC Musculoskeletal Disorders*, vol. 15, no. 1, article 119, 2014.

[24] Poster 1, http://www.jobogkrop.dk/Ondt-i-muskler-og-led/Ondt-i-nakke-skulder-og-arm/~/media/Files/MSB/Shop/PDF/Plakat-med-elastikoevelser-til-klinisk-personale.pdf.

[25] Poster 2, http://www.jobogkrop.dk/Ondt-i-muskler-og-led/Ondt-i-ryggen/~/media/Files/MSB/PDF/A2-Rygoevelser-v5-tryk.pdf.

[26] Poster 3, http://www.jobogkrop.dk/Ondt-i-muskler-og-led/Ondt-i-nakke-skulder-og-arm/~/media/Files/MSB/PDF/Exercises-with-elastic-band.pdf.

[27] M. Jakobsen, M. Brandt, E. Sundstrup, K. Jay, P. Aagaard, and L. Andersen, "Reliability of mechanical trunk responses during known and unknown trunk perturbations," *Journal of Applied Physiology*. In review.

[28] M. D. Jakobsen, E. Sundstrup, C. H. Andersen, P. Aagaard, and L. L. Andersen, "Muscle activity during leg strengthening exercise using free weights and elastic resistance: effects of ballistic vs controlled contractions," *Human Movement Science*, vol. 32, no. 1, pp. 65–78, 2013.

[29] E. Sundstrup, M. D. Jakobsen, C. H. Andersen, K. Jay, and L. L. Andersen, "Swiss ball abdominal crunch with added elastic resistance is an effective alternative to training machines," *International Journal of Sports Physical Therapy*, vol. 7, no. 4, pp. 372–380, 2012.

[30] J. Cohen, *Statistical Poer Analysis for the Behavioral Sciences*, Lawrence Erlbaum Associates, 1988.

[31] P. W. Hodges, G. L. Moseley, A. Gabrielsson, and S. C. Gandevia, "Experimental muscle pain changes feedforward postural responses of the trunk muscles," *Experimental Brain Research*, vol. 151, no. 2, pp. 262–271, 2003.

[32] M. Zedka, A. Prochazka, B. Knight, D. Gillard, and M. Gauthier, "Voluntary and reflex control of human back muscles during induced pain," *Journal of Physiology*, vol. 520, part 2, pp. 591–604, 1999.

[33] M. Lidegaard, R. B. Jensen, C. H. Andersen et al., "Effect of brief daily resistance training on occupational neck/shoulder muscle activity in office workers with chronic pain: randomized controlled trial," *BioMed Research International*, vol. 2013, Article ID 262386, 11 pages, 2013.

[34] P. Hodges, W. van den Hoorn, A. Dawson, and J. Cholewicki, "Changes in the mechanical properties of the trunk in low back pain may be associated with recurrence," *Journal of Biomechanics*, vol. 42, no. 1, pp. 61–66, 2009.

[35] M. T. Pedersen, M. Essendrop, J. H. Skotte, K. Jørgensen, and N. Fallentin, "Training can modify back muscle response to sudden trunk loading," *European Spine Journal*, vol. 13, no. 6, pp. 548–552, 2004.

[36] K. Jay, M. D. Jakobsen, E. Sundstrup et al., "Effects of kettlebell training on postural coordination and jump performance: a randomized controlled trial," *Journal of Strength and Conditioning Research*, vol. 27, no. 5, pp. 1202–1209, 2013.

[37] B. Bazrgari, A. Shirazi-Adl, and C. Larivière, "Trunk response analysis under sudden forward perturbations using a kinematics-driven model," *Journal of Biomechanics*, vol. 42, no. 9, pp. 1193–1200, 2009.

[38] S. A. Lavender, G. A. Mirka, R. W. Schoenmarklin, C. M. Sommerich, L. R. Sudhakar, and W. S. Marras, "The effects of preview and task symmetry on trunk muscle response to sudden loading," *Human Factors*, vol. 31, no. 1, pp. 101–115, 1989.

[39] M. L. Magnusson, A. Aleksiev, D. G. Wilder et al., "Unexpected load and asymmetric posture as etiologic factors in low back pain," *European Spine Journal*, vol. 5, no. 1, pp. 23–35, 1996.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 842804, 8 pages
http://dx.doi.org/10.1155/2015/842804

Review Article

The Role of Ankle Proprioception for Balance Control in relation to Sports Performance and Injury

Jia Han,[1,2] Judith Anson,[2] Gordon Waddington,[2] Roger Adams,[2] and Yu Liu[1]

[1] *School of Kinesiology, Shanghai University of Sport, Shanghai 200438, China*
[2] *Research Institute for Sport and Exercise, University of Canberra, Canberra, ACT 2600, Australia*

Correspondence should be addressed to Jia Han; jia.han@canberra.edu.au

Received 22 April 2015; Accepted 11 June 2015

Academic Editor: Massimiliano Pau

Balance control improvement is one of the most important goals in sports and exercise. Better balance is strongly positively associated with enhanced athletic performance and negatively associated with lower limb sports injuries. Proprioception plays an essential role in balance control, and ankle proprioception is arguably the most important. This paper reviews ankle proprioception and explores synergies with balance control, specifically in a sporting context. Central processing of ankle proprioceptive information, along with other sensory information, enables integration for balance control. When assessing ankle proprioception, the most generalizable findings arise from methods that are ecologically valid, allow proprioceptive signals to be integrated with general vision in the central nervous system, and reflect the signal-in-noise nature of central processing. Ankle proprioceptive intervention concepts driven by such a central processing theory are further proposed and discussed for the improvement of balance control in sport.

1. Introduction

In many sports, superior balance ability is necessary to achieve the highest competitive level and avoid lower limb injuries [1–3]. To control balance, the central nervous system (CNS) integrates visual, vestibular, and proprioceptive information to produce motor commands that coordinate the activation patterns of muscles [4–6]. Proprioception has been defined as one's ability to integrate the sensory signals from various mechanoreceptors to thereby determine body position and movements in space [7, 8], and it plays a crucial role in balance control [5, 6, 9–11]. Theoretically, proprioceptive information from every part of the body contributes to balance control. This includes visual proprioception, as demonstrated by Lee and Aronson [12], although in sport the visual channel is often occupied with processing information about opponents or ball flight, so other proprioceptive sources are needed [13]. Sensory reweighting theory, which holds that the CNS can shift reliance to more reliable sources of information to optimize balance control [5, 6, 9] suggests that, for example, where vision is being used for tracking an activity in the external environment, the CNS may rely more on proprioceptive information from particular parts of the body for balance control. Ankle proprioception may be one of the more important components contributing to balance control in sport, because during most sports activities, the ankle-foot complex is the only part of the body contacting the ground. Ankle proprioception provides essential information to enable adjustment of ankle positions and movements of the upper body, in order to successfully perform the complex motor tasks required in elite sport [14, 15].

Ankle proprioception can be altered by general [16] and sport-specific training [17–19], sport-related injuries [20–25], and sport-induced fatigue [26, 27], all of which may subsequently lead to altered balance ability. The assessment of ankle proprioception in healthy individuals and subjects with musculoskeletal or neurological disorders has been addressed in three recent reviews [5, 7, 28]. The purpose of this review is to explore the association between ankle proprioception and balance control in a sporting context, and their roles in sport performance and sport injury. This provides an opportunity to determine the most appropriate methods for assessing ankle proprioception in athletes and potential central processing mechanisms underlying balance control. Understanding the mechanism underlying ankle proprioception and balance control may foreshadow optimal interventions to improve balance control in sport.

2. Balance Control and Ankle Proprioception in Sports Performance

Balance ability and ankle proprioception are both related to competition level in a range of sports. A systematic review on balance ability and athletic performance found that static balance ability of rifle shooters and archers was associated with their shooting accuracy, and dynamic balance ability of ice hockey players displayed a significant relationship with maximum skating speed [3]. In addition, a recent study investigating balance ability of a group of athletes from soccer, handball, basketball, and volleyball found that the balance ability of male athletes was significantly correlated with their agility performance [29]. This evidence suggests that balance control is fundamental to sports performance.

Similarly, ankle proprioception and sports performance are related. Han et al. [18] measured ankle proprioception of 100 elite athletes from 5 different sports—aerobic gymnastics, soccer, swimming, badminton, and sports dancing—and found that ankle proprioception scores were significantly predictive of sport performance level, extending up to Olympic level. In a subsequent study [30], the authors assessed proprioception at the knee, spine, shoulder, and hand in addition to the ankle and found proprioception at the shoulder and spine were also significantly associated with competitive level in these elite athletes. Of these three critical body sites—the ankle, shoulder, and spine—ankle proprioception was correlated most strongly with sport competition level and was the most significant predictor of sports performance [30]. These findings highlighted the importance of ankle proprioception in sporting success.

Thus, although visual [1] and vestibular [31] functions play important roles in balance control in sport, ankle proprioception, within the proprioceptive system, appears to be the most critical for balance control contributing to sport performance.

3. Balance Control and Ankle Proprioception in Sports Injury

Both balance control and ankle proprioception are negatively associated with ankle injuries [2, 35]. The relationship between poor balance control and heightened injury risk was identified 30 years ago, when in 1984 Tropp et al. [36] found that ankle injuries were almost 4 times more prevalent in soccer players with poor balance in comparison to those with normal balance ability. Similarly, Watson [37] found hurdling athletes and Gaelic football players with poor balance had nearly twice as many ankle injuries relative to their counterparts with normal balance. In addition, balance ability was found to be significantly associated with ankle injury risk in both younger male and female basketball players [38]. A recent systematic review summarized the available evidence and suggested that poorer balance ability is an intrinsic factor associated with increased ankle injury risk [35].

Similar reports of the relationship between ankle proprioception and ankle injury risk are also noted in the literature. For example, a longitudinal study found ankle proprioception could predict ankle injuries in college basketball players [39]. In addition, basketball players with poorer ankle proprioception used an altered pattern of cocontraction of ankle plantarflexors and dorsiflexors, which in turn resulted in greater impact force at the moment of landing associated with higher risk of ankle injury [40]. Ankle proprioception is one of the intrinsic factors associated with ankle injury, as identified by Witchalls et al. in their systematic review [35].

Ankle injuries often lead to disruption of muscles and tendons with associated damage to inherent mechanoreceptors [5, 41], which detrimentally alter the quality of proprioceptive information required for balance control. Unrehabilitated, impaired ankle proprioception after ankle injury [20–25] can subsequently result in long-term deterioration of postural and balance control. Gymnasts, dancers, and military sportsmen with poorer ankle proprioception after injury demonstrate worse performance in both static and dynamic postural and balance control tasks [42–45]. In addition, the common motor program hypothesis [46] suggests that there will be bilateral impairments in ankle proprioception in both the injured and uninjured sides [22, 47]. This bilateral impairment is also evident in postural and balance performance relative to healthy controls [48].

These findings suggest that ankle proprioception is closely related to balance control in sport injuries, and balance ability may be significantly affected by impaired ankle proprioception after injuries.

4. Mechanisms Underlying Ankle Proprioception to Balance Control in Sport

Sensory noise and sensory information reweighting are two possible mechanisms for the optimal use of sensory information in balance control [5, 6, 9]. Both models highlight the role of central processing in balance control and may explain the importance of ankle proprioception for balance control in sport. For example, ankle proprioception is superior in gymnasts, sports dancers, badminton players, soccer players, and swimmers [18, 30, 49, 50], suggesting that through years of sport-specific practice, ankle proprioception may be processed more efficiently and reliably in the brain [7]. If the CNS uses a reweighting strategy relying on more reliable sources of information to optimize balance control [5, 6, 9], refined ankle proprioception, with its signal-to-noise ratio reduced through practice, could be one of these more reliable sources of proprioceptive information in particular sports.

In addition, the observation of bilateral deficits in both ankle proprioception and balance control after ankle injury [22, 47] favored a central motor program view of bilateral limb movement control [51]. The data indicate that a higher-order central mechanism may exist for proprioceptive

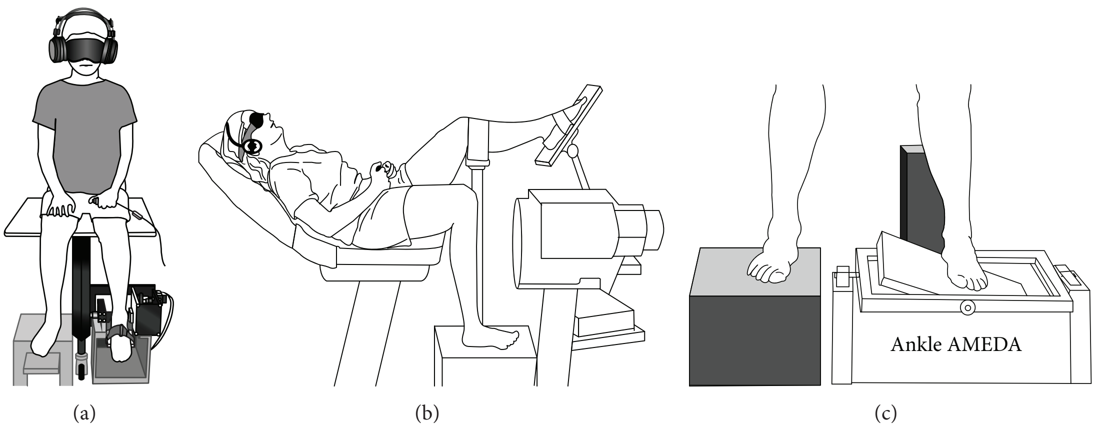


Figure 1: Examples of ankle proprioceptive assessment methods. (a) depicts the threshold to detection of passive motion (TTDPM) method, adapted from Yasuda et al. [32]; (b) depicts the joint position reproduction (JPR) method, adapted from Willems et al. [33]; and (c) depicts the active movement extent discrimination assessment (AMEDA) method, adapted from Symes et al. [34].

information processing contributing to postural and balance control [7, 52]. Indeed, a recent brain imaging study suggested that beyond peripheral reflex mechanisms, central processing of proprioceptive signals from the foot-ankle complex is essential for postural and balance control [53].

If central processing of proprioceptive information links ankle proprioception and balance control, then this has implications for ankle proprioceptive assessment. It suggests that the most appropriate measurement technologies are those that are relevant to normal function and that encompass ecologically valid components related to balance function [7]. This issue is also important for determination of the optimum ankle proprioceptive intervention to improve balance control in sport.

5. Selection of Ankle Proprioceptive Assessment Method Relevant to Balance in Sport

Proprioception can be assessed using different technologies/methodologies [5, 7, 28, 54]. There are three main technologies/methodologies used for ankle proprioceptive assessment. These are thresholds to detection of passive motion (TTDPM) [55–57], joint position reproduction (JPR) [26, 58, 59], and active movement extent discrimination assessment (AMEDA) [17, 60, 61]. The advantages and disadvantages of these testing protocols have been discussed in a recent review paper [7]. The current review focuses on the selection of appropriate measurement techniques relevant to balance function.

The three different technologies used for testing ankle proprioception are presented in Figure 1. It is clear from Figure 1 that some of the technologies adopt a nonweight bearing, either lying or sitting, position during testing and block both visual and audio information in order to assess "pure" ankle proprioception (Figures 1(a) and 1(b)). The ecological validity of these tests has been questioned however [62, 63] because the assessment conditions are markedly dissimilar to normal function and can therefore contribute little to understanding the role that ankle proprioception plays in balance control during sports and daily activities [7]. If central processing of ankle proprioception is crucial for balance control, ankle proprioceptive assessment should be conducted to reflect normal function that encompasses all proprioceptive signals arising from muscles, joints, and skin normally projected for integration in the CNS.

In a method designed to increase ecological validity, an upright, weight-bearing stance TTDPM option has been used, along with the AMEDA apparatus (Figure 1(c)), for ankle proprioception assessment [34, 56, 57, 64]. These techniques ensure activation of muscles, joint capsule compression, and skin stretch. More recently, Witchalls et al. [43, 65] have developed a "walk-across" AMEDA in order to assess dynamic ankle proprioception during normal gait, involving dynamic balance control.

In addition to ecological validity, assessment of ankle proprioception should acknowledge and incorporate the signal-noise nature of central processing [66–68]. It has been argued that when processing proprioceptive information, the brain has to deal with noise in the CNS arising from background or spontaneous neural activity [66, 69], which contributes to uncertainty in making decisions about positions and movements of a joint in space [7, 69]. Similarly, in balance control, the CNS has to process multiple sensory signals occurring against a noise background in order to determine body sway in space [6].

Waddington and Adams [70] applied signal detection theory (SDT) [71–73] to deal with noise-associated uncertainty in making judgments about ankle movements and positions. To do this, participants are required on any one trial to make an absolute judgment regarding one of five possible ankle movements. Nonparametric SDT Receiver Operating Characteristic (ROC) analysis [73] is then used to compare responses to pairs of ankle movements. The area under the ROC curve (AUC) [73] is used as the measurement of ankle proprioceptive sensitivity, representing a participant's ability to discriminate between the five ankle movements [7]. SDT gives a means to take an individual's uncertainty into account and produce an unbiased estimate of an individual's ankle proprioceptive performance [7].

In this way, the AMEDA technique both fulfills ecological validity and captures data in such a way as to address the signal-noise nature of central processing of proprioceptive information relevant to balance function. Although this method has not yet been used to determine the precise association between ankle proprioception measured with an AMEDA and balance ability in athletes, Guo et al. [56] has assessed ankle proprioception in an upright, weight-bearing stance using the TTDPM method and found that ankle proprioception can explain 53% and 44% of variance in anterior-posterior and medial-lateral posture sway directions, respectively.

6. Ankle Proprioceptive Intervention for Improvement of Balance Control in Sport

Both ankle proprioception and balance control are essential in sports [2, 3, 18, 40], and *passive* or *active* interventions to improve ankle proprioception and balance control, particularly after ankle injury, have been extensively reported in the literature [5, 11]. Regardless of the type of intervention—passive or active—central processing of ankle proprioception is likely to be critical for balance control. If this is the case, ankle proprioceptive intervention should focus on central processing mechanisms to improve balance control in order to enhance sports performance and minimize injuries.

6.1. Passive Intervention. A number of studies have explored effects of passive interventions, such as taping, bracing, compressing, or sport shoe insoles, on ankle proprioception [70, 74–78], with most finding passive techniques being not effective in improving ankle proprioception. Ankle taping and bracing, for example, are commonly used by athletes following a sprained ankle. Two systematic reviews consistently found ankle proprioception was not enhanced with the use of ankle tape or braces in athletes suffering repeated ankle sprains and with functional ankle instability [79, 80]. These findings suggest that ankle proprioception does not benefit from the restriction of ankle movement and/or from elastic resistance [81] imposed at the ankle joint.

In contrast, there is some evidence that the use of insoles, another passive intervention, has a positive effect on ankle proprioception in soccer players [70, 76]. It has been proposed that the use of textured insoles induces "essential noise" in the CNS, which in turn resulted in increased perception of information to support motor performance [82]. Consistent with this point of view, a systematic review found insoles with vibrating elements improved balance in older people [83]. However, not all types of insoles were found to be beneficial [83], presumably because only some proprioceptive signals combine with certain types of noise to enhance perception of proprioceptive information [82]. While ankle proprioception and balance control may be improved through a signal-noise resonance mechanism in the CNS [82], selection of appropriate techniques such as the height, texture, and vibration of particular insoles to optimize the signal-noise ratio currently needs further exploration if such design modifications are to be introduced into sports footwear.

6.2. Active Intervention. Various active exercise interventions, delivered in a task-specific paradigm, have been found to be effective for the improvement of ankle proprioception. It has been proposed that this occurs through neural mechanisms such as neural learning and neural plasticity [5]. Neural learning effects associated with ankle proprioception may be rapid. For example, Witchalls and colleagues [43] found athletes with chronic ankle instability improved their ankle proprioception in one session through ankle AMEDA test-retest practice. This improvement was thought to be due to central processing modifications. The authors [43] argue that mechanoreceptors at the ankle joint do not change during the repeated proprioceptive testing (the time is too short), and proprioceptive information originating from peripheral structures at the ankle does not significantly change either. Therefore, faster neural learning processes are likely to play the key role in improving ankle proprioception. Further research is needed to explore to what extent the improvement in ankle proprioception through neural learning translates to better balance control in the sporting context and whether such learning should be conducted as explicit or implicit learning [84].

In contrast, some neural changes may require weeks, months, or even years of practice. Several weeks of wobble-board training [17, 85, 86], Tai Chi exercise [56, 87, 88], and other specifically designed exercise programs [89–91] have been shown to improve ankle proprioception and balance control in athletes, university students, and older people, with or without ankle instability. Kiers et al. [92] argue that exercise on an unstable surface might not target ankle proprioception *per se* but rather trains the CNS to shift the weighting of sources of proprioceptive signals to improve balance. If this is the case, yachting and figure-skating athletes whose daily activities involve performing motor tasks on an unstable surface would benefit from exercise on a similar surface [93]. What is not known is whether combined exercise on both unstable and stable surfaces provides even greater benefits for ankle proprioception and balance control than just active training on unstable surfaces.

Apart from training surfaces, another issue associated with active ankle proprioceptive training is whether the training should focus on the injured side or should involve both sides after sports injury. Given a significant and positive correlation found between performance of both ankles in healthy and injured participants [22, 46, 61], ankle proprioceptive training should also involve the intact side [46, 61]. Some evidence suggests that motor skills are able to be transferred between hemispheres [94], indicating that training on the uninjured side could benefit the affected side. However, interhemispheric motor skill transfer may be affected by ageing [95] and limb dominance [96]. In addition, a recent study investigating sensory reweighting of proprioceptive information from each leg during balance control found that proprioceptive signals from each leg were weighted independently, and weighting of proprioceptive signals of one leg had no effect on the weight of the proprioceptive information of the other leg [9]. Taken together, proprioceptive training is likely to be most beneficial for improving ankle proprioception *per se* when conducted on each leg and by optimizing ankle proprioceptive information reweighting for balance control in sport. Future research is needed to elucidate the CNS process associated with active interventions.

6.3. Other Considerations. Although developing better proprioception and balance control through training is a common goal for athletes and there is mounting evidence suggesting that active interventions such as wobble board training aid in doing this, there may also be a significant genetic component to proprioceptive ability and balance control. This is likely to be more evident in elite athletes, who are striving to be the best of the best, where training levels are already extensive. In studies by Han et al. [18, 30], it was reported that ankle proprioception scores were not significantly correlated with years of training, suggesting that the amount of improvement in ankle proprioception associated with sports training may be constrained by biologically determined factors [30]. From studies of twins performing balance control tasks, it has been suggested that there may also be a genetic component contributing to balance control [97, 98]. If this is the case, in order to achieve the highest competitive level an athlete may also need to have genetic potential for better ankle proprioception and balance control. If correct, future sport talent identification may need to consider natural aptitude in both ankle proprioception and balance ability when selecting potential elite athletes.

7. Conclusion

Proprioception plays an essential role in balance control, and ankle proprioception is arguably the most important aspect of this. Central processing of ankle proprioceptive information, along with other sensory information, enables integration for postural and balance control. When assessing ankle proprioception for generalization to applied situations, the method used should have ecological validity and allow proprioceptive signals to be integrated in the central nervous system, in order to reflect the signal-noise nature of central processing in sports activities. In addition, ankle proprioceptive interventions, passive or active, should therefore be predicated on discriminating signal from noise in central processing, to attain optimal outcomes.

Conflict of Interests

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

Acknowledgment

This study is sponsored by Shanghai Pujiang Program Grant no. 15PJ1407600.

References

[1] H. Kiers, J. van Dieën, H. Dekkers, H. Wittink, and L. Vanhees, "A systematic review of the relationship between physical activities in sports or daily life and postural sway in upright stance," *Sports Medicine*, vol. 43, no. 11, pp. 1171–1189, 2013.

[2] C. Hrysomallis, "Relationship between balance ability, training and sports injury risk," *Sports Medicine*, vol. 37, no. 6, pp. 547–556, 2007.

[3] C. Hrysomallis, "Balance ability and athletic performance," *Sports Medicine*, vol. 41, no. 3, pp. 221–232, 2011.

[4] A. W. Shumway-Cook and M. H. Woollacott, *Motor Control: Translating Research into Clinical Practice*, Lippincott Williams & Wilkins, Baltimore, Md, USA, 4th edition, 2013.

[5] U. Röijezon, N. C. Clark, and J. Treleaven, "Proprioception in musculoskeletal rehabilitation. Part 1: basic science and principles of assessment and clinical interventions," *Manual Therapy*, vol. 20, no. 3, pp. 368–377, 2015.

[6] R. A. Speers, A. D. Kuo, and F. B. Horak, "Contributions of altered sensation and feedback responses to changes in coordination of postural control due to aging," *Gait & Posture*, vol. 16, no. 1, pp. 20–30, 2002.

[7] J. Han, G. Waddington, R. Adams, J. Anson, and Y. Liu, "Assessing proprioception: a critical review of methods," *Journal of Sport and Health Science*, 2015.

[8] D. J. Goble, "Proprioceptive acuity assessment via joint position matching: from basic science to general practice," *Physical Therapy*, vol. 90, no. 8, pp. 1176–1184, 2010.

[9] J. H. Pasma, T. A. Boonstra, S. F. Campfens, A. C. Schouten, and H. van der Kooij, "Sensory reweighting of proprioceptive information of the left and right leg during human balance control," *Journal of Neurophysiology*, vol. 108, no. 4, pp. 1138–1148, 2012.

[10] S. Bouisset and M.-C. Do, "Posture, dynamic stability, and voluntary movement," *Neurophysiologie Clinique*, vol. 38, no. 6, pp. 345–362, 2008.

[11] N. C. Clark, U. Röijezon, and J. Treleaven, "Proprioception in musculoskeletal rehabilitation. Part 2: clinical assessment and intervention," *Manual Therapy*, vol. 20, no. 3, pp. 378–387, 2015.

[12] D. N. Lee and E. Aronson, "Visual proprioceptive control of standing in human infants," *Perception & Psychophysics*, vol. 15, no. 3, pp. 529–532, 1974.

[13] K. Davids, *Visual Perception and Action in Sport*, Taylor & Francis, Eastbourne, UK, 1999.

[14] I. di Giulio, C. N. Maganaris, V. Baltzopoulos, and I. D. Loram, "The proprioceptive and agonist roles of gastrocnemius, soleus and tibialis anterior muscles in maintaining human upright posture," *The Journal of Physiology*, vol. 587, no. 10, pp. 2399–2416, 2009.

[15] S. Sasagawa, J. Ushiyama, K. Masani, M. Kouzaki, and H. Kanehisa, "Balance control under different passive contributions of the ankle extensors: quiet standing on inclined surfaces," *Experimental Brain Research*, vol. 196, no. 4, pp. 537–544, 2009.

[16] T. Winter, H. Beck, A. Walther, H. Zwipp, and S. Rein, "Influence of a proprioceptive training on functional ankle stability in young speed skaters—a prospective randomised study," *Journal of Sports Sciences*, vol. 33, no. 8, pp. 831–840, 2015.

[17] G. S. Waddington and R. D. Adams, "The effect of a 5-week wobble-board exercise intervention on ability to discriminate different degrees of ankle inversion, barefoot and wearing shoes: a study in healthy elderly," *Journal of the American Geriatrics Society*, vol. 52, no. 4, pp. 573–576, 2004.

[18] J. Han, J. Anson, G. Waddington, and R. Adams, "Sport attainment and proprioception," *International Journal of Sports Science and Coaching*, vol. 9, no. 1, pp. 159–170, 2014.

[19] A. Kynsburg, G. Panics, and T. Halasi, "Long-term neuromuscular training and ankle joint position sense," *Acta Physiologica Hungarica*, vol. 97, no. 2, pp. 183–191, 2010.

[20] J. B. Witchalls, G. Waddington, R. Adams, and P. Blanch, "Chronic ankle instability affects learning rate during repeated proprioception testing," *Physical Therapy in Sport*, vol. 15, no. 2, pp. 106–111, 2014.

[21] J. Hertel, "Sensorimotor deficits with ankle sprains and chronic ankle instability," *Clinics in Sports Medicine*, vol. 27, no. 3, pp. 353–370, 2008.

[22] G. Waddington and R. Adams, "Discrimination of active plantarflexion and inversion movements after ankle injury," *Australian Journal of Physiotherapy*, vol. 45, no. 1, pp. 7–13, 1999.

[23] J. Munn, S. J. Sullivan, and A. G. Schneiders, "Evidence of sensorimotor deficits in functional ankle instability: a systematic review with meta-analysis," *Journal of Science and Medicine in Sport*, vol. 13, no. 1, pp. 2–12, 2010.

[24] S. Yokoyama, N. Matsusaka, K. Gamada, M. Ozaki, and H. Shindo, "Position-specific deficit of joint position sense in ankles with chronic functional instability," *Journal of Sports Science and Medicine*, vol. 7, no. 4, pp. 480–485, 2008.

[25] T. Nakasa, K. Fukuhara, N. Adachi, and M. Ochi, "The deficit of joint position sense in the chronic unstable ankle as measured by inversion angle replication error," *Archives of Orthopaedic and Trauma Surgery*, vol. 128, no. 5, pp. 445–449, 2008.

[26] F. Mohammadi and A. Roozdar, "Effects of fatigue due to contraction of evertor muscles on the ankle joint position sense in male soccer players," *The American Journal of Sports Medicine*, vol. 38, no. 4, pp. 824–828, 2010.

[27] M. A. Sandrey and T. E. Kent, "The effects of eversion fatigue on frontal plane joint position sense in the ankle," *Journal of Sport Rehabilitation*, vol. 17, no. 3, pp. 257–268, 2008.

[28] S. Hillier, M. Immink, and D. Thewlis, "Assessing proprioception: a systematic review of possibilities," *Neurorehabilitation and Neural Repair*, 2015.

[29] D. Sekulic, M. Spasic, D. Mirkov, M. Cavar, and T. Sattler, "Gender-specific influences of balance, speed, and power on agility performance," *Journal of Strength and Conditioning Research*, vol. 27, no. 3, pp. 802–811, 2013.

[30] J. Han, G. Waddington, J. Anson, and R. Adams, "Level of competitive success achieved by elite athletes and multi-joint proprioceptive ability," *Journal of Science and Medicine in Sport*, vol. 18, no. 1, pp. 77–81, 2015.

[31] K. Aligene and E. Lin, "Vestibular and balance treatment of the concussed athlete," *NeuroRehabilitation*, vol. 32, no. 3, pp. 543–553, 2013.

[32] K. Yasuda, Y. Sato, N. Iimura, and H. Iwata, "Allocation of attentional resources toward a secondary cognitive task leads to compromised ankle proprioceptive performance in healthy young adults," *Rehabilitation Research and Practice*, vol. 2014, Article ID 170304, 7 pages, 2014.

[33] T. Willems, E. Witvrouw, J. Verstuyft, P. Vaes, and D. De Clercq, "Proprioception and muscle strength in subjects with a history of ankle sprains and chronic instability," *Journal of Athletic Training*, vol. 37, no. 4, pp. 487–493, 2002.

[34] M. Symes, G. Waddington, and R. Adams, "Depth of ankle inversion and discrimination of foot positions," *Perceptual and Motor Skills*, vol. 111, no. 2, pp. 475–484, 2010.

[35] J. Witchalls, P. Blanch, G. Waddington, and R. Adams, "Intrinsic functional deficits associated with increased risk of ankle injuries: a systematic review with meta-analysis," *British Journal of Sports Medicine*, vol. 46, no. 7, pp. 515–523, 2012.

[36] H. Tropp, J. Ekstrand, and J. Gillquist, "Stabilometry in functional instability of the ankle and its value in predicting injury," *Medicine & Science in Sports & Exercise*, vol. 16, no. 1, pp. 64–66, 1984.

[37] A. W. S. Watson, "Ankle sprains in players of the field-games Gaelic football and hurling," *The Journal of Sports Medicine and Physical Fitness*, vol. 39, no. 1, pp. 66–70, 1999.

[38] T. A. McGuine, J. J. Greene, T. Best, and G. Leverson, "Balance as a predictor of ankle injuries in high school basketball players," *Clinical Journal of Sport Medicine*, vol. 10, no. 4, pp. 239–244, 2000.

[39] K. A. Payne, K. Berg, and R. W. Latin, "Ankle injuries and ankle strength, flexibility, and proprioception in college basketball players," *Journal of Athletic Training*, vol. 32, no. 3, pp. 221–225, 1997.

[40] S. N. Fu and C. W. Y. Hui-Chan, "Are there any relationships among ankle proprioception acuity, pre-landing ankle muscle responses, and landing impact in man?" *Neuroscience Letters*, vol. 417, no. 2, pp. 123–127, 2007.

[41] C. C. Herb and J. Hertel, "Current concepts on the pathophysiology and management of recurrent ankle sprains and chronic ankle instability," *Current Physical Medicine and Rehabilitation Reports*, vol. 2, no. 1, pp. 25–34, 2014.

[42] J. B. Witchalls, P. Newman, G. Waddington, R. Adams, and P. Blanch, "Functional performance deficits associated with ligamentous instability at the ankle," *Journal of Science and Medicine in Sport*, vol. 16, no. 2, pp. 89–93, 2013.

[43] J. Witchalls, G. Waddington, P. Blanch, and R. Adams, "Ankle instability effects on joint position sense when stepping across the active movement extent discrimination apparatus," *Journal of Athletic Training*, vol. 47, no. 6, pp. 627–634, 2012.

[44] D. M. Forkm, C. Koczur, R. Battle, and R. A. Newton, "Evaluation of kinesthetic deficits indicative of balance control in gymnasts with unilateral chronic ankle sprains," *Journal of Orthopaedic & Sports Physical Therapy*, vol. 23, no. 4, pp. 245–250, 1996.

[45] J. Leanderson, E. Eriksson, C. Nilsson, and A. Wykman, "Proprioception in classical ballet dancers: a prospective study of the influence of an ankle sprain on proprioception in the ankle joint," *The American Journal of Sports Medicine*, vol. 24, no. 3, pp. 370–374, 1996.

[46] G. Waddington and R. Adams, "Ability to discriminate movements at the ankle and knee is joint specific," *Perceptual and Motor Skills*, vol. 89, no. 3, pp. 1037–1041, 1999.

[47] E. C. W. Lim and M. H. Tan, "Side-to-side difference in joint position sense and kinesthesia in unilateral functional ankle instability," *Foot & Ankle International*, vol. 30, no. 10, pp. 1011–1017, 2009.

[48] P. O. McKeon and J. Hertel, "Systematic review of postural control and lateral ankle instability, Part I: can deficits be detected with instrumented testing?" *Journal of Athletic Training*, vol. 43, no. 3, pp. 293–304, 2008.

[49] T. Aydin, Y. Yildiz, C. Yildiz, S. Atesalp, and T. A. Kalyon, "Proprioception of the ankle: a comparison between female teenaged gymnasts and controls," *Foot & Ankle International*, vol. 23, no. 2, pp. 123–129, 2002.

[50] T. Aydin, Y. Yildiz, C. Yildiz, and T. A. Kalyon, "Effects of extensive training on female teenage gymnasts' active and passive

ankle-joint position sense," *Journal of Sport Rehabilitation*, vol. 11, no. 1, pp. 1–10, 2002.

[51] J. J. Summers and J. G. Anson, "Current status of the motor program: revisited," *Human Movement Science*, vol. 28, no. 5, pp. 566–577, 2009.

[52] J. Han, G. Waddington, R. Adams, and J. Anson, "A proprioceptive ability factor underlying all proprioception tests? Response to Tremblay (2013)," *Perceptual and Motor Skills*, vol. 119, no. 1, pp. 301–304, 2014.

[53] D. J. Goble, J. P. Coxon, A. van Impe et al., "Brain activity during ankle proprioceptive stimulation predicts balance performance in young and older adults," *The Journal of Neuroscience*, vol. 31, no. 45, pp. 16344–16352, 2011.

[54] E. Hagert, "Proprioception of the wrist joint: a review of current concepts and possible implications on the rehabilitation of the wrist," *Journal of Hand Therapy*, vol. 23, no. 1, pp. 2–17, 2010.

[55] T. J. Hubbard and T. W. Kaminski, "Kinesthesia is not affected by functional ankle instability status," *Journal of Athletic Training*, vol. 37, no. 4, pp. 481–486, 2002.

[56] L.-Y. Guo, C.-P. Yang, Y.-L. You et al., "Underlying mechanisms of Tai-Chi-Chuan training for improving balance ability in the elders," *Chinese Journal of Integrative Medicine*, vol. 20, no. 6, pp. 409–415, 2014.

[57] J. Son, J. A. Ashton-Miller, and J. K. Richardson, "Do ankle orthoses improve ankle proprioceptive thresholds or unipedal balance in older persons with peripheral neuropathy?" *American Journal of Physical Medicine and Rehabilitation*, vol. 89, no. 5, pp. 369–375, 2010.

[58] J. H. You, S. Saliba, and E. Saliba, "Use of a combination of ankle pressure and SENSERite system to treat older adults with impaired ankle proprioception: a single-blind experimental study," *Archives of Physical Medicine and Rehabilitation*, vol. 90, no. 1, pp. 102–108, 2009.

[59] M. South and K. P. George, "The effect of peroneal muscle fatigue on ankle joint position sense," *Physical Therapy in Sport*, vol. 8, no. 2, pp. 82–87, 2007.

[60] J. Han, R. Adams, G. Waddington, and J. Anson, "Ability to discriminate movements at multiple joints around the body: global or site-specific," *Perceptual and Motor Skills*, vol. 116, no. 1, pp. 59–68, 2013.

[61] J. Han, J. Anson, G. Waddington, and R. Adams, "Proprioceptive performance of bilateral upper and lower limb joints: side-general and site-specific effects," *Experimental Brain Research*, vol. 226, no. 3, pp. 313–323, 2013.

[62] J. I. Laszlo, "Motor control and learning: how far do the experimental tasks restrict our theoretical insight?" in *Approaches to the Study of Motor Control and Learning*, J. J. Summers, Ed., pp. 47–79, Elsevier, Amsterdam, The Netherlands, 1992.

[63] J. A. Ashton-Miller, "Proprioceptive thresholds at the ankle: implications for the prevention of ligament injury," in *Proprioception and Neuromuscular Control in Joint Stability*, S. M. Lephart and F. H. Fu, Eds., pp. 279–289, Human Kinetics Publisher, Champaign, Ill, USA, 2000.

[64] G. Black, G. Waddington, and R. Adams, "Relative sensitivity of depth discrimination for ankle inversion and plantar flexion movements," *Perceptual and Motor Skills*, vol. 118, no. 1, pp. 115–125, 2014.

[65] J. Witchalls, G. Waddington, R. Adams, and P. Blanch, "Evaluation of the relative contribution of peripheral and focal vision to proprioceptive differentiation of underfoot inversion angles in young elite athletes," *Perceptual and Motor Skills*, vol. 117, no. 3, pp. 923–934, 2013.

[66] A. A. Faisal, L. P. J. Selen, and D. M. Wolpert, "Noise in the nervous system," *Nature Reviews Neuroscience*, vol. 9, no. 4, pp. 292–303, 2008.

[67] R. J. van Beers, P. Haggard, and D. M. Wolpert, "The role of execution noise in movement variability," *Journal of Neurophysiology*, vol. 91, no. 2, pp. 1050–1063, 2004.

[68] K. E. Jones, A. F. D. C. Hamilton, and D. M. Wolpert, "Sources of signal-dependent noise during isometric force production," *Journal of Neurophysiology*, vol. 88, no. 3, pp. 1533–1544, 2002.

[69] R. J. van Beers, P. Baraduc, and D. M. Wolpert, "Role of uncertainty in sensorimotor control," *Philosophical Transactions of the Royal Society B: Biological Sciences*, vol. 357, no. 1424, pp. 1137–1145, 2002.

[70] G. Waddington and R. Adams, "Textured insole effects on ankle movement discrimination while wearing athletic shoes," *Physical Therapy in Sport*, vol. 1, no. 4, pp. 119–128, 2000.

[71] J. A. Swets, *Tulips to Thresholds: Counterpart Careers of the Author and Signal Detection Theory*, Peninsula Publishing, Newport Beach, Calif, USA, 2010.

[72] J. A. Swets, R. M. Dawes, and J. Monahan, "Better decisions through science," *Scientific American*, vol. 283, no. 4, pp. 82–87, 2000.

[73] D. McNicol, *A Primer of Signal Detection Theory*, Routledge, New York, NY, USA, 2004.

[74] S. H. You, K. P. Granata, and L. K. Bunker, "Effects of circumferential ankle pressure on ankle proprioception, stiffness, and postural stability: a preliminary investigation," *Journal of Orthopaedic & Sports Physical Therapy*, vol. 34, no. 8, pp. 449–460, 2004.

[75] K. M. Refshauge, J. Raymond, S. L. Kilbreath, L. Pengel, and I. Heijnen, "The effect of ankle taping on detection of inversion-eversion movements in participants with recurrent ankle sprain," *The American Journal of Sports Medicine*, vol. 37, no. 2, pp. 371–375, 2009.

[76] G. Waddington and R. Adams, "Football boot insoles and sensitivity to extent of ankle inversion movement," *British Journal of Sports Medicine*, vol. 37, no. 2, pp. 170–174, 2003.

[77] D. M. Hopper, T. L. Grisbrook, M. Finucane, and K. Nosaka, "Effect of ankle taping on angle and force matching and strength of the plantar flexors," *Physical Therapy in Sport*, vol. 15, no. 4, pp. 254–260, 2014.

[78] S. H. Bentham, J. Hatcher, I. Horsley, and L. R. Mc Naughton, "The influence of an aircast sports stirrup ankle brace on the ankle joint proprioception of professional soccer players," *Sports Medicine, Training and Rehabilitation*, vol. 10, no. 4, pp. 223–233, 2001.

[79] J. Raymond, L. L. Nicholson, C. E. Hiller, and K. M. Refshauge, "The effect of ankle taping or bracing on proprioception in functional ankle instability: a systematic review and meta-analysis," *Journal of Science and Medicine in Sport*, vol. 15, no. 5, pp. 386–392, 2012.

[80] K. W. Janssen and S. J. Kamper, "Ankle taping and bracing for proprioception," *British Journal of Sports Medicine*, vol. 47, no. 8, pp. 527–528, 2013.

[81] J. Han, G. Waddington, J. Anson, and R. Adams, "Does elastic resistance affect finger pinch discrimination?" *Human Factors*, vol. 55, no. 5, pp. 976–984, 2013.

[82] K. Davids, R. Shultleworth, C. Button, I. Renshew, and P. Glazier, "'Essential noise'—enhancing variability of informational constraints benefits movement control: a comment on Waddington and Adams (2003)," *British Journal of Sports Medicine*, vol. 38, no. 5, pp. 601–605, 2004.

[83] J. M. Hijmans, J. H. B. Geertzen, P. U. Dijkstra, and K. Postema, "A systematic review of the effects of shoes and other ankle or foot appliances on balance in older people and people with peripheral nervous system disorders," *Gait and Posture*, vol. 25, no. 2, pp. 316–323, 2007.

[84] R. S. W. Masters, J. M. Poolton, B. Abernethy, and N. G. Patil, "Implicit learning of movement skills for surgery," *ANZ Journal of Surgery*, vol. 78, no. 12, pp. 1062–1064, 2008.

[85] G. Waddington, H. Seward, T. Wrigley, N. Lacey, and R. Adams, "Comparing wobble board and jump-landing training effects on knee and ankle movement discrimination," *Journal of Science and Medicine in Sport*, vol. 3, no. 4, pp. 449–459, 2000.

[86] G. Waddington, R. Adams, and A. Jones, "Wobble board (ankle disc) training effects on the discrimination of inversion movements," *Australian Journal of Physiotherapy*, vol. 45, no. 2, pp. 95–101, 1999.

[87] J. Liu, X.-Q. Wang, J.-J. Zheng et al., "Effects of Tai Chi versus proprioception exercise program on neuromuscular function of the ankle in elderly people: a randomized controlled trial," *Evidence-Based Complementary and Alternative Medicine*, vol. 2012, Article ID 265486, 8 pages, 2012.

[88] J. X. Li, D. Q. Xu, and Y. L. Hong, "Effects of 16-week Tai Chi intervention on postural stability and proprioception of knee and ankle in older people," *Age and Ageing*, vol. 37, no. 5, pp. 575–578, 2008.

[89] A. Martínez-Amat, F. Hita-Contreras, R. Lomas-Vega, I. Caballero Martinez, P. J. Alvarez, and E. Martínez-Lopez, "Effects of 12-week proprioception training program on postural stability, gait, and balance in older adults: a controlled clinical trial," *Journal of Strength and Conditioning Research*, vol. 27, no. 8, pp. 2180–2188, 2013.

[90] A. J. Y. Lee and W.-H. Lin, "Twelve-week biomechanical ankle platform system training on postural stability and ankle proprioception in subjects with unilateral functional ankle instability," *Clinical Biomechanics*, vol. 23, no. 8, pp. 1065–1072, 2008.

[91] E. Eils and D. Rosenbaum, "A multi-station proprioceptive exercise program in patients with ankle instability," *Medicine and Science in Sports and Exercise*, vol. 33, no. 12, pp. 1991–1998, 2001.

[92] H. Kiers, S. Brumagne, J. van Dieën, P. van Der Wees, and L. Vanhees, "Ankle proprioception is not targeted by exercises on an unstable surface," *European Journal of Applied Physiology*, vol. 112, no. 4, pp. 1577–1585, 2012.

[93] F. A. Hazime, P. Allard, M. R. Ide, C. M. Siqueira, C. F. Amorim, and C. Tanaka, "Postural control under visual and proprioceptive perturbations during double and single limb stances: insights for balance training," *Journal of Bodywork and Movement Therapies*, vol. 16, no. 2, pp. 224–229, 2012.

[94] M. R. Hinder, T. J. Carroll, and J. J. Summers, "Inter-limb transfer of ballistic motor skill following non-dominant limb training in young and older adults," *Experimental Brain Research*, vol. 227, no. 1, pp. 19–29, 2013.

[95] S. Graziadio, K. Nazarpour, S. Gretenkord, A. Jackson, and J. A. Eyre, "Greater intermanual transfer in the elderly suggests age-related bilateral motor cortex activation is compensatory," *Journal of Motor Behavior*, vol. 47, no. 1, pp. 47–55, 2015.

[96] J. Han, G. Waddington, R. Adams, and J. Anson, "Bimanual proprioceptive performance differs for right- and left-handed individuals," *Neuroscience Letters*, vol. 542, pp. 37–41, 2013.

[97] S. Pajala, P. Era, M. Koskenvuo et al., "Contribution of genetic and environmental factors to individual differences in maximal walking speed with and without second task in older women," *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, vol. 60, no. 10, pp. 1299–1303, 2005.

[98] S. Pajala, P. Era, M. Koskenvuo et al., "Contribution of genetic and environmental effects to postural balance in older female twins," *Journal of Applied Physiology*, vol. 96, no. 1, pp. 308–315, 2004.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 689610, 6 pages
http://dx.doi.org/10.1155/2015/689610

Research Article

Correlation between Trunk Posture and Neck Reposition Sense among Subjects with Forward Head Neck Postures

Han Suk Lee,[1] Hyung Kuk Chung,[2] and Sun Wook Park[3]

[1] *Department of Physical Therapy, Faculty of Health Science, Eulji University, 212 Yangji-dong, Sujeong-gu, Seongnam, Gyeonggi-do 461-713, Republic of Korea*
[2] *Department of Physical Therapy, Ansan University, 155 Ansan University Road, Ansan, Gyeonggi-do 426-701, Republic of Korea*
[3] *Department of Physical Therapy, Samsung Seoul Hospital, 81 Irwon Ro, Gang Nam-gu, Seoul 135-710, Republic of Korea*

Correspondence should be addressed to Han Suk Lee; leehansuk21@hanmail.net

Received 9 April 2015; Revised 30 May 2015; Accepted 2 June 2015

Academic Editor: Thierry Paillard

Objective. To assess the correlation of abnormal trunk postures and reposition sense of subjects with forward head neck posture (FHP). *Methods*. In all, postures of 41 subjects were evaluated and the FHP and trunk posture including shoulder, scapular level, pelvic side, and anterior tilting degrees were analyzed. We used the head repositioning accuracy (HRA) test to evaluate neck position senses of neck flexion, neck extension, neck right and left side flexion, and neck right and left rotation and calculated the root mean square error in trials for each subject. Spearman's rank correlation coefficients and regression analysis were used to assess the degree of correlation between the trunk posture and HRA value, and a significance level of $\alpha = 0.05$ was considered. *Results*. There were significant correlations between the HRA value of right side neck flexion and pelvic side tilt angle ($p < 0.05$). If pelvic side tilting angle increases by 1 degree, right side neck flexion increased by 0.76 degrees ($p = 0.026$). However, there were no significant correlations between other neck motions and trunk postures. *Conclusion*. Verifying pelvic postures should be prioritized when movement is limited due to the vitiation of the proprioceptive sense of neck caused by FHP.

1. Introduction

Forward head neck posture (FHP) is caused by maintaining an abnormal posture for a long time.

This posture shortens the sternoclenoidomastoid (SCM) and scalenus anterior but lengthens the levator scapulae and semispinalis capitis posterior major [1, 2]. Moreover, this posture accelerates neck extensor activity because of upper cervical excessive extension [3, 4]. The activities of the upper and lower trapezius increase as well with this posture [5].

FHP can produce problems related to the proprioception of muscles [6], such as mechanoreceptor function, and alter the sensitivity of spindles of the neck muscles mentioned previously, as well as inducing the loss of kinesthetic acuity of neck motions [7].

Proprioceptive dysfunction [8, 9], reposition sense [7], dizziness [10], coordination [11], balance [12], or others are also affected. In order to solve these issues, various treatments such as craniocervical flexion training [13], eye-head neck coordination exercise [11, 14, 15], mobilization, manipulation [16–18], and cocontraction exercises [19] have been developed.

However, most of these studies only assessed chronic neck pain patients [7–9, 11, 12], not patients with FHP.

Nagai et al. [9] suggested patients with neck pain might have a greater association with total head excursion rather than FHP. This study found that pilots with neck pain had limited motions of the neck but did not have problems related to postures, including FHP or shoulder posture, indicating that neck pain is not always accompanied by FHP.

Therefore, the proprioception of patients with FHP, not those with neck pain, should be studied. Additionally, most of the studies on improving the neck reposition sense focused on the neck itself.

The upper body posture of FHP subjects, similar to that in thoracic kyphosis [20] or rounder shoulders [21, 22], can easily get affected because the neck muscles are anatomically

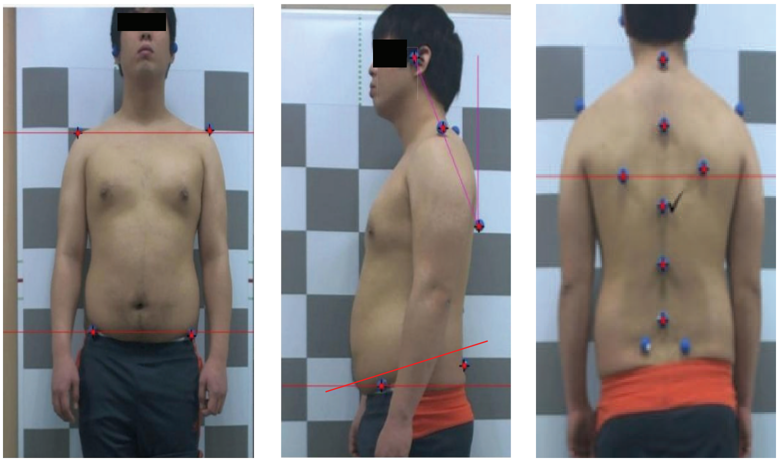


Figure 1: Anterior, lateral, and posterior view.

connected to the trunk. Therefore, if we consider the anatomical orientation of the muscles, there could be problems not only with thoracic kyphosis or rounder shoulder but also with other postures of a trunk such as shoulder or pelvic tilt.

Falla et al. [23] found that FHP associated with prolonged sitting can aggravate neck pain, and a neck pain patient with FHP has reduced ability to maintain an upright posture. Ahn [2] and Murphy et al. [24] suggested that pelvic distortion can cause dysfunctions in the cervical spine, and cervical range of motion improved after the pelvic distortion was corrected. Furthermore, the neck muscle is connected to the trunk muscles with fascia [25]. Therefore, trunk posture should be considered if correction of neck problem is needed.

As mentioned above, a FHP subject possibly has a problem with the repositioning sense of the neck, which can be related with the posture of the trunk owing to the anatomical connection with the neck muscles.

The objective of this study was to identify if accuracy of neck motion is affected by trunk posture and present the baseline data on therapy in the clinic in order to ameliorate repositioning sense of the neck in subjects with FHP.

Therefore, we hypothesized that FHP causes loss of kinesthetic accuracy of neck motions, and this kinesthetic accuracy can be affected by trunk posture, including shoulder tilting angle, scapular level, pelvic side tilting, and anterior tilting angle.

2. Material and Method

2.1. Subjects. From December 2014 to March 2015, 41 subjects (mean age 23.7 ± 2.7 years, mean height 173.6 ± 6.6 cm, mean weight 68.1 ± 10.1 kg, and mean FHP 6.9 ± 2.6 cm) were recruited in the study. The Eulji University approved the study (Grant number EU 14-61), and all subjects were informed of the purpose of this study and provided their written informed consent prior to their participation. This study adhered to the ethical principles of the Declaration of Helsinki. Inclusion criteria were as follows: FHP above 2.5 cm, age 20 to 30 years, no history of concussion or mild neck injury in the previous 12 months, and no other past neurological disorder or fracture.

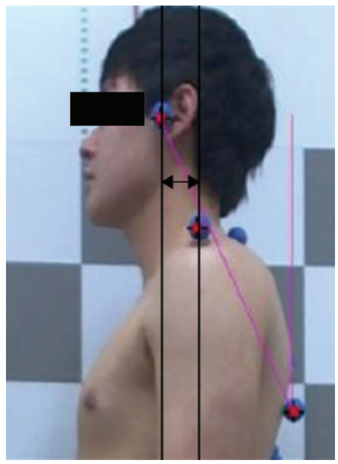


Figure 2: Measurement of FHP.

2.2. Experimental Process. We measured posture using the Body Style S-8.0 (South Korea, LU Commerce) and used the Body Style Analyzer (System Software) to evaluate the posture. We used body markers over each landmark, including the tragus of the ear, the spinous process of the C7 vertebra, acromion, anterior superior iliac spine (ASIS), and the inferior angle of the scapular, posterior superior iliac spine (PSIS), iliac crest, upper thorax, middle thorax, and lower thorax. Then, the subjects stood on the posture pad and photographs of subjects were taken in the lateral, anterior, and posterior views (Figure 1). Data of photography was transferred to the Body Style Analyzer (South Korea, LU Commerce). We analyzed FHP in the lateral view (Figure 2), and the intrarater and interrater evaluations of photogrammetry findings in the standing sagittal posture of the cervical spine were found to be reliable [26, 27]. The distance from the line through acromion to the line through the external auditory meatus was measured for FHP. FHP was calculated using the Body Style Analyzer with markings at the ear tragus and the acromion (Figure 2). If the distance was 2.5–5 cm, it was defined as moderate FHP, and if the distance was >5 cm,

Table 1: The evaluation of HRA of neck motions and trunk posture.

	HRA of NF	HRA of NE	HRA of NRSF	HRA of NLSF	HRA of NRR	HRA of NLR	STA	PSTA	PAT	ScL
M ± SD	9.88 ± 5.46	9.69 ± 4.53	8.70 ± 3.99	10.26 ± 6.23	9.54 ± 3.76	9.80 ± 5.19	1.44 ± 1.05	1.52 ± 1.01	11.76 ± 6.39	2.37 ± 1.93

M: mean, SD: standard deviation, and unit: degree.
HRA: head repositioning accuracy, NF: neck flexion, NE: neck extension, NRSF: neck right side flexion, NLSF: neck left side flexion, NRR: neck right rotation, NLR: neck left rotation, STA: shoulder tilting angle, PAT: pelvic anterior tilting, PSTA: pelvic side tilting angle, and ScL: scapular level.

it was defined as severe FHP [28]. We recruited subjects with FHP >2.5 cm.

The trunk posture, including shoulder tilting, scapular level, pelvic tilting, and anterior tilting degrees in anterior, posterior, and lateral views, was obtained using the Body Style Analyzer.

For neck reposition sense testing, we used the head repositioning accuracy (HRA) test [7, 9, 29] because the head-to-neutral test has been reported to be more sensitive than the head-to-target test [30]. First, subjects were made to sit on a wooden chair with hips and knees at approximately 90° flexion and feet hip-width apart. The HRA test was performed to measure differences in measurements between the reference positions (position 0) and return positions. Equipment with a laser (Figure 3) was firmly placed on the subjects' heads. With their head in a natural resting position, the subjects were requested to focus on a target that was positioned at the eye level. All subjects were then instructed to close their eyes with a sleep shade and were instructed to memorize this position because this was the reference position. Then, they performed a full neck flexion at their preferred speed and held this position for 5 s. After this, the subjects were instructed to return to the reference position with their preferred speed. The stopping point of the laser beam was marked with a dot that was the return position.

The projection point on the abscissa and ordinate axes were measured (X, Y), and each coordinate was given a positive or negative value according to its position relative to the corresponding axis. Using these 2 values, the subject's HRA was then calculated trigonometrically. This measurement represented the direct distance between the points (the return position) on which the light beam stopped on the target to point 0 (the reference) of the target. For comparison of the absolute values for the horizontal values for the horizontal (X) and vertical (Y) components of the repositioning error, the negative signs were removed by calculating the RMS values [31]. Three repetitions of HRA to the reference position were performed and then the mean value of the trails was calculated. The same procedure was followed for extension, rotation, and side flexion, which were randomly performed (Figure 3).

2.3. Statistical Analysis. All statistical analyses were performed using IBM SPSS Statistics (version 20.0, IBM Corporation, South Korea). Descriptive statistics (mean and standard deviations) were calculated for each variable. Spearman's rank correlation coefficients and regression analysis were used to assess the degree of correlation between postural evaluation items and the value of each joint reposition sense, and the significance level of $\alpha = 0.05$ was considered.

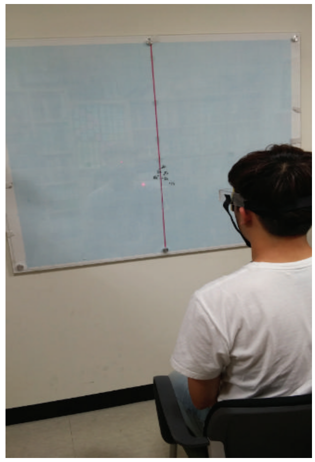


Figure 3: Measurement of head reposition accuracy.

The root mean square error (RMSE) among the trials for each subject was defined by the following equations [32]:

$$E^2 = x^2 + y^2,$$
$$\mathrm{RMSE}^2 = \frac{1}{m}\sum_{i=1}^{m} E_i^2. \tag{1}$$

E denoted the differences between the initial reference position (x) and the final position (y) when repositioning from either flexed, side flexed, extended, or rotated neck position, and m denoted the trial number.

3. Results

The mean HRA values of neck flexion and extension were 9.88 and 9.68, respectively. The mean HRA values of right and left side neck flexion were 8.70 and 10.26, respectively. The mean HRA values of right and left neck rotation were 9.54 and 9.80, respectively. The mean shoulder tilting angle was 1.44 degrees, and pelvic side tilting angle was 1.52 degrees. The pelvic anterior tilting angle was 11.76 degrees, and the scapular level angle was 2.37 degrees (Table 1).

There were significant correlations between the HRA of right side neck flexion and pelvic side tilt angle ($p < 0.05$). If pelvic side tilting angle increases by 1 degree, right side neck flexion increased by 0.76 degrees ($p = 0.026$) (Table 2). However, there was no significant correlation between other HRA values including those for neck flexion, extension, right side flexion, right rotation and left rotation, and trunk posture ($p > 0.05$) (Table 3).

Table 2: Spearman's rank correlation coefficient between HRA of neck motions and trunk posture.

	HRA of NF	HRA of NE	HRA of NRSF	HRA of NLSF	HRA of NRR	HRA of NLR	STA	PSTA	PAT	ScL
HRA of NF	1.000	.020	−.018	.186	.118	.068	−.167	.173	.171	−.049
HRA of NE	.020	1.000	−.008	.078	.378*	.166	.224	−.111	−.003	−.213
HRA of NRSF	−.018	−.008	1.000	.410**	.016	−.035	.132	.376*	.000	.045
HRA of NLSF	.186	.078	.410**	1.000	.178	.121	.117	.243	−.188	.090
HRA of NRR	.118	.378*	.016	.178	1.000	.290	.207	−.029	.016	−.072
HRA of NLR	.068	.166	−.035	.121	.290	1.000	−.069	−.146	−.071	−.100
STA	−.167	.224	.132	.117	.207	−.069	1.000	.157	−.185	.266
PST	.173	−.111	.376*	.243	−.029	−.146	.157	1.000	.169	.023
PAT	.171	−.003	.000	−.188	.016	−.071	−.185	.169	1.000	−.066
SL	−.049	−.213	.045	.090	−.072	−.100	.266	.023	−.066	1.000

HRA: head repositioning accuracy, unit: degree.
NF: neck flexion, NE: neck extension, NRSF: neck right side flexion, NLSF: neck left side flexion, NRR: neck right rotation, NLR: neck left rotation, STA: shoulder tilting angle, PAT: pelvic anterior tilting, PSTA: pelvic side tilting angle, and ScL: scapular level.
$^{*}p < 0.05$, $^{**}p < 0.01$.

Table 3: Regression analysis of HRA of right side flexion and pelvic side tilting angle.

Predictor variable	B	β	T
HRA of right side flexion	0.760	0.347	2.30*

$^{*}p < 0.05$.

4. Discussion

FHP is affected by stress and incorrect postures. Owing to industrial development, the population of subjects with FHP has been increasing [28]. In particular, workers who use computers in their offices are likely at risk for FHP.

FHP can cause problems with proprioception of the neck muscles, and, therefore, a treatment plan is necessary for such patients. Moreover, proprioception can improve with direct treatment of neck muscle or with indirect treatment of the trunk posture, including treatment of the pelvic posture [2, 24, 33]. However, research on the indirect method is limited. In addition, further research is needed on the relationship between proprioception and trunk posture, before an indirect method can be developed.

Therefore, this study explored the correlation between HRA and trunk posture in 41 subjects with FHP to determine the relationship between proprioception of neck and trunk posture. We found a significant correlation between the right side flexion reposition sense of the neck and the pelvic side tilting angle.

Black et al. [33] found that a change in lumbar posture was associated with a compensatory change in cervical position. Murphy et al. [24] used manipulative therapy on the cervical spine to relieve low back pain. Nansel et al. [34] found that cervical spine manipulation has significant effects on the tone of the lumbopelvic musculature, particularly in the gluteal region, and Hyoung et al. [35] found that increasing cervical motion after ankle joint therapy is helpful.

Corrective exercises for FHP had a positive effect on spinal posture in patients with lumbosacral radiculopathy [36] or adolescent idiopathic scoliosis [37].

According to earlier studies, function of cervical improved after the patient received therapy for trunk and ankle region. In this study, we found similar results with those of previous studies that lumbar posture is related to cervical motion.

However, the mechanism underlying how treatment on the cervical region affects the pelvic area is unknown.

Nansel et al. [34] and Murphy et al. [24] suggested this effect of treatment may be because of its effect on the tonic neck reflex (TNR). TNR alters the tone of the trunk and extremities in two ways. One is via afferents from muscle spindles to the vestibular nucleus and the pontine and medullary reticular formation. The other is via signals from the upper cervical afferents sent to propriospinal neurons.

Therefore, if cervical dysfunction is corrected, the tone normalizes with normal patterns of the TNR, and pelvic distortion will improve.

Ahn [2] and Hyoung et al. [35] explained the treatment effect using the mechanical chain of joint. They assumed that the entire body is connected in a chain that affects each segment.

In our opinion, this relationship may be explained with two reasons: the fascia and functional structure of quadratus lumborum (QL) and scalenus muscle.

With respect to the fascia, the agonist muscles of neck side flexion and the pelvic side tilting angle are connected via the fascia on the lateral line [25].

The QL is not directly part of the lateral line according to anatomy trains' rule. However, with respect to the functional structure, the QL uniquely works as a lateral flexor of the trunk and the scalenus works as a lateral flexor of neck, similar to the QL. The QL pulls from one end of the rib cage and the scalenus from the other end. Therefore, the two muscles are very close related to functional structure. If the scalenus pulls the ribcage, it affects the QL as well, thus affecting pelvic posture. Thus, if the pelvic posture is fixed well, the proprioception of the muscles that affects movements of the neck can be refined well.

Nejati et al. [22] suggested that shoulder posture was not correlated with neck pain. However, Szeto et al. [38]

found that subjects with neck and shoulder discomfort had protracted acromion posture. Lau et al. [39] specified that the sagittal posture of the thoracic spine had a very close relationship with neck pain severity and disability and suggested that thoracic posture correction would help prevent neck pain. Lynch et al. [40] found significant interactions between movement of forward head translation and those of forward shoulder translation. The results did not concur with those previously described.

The conflicting result was probably because of the lack of a gold standard in clinical measurement of posture. The methods of measurement were different in each study. Nejati et al. [22], Szeto et al. [38], and Lynch et al. [40] studied shoulder kyphosis and protraction, but Lau et al. researched the upper thoracic angle. The measurements of posture also differed among studies, indicating that there is no gold standard for the clinical measurement of posture to reflect the actual curvatures of the spine.

We could not find a correlation between shoulder posture and any neck motion, which is similar to the result of Nejati et al.'s study [22]. However, the research methodology was not the same as ours. Therefore, additional studies are needed in the future, involving the same clinical measurement. Moreover, the standard measurement of posture should be developed that can be used easily in the clinic.

Consequently, in the future, to improve the reposition sense of subjects with FHP, first, pelvic posture should be checked, and then that of its related muscles should be verified using electromyography.

The limitation of this study was that it was difficult to generalize the results owing to small sample size. In addition, in the study, we could not determine the mechanism of relationship between neck reposition sense and trunk posture. So, we will be performing an electromyography study and other equipment to confirm the mechanism in the future.

5. Conclusion

In conclusion, in the case of FHP subjects, the higher the pelvis side tilting angle, the worse the HRA value of lateral neck flexion. This might be the anatomical structure of the muscles around pelvic area and neck with fascia and functional structure of QL and scalenus muscle and the mechanism of that result should be studied in the future. Therefore, verifying pelvic posture should be prioritized when movement is limited due to vitiation of the proprioceptive sense of the neck owing to FHP.

Conflict of Interests

Han Suk Lee declares that there is no conflict of interests regarding the publication of this paper. Hyung Kuk Chung declares that there is no conflict of interests regarding the publication of this paper. Sun Wook Park declares that there is no conflict of interests regarding the publication of this paper.

References

[1] D. A. Neuman, *Kinesiology of the Musculoskeletal System*, Mosby, Singapore, 2nd edition, 2009.

[2] M. Ahn, *The Effect of Postural Correction on Pelvis Balance and Cervical Range of Motion*, Deagu University, Gyeongsan, Republic of Korea, 2004.

[3] L. Straker and K. Mekhora, "An evaluation of visual display unit placement by electromyography, posture, discomfort and preference," *International Journal of Industrial Ergonomics*, vol. 26, no. 3, pp. 389–398, 2000.

[4] S. J. Edmondston, M. Sharp, A. Symes, N. Alhabib, and G. T. Allison, "Changes in mechanical load and extensor muscle activity in the cervico-thoracic spine induced by sitting posture modification," *Ergonomics*, vol. 54, no. 2, pp. 179–186, 2011.

[5] J.-H. Weon, J.-S. Oh, H.-S. Cynn, Y.-W. Kim, O.-Y. Kwon, and C.-H. Yi, "Influence of forward head posture on scapular upward rotators during isometric shoulder flexion," *Journal of Bodywork and Movement Therapies*, vol. 14, no. 4, pp. 367–374, 2010.

[6] M.-Y. Lee, H. Y. Lee, and M.-S. Yong, "Characteristics of cervical position sense in subjects with forward head posture," *Journal of Physical Therapy Science*, vol. 26, no. 11, pp. 1741–1743, 2014.

[7] C.-H. Cheng, J.-L. Wang, J.-J. Lin, S.-F. Wang, and K.-H. Lin, "Position accuracy and electromyographic responses during head reposition in young adults with chronic neck pain," *Journal of Electromyography and Kinesiology*, vol. 20, no. 5, pp. 1014–1020, 2010.

[8] H. V. Heikkilä and B.-I. Wenngren, "Cervicocephalic kinesthetic sensibility, active range of cervical motion, and oculomotor function in patients with whiplash injury," *Archives of Physical Medicine and Rehabilitation*, vol. 79, no. 9, pp. 1089–1094, 1998.

[9] T. Nagai, J. P. Abt, T. C. Sell et al., "Neck proprioception, strength, flexibility, and posture in pilots with and without neck pain history," *Aviation Space and Environmental Medicine*, vol. 85, no. 5, pp. 529–535, 2014.

[10] J. Treleaven, G. Jull, and M. Sterling, "Dizziness and unsteadiness following whiplash injury: characteristic features and relationship with cervical joint position error," *Journal of Rehabilitation Medicine*, vol. 35, no. 1, pp. 36–43, 2003.

[11] U. Röijezon, M. Björklund, M. Bergenheim, and M. Djupsjöbacka, "A novel method for neck coordination exercise—a pilot study on persons with chronic non-specific neck pain," *Journal of Neuroengineering and Rehabilitation*, vol. 5, no. 1, p. 36, 2008.

[12] P. Michaelson, M. Michaelson, S. Jaric, M. L. Latash, P. Sjölander, and M. Djupsjöbacka, "Vertical posture and head stability in patients with chronic neck pain," *Journal of Rehabilitation Medicine*, vol. 35, no. 5, pp. 229–235, 2003.

[13] G. Jull, D. Falla, J. Treleaven, P. Hodges, and B. Vicenzino, "Retraining cervical joint position sense: the effect of two exercise regimes," *Journal of Orthopaedic Research*, vol. 25, no. 3, pp. 404–412, 2007.

[14] B. K. Humphreys and P. M. Irgens, "The effect of a rehabilitation exercise program on head repositioning accuracy and reported levels of pain in chronic neck pain subjects," *Journal of Whiplash & Related Disorders*, vol. 1, no. 1, pp. 99–112, 2002.

[15] C. M. Petersen, C. L. Zimmermann, and R. Tang, "Proprioception interventions to improve cervical position sense in cervical pathology," *International Journal of Therapy and Rehabilitation*, vol. 20, no. 3, pp. 154–163, 2013.

[16] A. Gross, J. Miller, J. D'Sylva et al., "Manipulation or mobilisation for neck pain: a Cochrane Review," *Manual Therapy*, vol. 15, no. 4, pp. 315–333, 2010.

[17] R. G. Rogers, "The effects of spinal manipulation on cervical kinesthesia in patients with chronic neck pain: a pilot study," *Journal of Manipulative and Physiological Therapeutics*, vol. 20, no. 2, pp. 80–85, 1997.

[18] P. J. Palmgren, P. J. Sandström, F. J. Lundqvist, and H. Heikkilä, "Improvement after chiropractic care in cervicocephalic kinesthetic sensibility and subjective pain intensity in patients with nontraumatic chronic neck pain," *Journal of Manipulative and Physiological Therapeutics*, vol. 29, no. 2, pp. 100–106, 2006.

[19] B. Armstrong, P. McNair, and D. Taylor, "Head and neck position sense," *Sports Medicine*, vol. 38, no. 2, pp. 101–117, 2008.

[20] J. Quek, Y.-H. Pua, R. A. Clark, and A. L. Bryant, "Effects of thoracic kyphosis and forward head posture on cervical range of motion in older adults," *Manual Therapy*, vol. 18, no. 1, pp. 65–71, 2013.

[21] B. Braun and L. Amundson, "Quantitative assessment of head and shoulder posture," *Archives of Physical Medicine and Rehabilitation*, vol. 70, no. 4, pp. 322–329, 1989.

[22] P. Nejati, S. Lotfian, A. Moezy, A. Moezy, and M. Nejati, "The relationship of forward head posture and rounded shoulders with neck pain in Iranian office workers," *Medical Journal of the Islamic Republic of Iran*, vol. 28, p. 26, 2014.

[23] D. Falla, G. Jull, T. Russell, B. Vicenzino, and P. Hodges, "Effect of neck exercise on sitting posture in patients with chronic neck pain," *Physical Therapy*, vol. 87, no. 4, pp. 408–417, 2007.

[24] D. R. Murphy, B. M. Carr, and R. J. Tyszkowski, "A possible cervical cause of low back pain: pelvic distortion," *Journal of Bodywork and Movement Therapies*, vol. 4, no. 2, pp. 83–89, 2000.

[25] W. Tomas and Myers, *Anatomy Train: Myofascial Merdians for Manual & Movement Therapist*, Churchill Livingstone, Beijing, China, 2014.

[26] R. M. Ruivo, P. Pezarat-Correia, and A. I. Carita, "Intrarater and interrater reliability of photographic measurement of upper-body standing posture of adolescents," *Journal of Manipulative and Physiological Therapeutics*, vol. 38, no. 1, pp. 74–80, 2015.

[27] J. L. do Rosario, "Photographic analysis of human posture: a literature review," *Journal of Bodywork and Movement Therapies*, vol. 18, no. 1, pp. 56–61, 2014.

[28] Y. J. Kim, *The effect of ballet program on turtle neck syndrome in office workers [M.S. thesis]*, Graduate School Hanyang University, Seoul, Republic of Korea, 2012.

[29] E. F. Owens Jr., C. N. R. Henderson, M. R. Gudavalli, and J. G. Pickar, "Head repositioning errors in normal student volunteers: a possible tool to assess the neck's neuromuscular system," *Chiropractic & Osteopathy*, vol. 14, no. 1, article 5, 7 pages, 2006.

[30] E. Kristjansson, P. Dall'Alba, and G. Jull, "A study of five cervicopephalic relocation tests in three different subject groups," *Clinical Rehabilitation*, vol. 17, no. 7, pp. 768–774, 2003.

[31] G. D. Rix and J. Bagust, "Cervicocephalic kinesthetic sensibility in patients with chronic, nontraumatic cervical spine pain," *Archives of Physical Medicine and Rehabilitation*, vol. 82, no. 7, pp. 911–919, 2001.

[32] C. C. Teng, H. Chai, D.-M. Lai, and S.-F. Wang, "Cervicocephalic kinesthetic sensibility in young and middle-aged adults with or without a history of mild neck pain," *Manual Therapy*, vol. 12, no. 1, pp. 22–28, 2007.

[33] K. M. Black, P. McClure, and M. Polansky, "The influence of different sitting positions on cervical and lumbar posture," *Spine*, vol. 21, no. 1, pp. 65–70, 1996.

[34] D. D. Nansel, T. Waldorf, and R. Cooperstein, "Effect of cervical spinal adjustments on lumbar paraspinal muscle tone: evidence for facilitation of intersegmental tonic neck reflexes," *Journal of Manipulative and Physiological Therapeutics*, vol. 16, no. 2, pp. 91–95, 1993.

[35] I. H. Hyoung, M. Ahn, S. S. Bae, S. E. Mun, and J. B. Chae, "The effects of GCM ankle joint therapy on shoulder and neck," *The Journal of Korean Society of Physical Therapy*, vol. 15, no. 4, pp. 479–487, 2003.

[36] I. M. Moustafa and A. A. Diab, "The effect of adding forward head posture corrective exercises in the management of lumbosacral radiculopathy: a randomized controlled study," *Journal of Manipulative and Physiological Therapeutics*, vol. 38, no. 3, pp. 167–178, 2015.

[37] A. A. Diab, "The role of forward head correction in management of adolescent idiopathic scoliotic patients: a randomized controlled trial," *Clinical rehabilitation*, vol. 26, no. 12, pp. 1123–1132, 2012.

[38] G. P. Y. Szeto, L. Straker, and S. Raine, "A field comparison of neck and shoulder postures in symptomatic and asymptomatic office workers," *Applied Ergonomics*, vol. 33, no. 1, pp. 75–84, 2002.

[39] K. T. Lau, K. Y. Cheung, K. B. Chan, M. H. Chan, K. Y. Lo, and T. T. Wing Chiu, "Relationships between sagittal postures of thoracic and cervical spine, presence of neck pain, neck pain severity and disability," *Manual Therapy*, vol. 15, no. 5, pp. 457–462, 2010.

[40] S. S. Lynch, C. A. Thigpen, J. P. Mihalik, W. E. Prentice, and D. Padua, "The effects of an exercise intervention on forward head and rounded shoulder postures in elite swimmers," *British Journal of Sports Medicine*, vol. 44, no. 5, pp. 376–381, 2010.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 716042, 8 pages
http://dx.doi.org/10.1155/2015/716042

Research Article

Relationships of Balance, Gait Performance, and Functional Outcome in Chronic Stroke Patients: A Comparison of Left and Right Lesions

Priscila Garcia Lopes,[1] José Augusto Fernandes Lopes,[1] Christina Moran Brito,[1] Fábio Marcon Alfieri,[1,2] and Linamara Rizzo Battistella[1,3]

[1] *Clinical Research Center, Institute of Physical Medicine and Rehabilitation, University of São Paulo School of Medicine, Rua Domingo de Soto 100, Vila Mariana, 04116-030 São Paulo, SP, Brazil*
[2] *Health Promotion and Physical Therapy Faculty, São Paulo Adventist University Center, São Paulo, Brazil*
[3] *School of Medicine, University of São Paulo, São Paulo, Brazil*

Correspondence should be addressed to Fábio Marcon Alfieri; fabiomarcon@bol.com.br

Received 4 March 2015; Revised 28 May 2015; Accepted 25 June 2015

Academic Editor: Massimiliano Pau

Introduction. This study compared the balance by center of pressure (COP) and its relationship with gait parameters and functional independence in left (LH) and right (RH) chronic stroke patients. *Methods*. In this cross-sectional study, twenty-one hemiparetic stroke patients were assessed for Functional Independence Measure (FIM), balance with a force platform, and gait in the Motion Analysis Laboratory. *Results*. The amplitudes of the COP in the anteroposterior and mediolateral directions were similar in both groups. The anteroposterior direction was greater than the mediolateral direction. Only the temporal parameters showed any statistically significant differences. The LH showed a significant correlation between stride length, step length, and gait velocity with COP velocity sway for the healthy and paretic lower limbs. In both groups, the area of COP was significantly correlated with stride length. Motor FIM was significantly correlated with the COP in the LH group. *Conclusion*. There was no difference in the performance of balance, gait, and functional independence between groups. The correlation of the COP sway area with stride length in both groups can serve as a guideline in the rehabilitation of these patients where training the static balance may reflect the improvement of the stride length.

1. Introduction

Stroke is the leading cause of disability in adults. Forty percent of stroke patients exhibit moderate functional impairment, and 15% to 30% exhibit severe disabilities [1]. Although intensive rehabilitation, evaluation of the balance, gait, and functional independence are offered to many patients within six months of a stroke, many of them continue to have motor deficits [2]. Adequate therapy increases chronic patient survival, [3] emphasizing the importance of evaluating the overall motor profile following the initial recovery period. Therefore, research in the chronic phase of the stroke is also important.

Postural instability is a common finding and is cited as the leading cause of falls and limited functional independence in stroke patients [4–6]. Posture or balance deficits are common mainly because the unaffected limb bears a greater proportion of the body weight [6–10]. In hemiparetic patients, postural oscillation while standing upright is characterized by an asymmetric profile with larger oscillations on the paretic side than the nonparetic side and low temporal synchronization between oscillations of the lower limbs and the pelvis and between the lower limbs. Difficulty in stabilizing the pelvis and the distal segments of the lower limb on the affected side are reflected in the increase in postural oscillation of hemiparetic patients [11].

Evidence of the differences between the functional consequences of strokes in the left and right hemispheres is particularly interesting. The left hemisphere is more important for motor control, while the right hemisphere is more important for spatial orientation [12]. Motor activities requiring planning and coordination are more dependent on the left hemisphere and are strongly affected in individuals with right side hemiparesis [13, 14]. Right hemisphere lesions are more likely to result in deficits in attention and contralateral perception [15] and stabilization of the position in relation to lesion of left side [14]. The right hemisphere integrates sensorimotor information which is critical for maintaining posture and maintaining sitting or standing positions [15].

Left hemiparetic patients exhibit poorer postural balance in sitting and standing positions compared to right hemiparetic patients, and patients who have not received adequate therapy exhibit high degrees of postural alterations [4]. Some studies have assessed balance by analyzing postural instability during standing in hemiparetic stroke patients [5, 6, 8, 9, 16–19]. These studies were performed mainly in the first year after stroke. Postural instability is assessed through center of pressure (COP) sway analysis, and COP position can be assessed directly using force platforms during the evaluation of posture and gait.

Some studies have reported higher anterior-posterior oscillation than mediolateral oscillation in static COP assessments [20–22]. Rode et al. [8] compared postural oscillation in 15 right and 15 left hemiparetic patients and found that the latter group exhibited larger areas of oscillation and mediolateral displacement. Other studies have attempted to correlate static balance data with gait parameters [20, 23].

The correlations of balance and gait parameters are important for the assessment and rehabilitation of patients because a reliable correlation could mean that resources used to improve balance could also influence gait. In clinical practice, it is clear that delay in therapy leads to poorer postural control among the left hemiparetic patients, but after one year has passed and rehabilitation is finished, monitoring and comparing with right side hemiparetics is difficult. It would be interesting to discover the possible mechanisms involved in the control of posture, the regulation of skeletal muscle during gait, and the oscillations of the COP that maintain corporal stability in hemiparetic patients. As the stroke sufferer's balance is impaired and can lead to consequences such as falls, knowing the questions related to balance and gait will be important in order for these two physical capabilities to be better understood in hemiparetic subjects. It is also believed that this will lead to a better direction regarding the rehabilitation of these patients.

We hypothesized that the left hemiparetic patients still have poorer balance in the chronic phase as well as in early stage [4]. Therefore the purpose of this study was to compare the balance by COP sway and its relationship with gait parameters and functional independence in left or right chronic stroke patients.

2. Methods

This cross-sectional study was conducted at the Institute of Physical Medicine and Rehabilitation at the Clinical Hospital of the Medical School of the University of São Paulo (IMREA-HCFMUSP). Twenty-seven hemiparetic chronic stroke patients treated at the hemiplegia outpatient clinic were invited to participate in the study. Selected patients were informed of the study's aims and procedures, and they or their caretakers signed an informed consent form. This research project was approved by the Committee on Ethics and Research (CAPPesq) under protocol number 0280/09.

The following inclusion criteria were used: hemiparesis resulting from a stroke that occurred at least 12 months prior to the study; age between 45 and 65 years; pattern of hemiparesis featuring brachial predominance; ability to walk 10 meters unassisted; ability to remain standing upright for 60 seconds unassisted; and right side dominance (right-handed individuals). Exclusion criteria were as follows: cognitive impairments affecting comprehension and disabilities arising from other conditions, such as deformity or pain.

2.1. Clinical Assessment. Patient clinical records containing personal data, clinical diagnosis, time of the lesion, and Functional Independence Measure FIM [24] score were evaluated. The FIM quantitatively evaluates the care demanded by a person to perform a series of motor and cognitive tasks of daily living. Among the activities evaluated are self-care, transfers, locomotion, sphincter control, communication, and social cognition—including memory, social interaction, and problem solving. Each of these activities is evaluated and receives a score ranging from one (total dependence) to seven (complete independence) and the total score ranges from 18 to 126. Two FIM areas describe the motor score ranging from 13 to 91 points and cognitive score ranging from 5 to 35 points [24].

First, the patient's right or left dominance was assessed by asking about the dominant upper limb (writing hand). The physical assessment involved tests of muscle strength and tone. Muscle strength was tested according to the Kendall scale [25] and spasticity was tested via the modified Ashworth scale [26, 27]. Quadriceps strength was chosen as the representative assessment for the lower limb, and brachial biceps strength was chosen for the upper limb. Spasticity was measured in the gastrocnemius muscle because this muscle is important for the ankle strategy in maintaining postural control [21].

2.2. Postural Control Assessment. The postural control assessment was performed with patients standing upright. Patients remained standing with their arms hanging alongside their body, eyes fixed on a point on the wall, and feet set on a force platform (AMTI OR6-7 version 2.0/2004, installed at the Motion Analysis Laboratory). McIlroy and Maki [28] showed that foot separation in a preferred stance position is correlated to subject height and is considerably larger than that usually standardized for posturography suggesting a standard position with 17 cm separation or 11 percent of subject height would avoid uncomfortable foot positions. To accommodate the increased instability expected in chronic stroke patients while still taking anthropometric variations into consideration a foot separation equal to the length of the patient's feet was adopted as measured between the midpoints of both heels. The feet were positioned parallel to each other on the platform. After calibration, COP position was recorded by the force platform.

Patients were asked to hold the position for 60 seconds, and the assessment was repeated three times. The data were acquired at 100 Hz and subsampled at 10 Hz as described by Raymakers et al. [29]. To avoid disturbances from the initial stabilization of the subject, the first 10 seconds of each record were discarded. The center of pressure (COP) sway variables measured were anterior-posterior and mediolateral oscillation amplitude, COP sway area, and average velocity [29]. Sway amplitude was calculated as the difference between the maximum and minimum coordinates of the COP in each direction and was expressed in centimeters (cm); the rectangular area that covered the whole COP trajectory was calculated by multiplying anterior-posterior and mediolateral amplitudes and was expressed in cm^2. Average sway velocity was calculated by dividing the total length of the COP trajectory by the duration of the recording and expressed in centimeters per second (cm/s). A therapist remained by the patient's side during the procedures for safety in case the patient lost his or her balance. Posturography (postural control) data were recorded by the EVaRT 5.0 software (Motion Analysis Corporation) and we used a routine developed and processed in the Matlab 2008a software (Mathworks) to process the data.

2.3. Gait Assessment. To obtain spatial and temporal gait parameters, reflective markers were placed on the patient's heels on the lateral and medial malleoli between the first and second metatarsals and on the sacrum. Considering that joint angles would not be analyzed, this is the minimum subset of the modified Helen Hayes market set [30] that allows the calculation of the desired gait parameters using specialized computer software (Orthotrak 6.2, Motion Analysis Corporation, Santa Rosa, CA, USA). Patients were then asked to walk along a preestablished path measuring 10 meters in length in the Motion Analysis Laboratory. The gait parameters considered were step and stride lengths, gait velocity, and gait cadence. Step and stride lengths were expressed in centimeters, velocity in centimeters per second, and cadence as the number of steps per minute. Temporal parameters considered as percentages of the gait cycle time were also measured: stance phase is the whole period of time when the foot is in contact with the ground; swing phase is the period when the foot is not in contact with the ground; double stance onset is the period when both feet are in contact with the floor at the beginning of the gait cycle; single stance is defined as the percentage of time of the gait cycle when only one foot is in contact with the ground. The data from the markers were captured by eight Hawk System digital cameras operating at 100 Hz using software provided by their manufacturer (EVaRT 5.0, Motion Analysis Corporation, Santa Rosa, CA).

2.4. Data Analysis. All data were distributed normally, according to the Kolmogorov-Smirnov test, with $p < 0.05$. Parametric tests, such as Student's t-test, were used to compare means to analyze static postural control. The level of significance was set at $p \leq 0.05$. The COP velocity and area of oscillation for left and right hemiparesis groups were compared. Gait data were also compared between groups using Student's t-test. Finally, possible correlations between the gait parameters and static COP sway, velocity, and area were evaluated using Pearson's correlation coefficient, where 0 to 0.30 was interpreted as a weak correlation, 0.30 to 0.70 as a moderate correlation, and over 0.70 as a strong correlation. The static COP sway measured (mean velocity and area of oscillation) was evaluated for correlation with gait data and with the FIM.

3. Results

Of the 27 patients contacted from the IMREA-HCFMUSP, six did not complete evaluations; thus 21 participants remained: nine in the right hemiparetic group (RH) group and 12 in the left hemiparetic group (LH). Table 1 describes the characteristics regarding mean age, time since injury, Functional Independence Measure (FIM), muscle strength, and spasticity of all patients and of each group. The groups are similar, except for the motor FIM, which had a higher score for the RH.

In assessing the extent of COP sway in the anterior-posterior (RH—3.0 ± 1.4 cm and LH—3.1 ± 1.2) and mediolateral directions (RH—1.7 ± 1.2 and LH—1.5 ± 0.5 cm) and the average of the velocity (RH—1.9 ± 1.1 and LH—1.6 ± 0.8 cm/s) it was found that the values were similar ($p > 0.05$) in the right and left sides.

Between groups, only the temporal parameters showed any statistically significant differences. All patients spent more time in the stance phase for the healthy lower limb; specifically the single stance time was significantly different in both the left ($p = 0.0004$) and the right ($p = 0.001$) hemiparesis groups.

In general, the groups showed that the total stance time was longer for the healthy lower limb than for the affected limb; as for the analyzed gait variables (stance, swing, and single stance), there were statistically significant differences between healthy and affected limbs.

There were no statistically significant differences between the affected and healthy sides in either group of patients for the remaining parameters, such as length of step and stride, gait velocity, initial stance, and cadence. The step length of the affected limb was longer than that of the healthy limb in most of the patients in both groups. Meanwhile, the stride length of the healthy limb was greater than that of the affected limb in both groups.

In both groups the mean velocity and cadence were similar and had low standard deviations; there were no statistically significant differences in these gait parameters between patient groups. The LH had a mean velocity of 40.7 (±3.6) cm/s and a mean cadence of 74 (±0.99) steps/minute; the RH had a mean velocity of 40 (±1.5) cm/s and a cadence of 77.6 (±3.6) steps/minute. Higher values were recorded for the healthy lower limb in most patients (LH, $n = 6$; and RH, $n = 3$).

Correlations between the gait parameters and static COP velocity were also analyzed (Table 2). The left hemiparetic group showed a significant ($p < 0.05$) correlation

Table 1: Characteristics of the studied patients: motor and total FIM, muscle strength, and spasticity.

	Total (n = 21)	RH (n = 9)	LH (n = 12)	p
Male/female	15/6	7/2	8/4	0.65
Mean age (years) (SD)	55.3 (±5.9)	54.2 (±2.8)	56.4 (±7.4)	0.41
Time since lesion (months)	32.6 (±17.7)	37.4 (±19.9)	29 (±15.7)	0.29
BMI (Kg/m^2)	30.1 (±5.6)	29.5 (±6.8)	30.60 (±5)	0.69
Weight (kg)	73.2 (±10.9)	75.3 (±11.7)	71.4 (±11.6)	0.26
Height (m)	1.5 (±9.36)	1.6 (±9.5)	1.5 (±8.9)	0.20
FIM	Motor	81 (±5.0)	73.7 (±8.7)	0.03
	Total	107 (±16.0)	107 (±8.7)	0.32
Muscle strength	Biceps brachial	3.0 (±0.9)	2.4 (±1.7)	0.4
	Quadriceps	3.8 (±0.4)	3.9 (±0.7)	0.9
Spasticity	Biceps brachial	1.2 (±0.7)	1.0 (±0.7)	0.3
	Gastrocnemius	1.1 (±0.7)	0.8 (±0.7)	0.16

Note: FIM: functional independence measure; RH: right hemiparetic group; LH: left hemiparetic group; SD: standard deviation.

Table 2: Correlation between velocity of COP sway and gait parameters.

Gait parameters	LH		RH	
	Healthy limb	Affected limb	Healthy limb	Affected limb
Step length	−0.60*	−0.62*	0.38	0.36
Stride length	−0.72*	−0.72*	0.46	0.41
Gait velocity	−0.68*	−0.70*	−0.05	−0.19
Stance phase	0.76*	0.27	0.47	−0.01
Swing phase	−0.76*	−0.27	−0.47	0.01
Double stance onset	0.70*	0.53	0.07	0.61
Single stance	−0.27	−0.76*	0.01	−0.47
Cadence	−0.46	−0.45	−0.58	−0.62

*Pearson's correlation coefficient (r) with $p < 0.05$; RH: right hemiparetic group; LH: left hemiparetic group.

between stride length, step length, and gait velocity with COP velocity sway for the healthy and paretic lower limbs. In the healthy limb, the parameters like swing and stance were significantly correlated with the COP velocity.

Table 3 shows the correlation coefficients between the area of COP, sway, and gait parameters. In both groups the area of COP was significantly correlated with stride length. In the right group the step length was significantly correlated with the area of COP but only in the healthy limb. By contrast, these variables were moderately correlated in both limbs of left hemiparetic patients. In this group gait velocity and COP area were significantly correlated. The area of COP in the healthy limb was significantly correlated with durations of the stance and swing phases and in the affected limb it was significantly correlated with the single stance duration and double stance onset.

The correlations between the FIM and static balance data were also analyzed. Both groups had their area of COP sway moderately correlated with motor and the FIM total. Motor FIM was significantly correlated with the area of COP sway in the left hemiparesis group. The velocity of COP sway was not significantly correlated with the FIM (Table 4).

4. Discussion

The main aim of this study was to compare the balance by COP sway and its relation with gait parameters and functional independence in left and right chronic stroke patients. The results showed no differences in the performance of balance, gait, and functional independence between groups with chronic damage to the right or left hemisphere. The chronicity of the patients in this study may have contributed to higher motor adaptation.

As in the present study, Peurala et al. [31] also found no difference in the COP oscillation velocity in either the sagittal or frontal plane among individuals with chronic hemiparesis. Ioffe et al. [32] conducted a study about learning postural control with two groups of hemiparetic patients. Patients were trained in 10 sessions consisting of 2 activities in which they had to displace COP visualized on a screen. In one postural control learning activity, patients with left hemiparesis exhibited a long delay during the initial sessions, while patients with right hemiparesis learned faster; however, the learning speed was similar in both groups after 10 days. The authors argued that specific control of the COP trajectory may require a large amount of sensory information, which is associated with the right hemisphere.

Table 3: Correlation between the area of COP sway and gait parameters.

Gait parameters	LH		RH	
	Healthy limb	Affected limb	Healthy limb	Affected limb
	OE	OE	OE	OE
Step length	−0.48	−0.55	0.74*	0.49
Stride length	−0.78*	−0.79*	0.76*	0.71*
Gait velocity	−0.71*	−0.72*	0.31	0.18
Stance phase	0.74*	0.27	0.41	0.29
Swing phase	−0.74*	−0.27	−0.41	−0.29
Double stance onset	0.38	0.67*	0.31	0.59
Single stance	−0.27	−0.74*	−0.29	−0.41
Cadence	−0.40	−0.38	0.44	0.49

*Pearson's correlation coefficient with $p < 0.05$; R: right hemiparetic group; L: left hemiparetic group.

Table 4: Correlations between FIM and balance (COP).

	L Hemiparesis		R Hemiparesis	
	Total FIM	Motor FIM	Total FIM	Motor FIM
Velocity of oscillation	−0.19	−0.28	−0.04	0.12
Area of oscillation	−0.50	−0.59*	0.52	0.62

*Pearson's correlation coefficient (r) with $p < 0.05$.

Many studies have been conducted on patients during the first months after a stroke. The evidence that patients can still learn with time and stimulation raises questions about the evolution of chronic patients who were or are still being subjected to rehabilitation training and about the lateralization of lesion effects. The participants in this study all suffered from lesions more than 12 months before being tested, with an average time of 32 months since lesion. Muscle strength and tone were evaluated to characterize the pattern of injury patients in the area of the middle cerebral artery, which corresponds to a greater involvement of the upper limb in the hemiparesis. The average age of patients (55 years) of this study was not regarded as elderly, and this may not have affected the balance analysis. According to Ruwer et al. [33] aging affects the ability of the central nervous system to process the vestibular, visual, and proprioceptive signals responsible for maintaining body balance and reducing the ability to modify adaptive reflexes. In this study, the mean static COP sway was not statistically significant. When static balance was tested on force platforms, patients in both groups exhibited greater anterior-posterior COP sway than mediolateral one. According to a literature review on falls in stroke patients by Weerdesteyn et al. [34], these patients exhibit greater body oscillation, especially in the frontal plane, and rely more on the healthy limb to maintain balance. Depending on how oscillation is measured, it can be as much as 1.5 to 5 times greater in stroke patients than in healthy individuals.

Despite this result, mediolateral sway continues to be considered as the best prognostic parameter for revealing balance and risk of falling in hemiparetic patients [6, 29, 35]. To Kadaba et al. [30], the anteroposterior displacement of the COP was faster than the medial-lateral one only among patients after a stroke; it did not occur among healthy subjects.

Although the functional independence results did not reveal a significant difference between the groups, according to Peurala et al. [31], we can assume that the loss of the dominant limb has a greater effect on daily activities than loss of the nondominant limb. The right hemiparesis group represents the loss of the dominant limb because all patients were right-handed. Right hemiparetic patients exhibited better functional abilities, especially in activities involving standing upright, balance, and gait. Areas involved in body schema and spatial perception are affected in left hemiparetic patients. From a neurophysiological perspective, this factor contributes negatively to kinesthetic sensation and perception, eventually leading to the neglect of the affected half of the body [8, 10].

The results for temporal gait parameters, such as duration of the stance and swing phases, were consistent with those in the literature [36, 37]. Dynamic analysis revealed that the healthy limb supported the body weight longer in both groups and spent significantly less time in the swing phase compared with the affected limb. This shows that, despite time and rehabilitation, a persistent difference in muscle strength and perception in the affected lower limb affects gait symmetry. The remaining gait parameters assessed, such as step length, stride length, velocity, stance, and cadence, did not differ significantly between healthy and affected limbs. In stroke patients, the lengths of steps and strides are typically different between healthy and affected limbs, and the length of the gait cycle increases, the stance time increases, the duration of the swing phase decreases, the duration of the double stance increases, cadence and velocity decrease [11, 16, 20, 36–40], and stride width increases [41].

The asymmetry in the propulsion of the lower limbs during gait also increases, as reflected by the difference in ground reaction force [42]. However, the groups of patients in this study did not necessarily follow this pattern, likely indicating adaptation and reduction of motor asymmetry over time.

Nardone et al. [20] found a correlation between COP positioning and muscle strength in the nonaffected lower limb during the single stance phase, with COP remaining greater in the healthy limb. There was a positive correlation between single stance duration and COP sway in the healthy lower limb. The asymmetry found in hemiparetic patients affects gait by increasing the time and effort required to shift the body weight to the affected limb. The degree of asymmetry, as measured by stabilometry, was correlated with the level of difficulty reflected in the gait parameters.

Correlations between COP velocity sway, gait parameters, COP area, and motor FIM were greater in the left hemiparesis group. The greater correlation between COP data and gait parameters in patients with right hemisphere lesions might be due to the role of this hemisphere in sensorimotor integration, which is critical for postural maintenance [15]. Right hemisphere lesions have a very large impact and can cause body schema alterations, contralateral neglect, altered postural alignment [13], and visuomotor impairment [9].

The COP area displacement had more positive correlations with gait parameters in both groups relative to the other parameters. This might be due to the mediolateral component of displacement; the area is calculated based on anterior-posterior and mediolateral displacements, and this parameter is significantly different in hemiparetic patients and healthy individuals [8, 16, 20, 32]. Paillex and So [6] observed reductions of lateral COP displacement and the area of COP displacement after rehabilitation in hemiparetic patients. These results were correlated with reduced muscle strength in the thigh adductors and abductors. A quantitative study found correlations among gait performance, postural stability, and functional assessments in hemiparetic patients. Their results showed that the ability to maintain stability while standing and the ability to shift the center of mass were significantly correlated with gait velocity, stride length, and step length in the paretic limb. After correlating the balance scores obtained by the Fugl-Meyer Assessment with gait variables and stability while in a standing position, they concluded that hemiparetic patients compensate for a lack of balance with smaller steps and a slower gait [23]. In the present study, the static balance area was moderately correlated with COP sway and the FIM in both groups; however, static COP velocity was not correlated with the FIM.

We believe that, with time and rehabilitation treatment, a sensorimotor reorganization makes similar performance improvements in balance and gait test regardless of the side lesioned. However, it is believed that rehabilitation programs should be alert to improve balance especially in the anteroposterior direction and include therapeutic exercises that stimulate the different receptors. The COP sway area was correlated with stride length in both groups. This data can serve as a guideline in the rehabilitation of these patients where training the static balance may reflect the improvement of the stride length.

As a limiting factor, we point out the size of our sample, but we believe that the results are not invalidated due to the range of objective assessments carried out. However, we believe that further research should be carried out considering the standardization of brain injury size and the influence of treatment programs on the improvement of postural control and gait parameters associated with the functionality of this population.

5. Conclusion

There was no difference in the performance of balance, gait, and functional independence between the groups of chronic hemiparetic stroke patients when comparing left hemisphere lesion and right hemisphere lesion. The chronicity of the patients in this study and rehabilitation treatment may have contributed to higher motor adaptation. The correlation of the COP sway area with stride length in both groups can serve as a guideline in the rehabilitation of these patients where training the static balance may reflect the improvement of the stride length.

Although the group with left hemiparesis has shown better correlation of COP and gait parameters, improved functionality, and gait that are routinely worked in rehabilitation programs, this may be related to the improvement of balance. Yet, it is believed that this should also be emphasized so that these variables are also improved. In particular, the anteroposterior direction should be emphasized in chronic stroke patients.

Conflict of Interests

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

Authors' Contribution

All authors contributed equally to this work.

Acknowledgments

The authors thank Daniel Gustavo Goroso for assistance with data analysis and for statistical assistance. The authors would like to thank IMREA/HC/FMUSP and CAPES (Coordenação de Aperfeiçoamento de Pessoal de Nível Superior) for partly supporting this study. This paper was produced as part of the activities of FAPESP Research, Innovation and Dissemination Center for Neuromathematics (Grant no. 2013/07699-0, São Paulo Research Foundation) and was supported by School of Medicine of University of São Paulo-Center for Advanced Studies in Rehabilitation-CEAR.

References

[1] P. W. Duncan, R. Zorowitz, B. Bates et al., "Management of adult stroke rehabilitation care: a clinical practice guideline," *Stroke*, vol. 36, no. 9, pp. e100–e143, 2005.

[2] L. F. Teixeira-Salmela, E. S. G. Oliveira, E. G. S. Santana, and G. P. Resende, "Muscle strengthening and physical conditioning in hemiplegic patients," *Acta Fisiátrica*, vol. 7, no. 3, pp. 108–118, 2000.

[3] L. B. Goldstein, "Acute ischemic stroke treatment in 2007," *Circulation*, vol. 116, no. 13, pp. 1504–1514, 2007.

[4] M. De Sèze, L. Wiart, A. Bon-Saint-Côme et al., "Rehabilitation of postural disturbances of hemiplegic patients by using trunk control retraining during exploratory exercises," *Archives of Physical Medicine and Rehabilitation*, vol. 82, no. 6, pp. 793–800, 2001.

[5] T. Ikai, T. Kamikubo, I. Takehara, M. Nishi, and S. Miyano, "Dynamic postural control in patients with hemiparesis," *American Journal of Physical Medicine and Rehabilitation*, vol. 82, no. 6, pp. 463–469, 2003.

[6] R. Paillex and A. So, "Changes in the standing posture of stroke patients during rehabilitation," *Gait and Posture*, vol. 21, no. 4, pp. 403–409, 2005.

[7] Y.-C. Pai, M. W. Rogers, L. D. Hedman, T. A. Hanke, and C. J. Winstein, "Alterations in weight-transfer capabilities in adults with hemiparesis," *Physical Therapy*, vol. 74, no. 7, pp. 647–659, 1994.

[8] G. Rode, C. Tiliket, and D. Boisson, "Predominance of postural imbalance in left hemiparetic patients," *Scandinavian Journal of Rehabilitation Medicine*, vol. 29, no. 1, pp. 11–16, 1997.

[9] G. Rode, C. Tiliket, P. Charlopain, and D. Boisson, "Postural asymmetry reduction by vestibular caloric stimulation in left hemiparetic patients," *Scandinavian Journal of Rehabilitation Medicine*, vol. 30, no. 1, pp. 9–14, 1998.

[10] E. F. Chagas and M. Tavares, "Symmetry and weight-transfer in hemiplegic patients: relationship between this condition and functional activity performance," *Revista de Fisioterapia da Universidade de São Paulo*, vol. 8, no. 1, pp. 40–50, 2001.

[11] R. Dickstein, S. Shefi, E. Marcovitz, and Y. Villa, "Anticipatory postural adjustment in selected trunk muscles in post stroke hemiparetic patients," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 2, pp. 261–267, 2004.

[12] A. Goto, S. Okuda, S. Ito et al., "Locomotion outcome in hemiplegic patients with middle cerebral artery infarction: the difference between right- and left-sided lesions," *Journal of Stroke and Cerebrovascular Diseases*, vol. 18, no. 1, pp. 60–67, 2009.

[13] M. C. Voos and L. E. P. Valle, "Comparative study between hemispheres affected in encephalic vascular accidents and functional evolution in right-handed individuals," *Revista Brasileira de Fisioterapia*, vol. 12, no. 2, pp. 113–120, 2008.

[14] S. Y. Schaefer, P. K. Mutha, K. Y. Haaland, and R. L. Sainburg, "Hemispheric specialization for movement control produces dissociable differences in online corrections after stroke," *Cerebral Cortex*, vol. 22, no. 6, pp. 1407–1419, 2012.

[15] M. C. Voos, M. E. P. Piemonte, and L. E. P. Valle, "Functional asymmetries in hemiparetic patients: a literature review," *Fisioterapia e Pesquisa*, vol. 14, no. 1, pp. 79–87, 2007.

[16] E. B. Titianova and I. M. Tarkka, "Asymmetry in walking performance and postural sway in patients with chronic unilateral cerebral infarction," *Journal of Rehabilitation Research and Development*, vol. 32, no. 3, pp. 236–244, 1995.

[17] I. V. Bonan, F. M. Colle, J. P. Guichard et al., "Reliance on visual information after stroke. Part I: balance on dynamic posturography," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 2, pp. 268–273, 2004.

[18] B. P. Cunha, S. R. Alouche, I. M. G. Araujo, and S. M. S. F. Freitas, "Individuals with post-stroke hemiparesis are able to use additional sensory information to reduce postural sway," *Neuroscience Letters*, vol. 513, no. 1, pp. 6–11, 2012.

[19] P. Rougier and S. Boudrahem, "Effects of visual feedback of center-of-pressure displacements on undisturbed upright postural control of hemiparetic stroke patients," *Restorative Neurology and Neuroscience*, vol. 28, no. 6, pp. 749–759, 2010.

[20] A. Nardone, M. Godi, M. Grasso, S. Guglielmetti, and M. Schieppati, "Stabilometry is a predictor of gait performance in chronic hemiparetic stroke patients," *Gait and Posture*, vol. 30, no. 1, pp. 5–10, 2009.

[21] A. Shumway-Cook and M. Woollacott, *Motor Control: Theory and Practical Applications*, Williams and Wilkins, Baltimore, Md, USA, 1995.

[22] D. S. Marigold and J. J. Eng, "The relationship of asymmetric weight-bearing with postural sway and visual reliance in stroke," *Gait and Posture*, vol. 23, no. 2, pp. 249–255, 2006.

[23] M. A. Dettmann, M. T. Linder, and S. B. Sepic, "Relationships among walking performance, postural stability, and functional assessments of the hemiplegic patient," *The American Journal of Physical Medicine*, vol. 66, no. 2, pp. 77–90, 1987.

[24] M. Riberto, M. H. Miyazaki, S. S. H. Jucá, H. Sakamoto, P. P. N. Pinto, and L. R. Battistella, "Validation of a Brazilian version of the functional independence measure," *Acta Fisiátrica*, vol. 11, no. 2, pp. 72–76, 2004.

[25] F. P. Kendall, E. K. Mcmreary, and P. G. Provance, *Muscle Tests and Functions*, Manole, São Paulo, Brazil, 4th edition, 1995.

[26] R. W. Bohannon and M. B. Smith, "Interrater reliability of a modified Ashworth scale of muscle spasticity," *Physical Therapy*, vol. 67, no. 2, pp. 206–207, 1987.

[27] A. D. Pandyan, G. R. Johnson, C. I. M. Price, R. H. Curless, M. P. Barnes, and H. Rodgers, "A review of the properties and limitations of the Ashworth and modified Ashworth Scales as measures of spasticity," *Clinical Rehabilitation*, vol. 13, no. 5, pp. 373–383, 1999.

[28] W. E. McIlroy and B. E. Maki, "Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing," *Clinical Biomechanics*, vol. 12, no. 1, pp. 66–70, 1997.

[29] J. A. Raymakers, M. M. Samson, and H. J. J. Verhaar, "The assessment of body sway and the choice of the stability parameter(s)," *Gait and Posture*, vol. 21, no. 1, pp. 48–58, 2005.

[30] M. P. Kadaba, H. K. Ramakrishnan, and M. E. Wootten, "Measurement of lower extremity kinematics during level walking," *Journal of Orthopaedic Research*, vol. 8, no. 3, pp. 383–392, 1990.

[31] S. H. Peurala, P. Könönen, K. Pitkänen, J. Sivenius, and I. M. Tarkka, "Postural instability in patients with chronic stroke," *Restorative Neurology and Neuroscience*, vol. 25, no. 2, pp. 101–108, 2007.

[32] M. E. Ioffe, L. A. Chernikova, R. M. Umarova, N. A. Katsuba, and M. A. Kulikov, "Learning postural tasks in hemiparetic patients with lesions of left versus right hemisphere," *Experimental Brain Research*, vol. 201, no. 4, pp. 753–761, 2010.

[33] S. L. Ruwer, A. G. Rossi, and L. F. Simon, "Balance in the elderly," *Revista Brasileira de Otorrinolaringologia*, vol. 71, no. 3, pp. 298–303, 2005.

[34] V. Weerdesteyn, M. de Niet, H. J. R. van Duijnhoven, and A. C. H. Geurts, "Falls in individuals with stroke," *Journal of Rehabilitation Research and Development*, vol. 45, no. 8, pp. 1195–1214, 2008.

[35] M. De Haart, A. C. Geurts, S. C. Huidekoper, L. Fasotti, and J. van Limbeek, "Recovery of standing balance in post acute stroke patients: a rehabilitation cohort study," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 6, pp. 886–895, 2004.

[36] H. Corriveau, R. Hébert, M. Raîche, and F. Prince, "Evaluation of postural stability in the elderly with stroke," *Archives of Physical Medicine and Rehabilitation*, vol. 85, no. 7, pp. 1095–1101, 2004.

[37] L. M. Saadi, R. Emygdio, C. L. C. Moliaro, and M. D. D'Angelo, "Computerized analysis of gait time parameters in 25 hemiplegic patients," *Revista Medicina de Reabilitação*, vol. 37-38, pp. 18–26, 1994.

[38] M. Saad and L. R. Battistella, *Gait Analysis*, Lemos Editorial, São Paulo, Brazil, 1997.

[39] P. G. Lopes, J. C. P. Vasconcelos, A. M. Ramos, M. C. S. Moreira, and J. A. F. Lopes, "Effect of biofeedback therapy by surface electromyography in hemiparetic gait knee flexion," *Acta Fisiátrica*, vol. 11, no. 3, pp. 125–131, 2004.

[40] B. Manor, K. Hu, P. Zhao et al., "Altered control of postural sway following cerebral infarction: a cross-sectional analysis," *Neurology*, vol. 74, no. 6, pp. 458–464, 2010.

[41] P. M. Davies, *Exactly at the Center*, Manole, São Paulo, Brazil, 1996.

[42] M. G. Bowden, C. K. Balasubramanian, R. R. Neptune, and S. A. Kautz, "Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking," *Stroke*, vol. 37, no. 3, pp. 872–876, 2006.

Hindawi Publishing Corporation
BioMed Research International
Volume 2015, Article ID 123848, 15 pages
http://dx.doi.org/10.1155/2015/123848

Review Article

Effects of Exercise on Spinal Deformities and Quality of Life in Patients with Adolescent Idiopathic Scoliosis

Shahnawaz Anwer,[1,2] Ahmad Alghadir,[1] Md. Abu Shaphe,[3] and Dilshad Anwar[4]

[1] *Rehabilitation Research Chair, Department of Rehabilitation Sciences, College of Applied Medical Sciences, King Saud University, Riyadh 11433, Saudi Arabia*
[2] *Dr. D. Y. Patil College of Physiotherapy, Dr. D. Y. Patil Vidyapeeth, Pune, India*
[3] *Department of Physiotherapy, College of Applied Medical Sciences, Jazan University, Saudi Arabia*
[4] *Department of Orthopedics, JNMC, AMU, Aligarh, India*

Correspondence should be addressed to Shahnawaz Anwer; anwer_shahnawazphysio@rediffmail.com

Received 22 April 2015; Revised 15 June 2015; Accepted 18 June 2015

Academic Editor: Massimiliano Pau

Objectives. This systematic review was conducted to examine the effects of exercise on spinal deformities and quality of life in patients with adolescent idiopathic scoliosis (AIS). *Data Sources*. Electronic databases, including PubMed, CINAHL, Embase, Scopus, Cochrane Register of Controlled Trials, PEDro, and Web of Science, were searched for research articles published from the earliest available dates up to May 31, 2015, using the key words “exercise,” “postural correction,” “posture,” “postural curve,” “Cobb’s angle,” “quality of life,” and “spinal deformities,” combined with the Medical Subject Heading “scoliosis.” *Study Selection*. This systematic review was restricted to randomized and nonrandomized controlled trials on AIS published in English language. The quality of selected studies was assessed by the PEDro scale, the Cochrane Collaboration’s tool, and the Grading of Recommendations Assessment, Development, and Evaluation System (GRADE). *Data Extraction*. Descriptive data were collected from each study. The outcome measures of interest were Cobb angle, trunk rotation, thoracic kyphosis, lumbar kyphosis, vertebral rotation, and quality of life. *Data Synthesis*. A total of 30 studies were assessed for eligibility. Six of the 9 selected studies reached high methodological quality on the PEDro scale. Meta-analysis revealed moderate-quality evidence that exercise interventions reduce the Cobb angle, angle of trunk rotation, thoracic kyphosis, and lumbar lordosis and low-quality evidence that exercise interventions reduce average lateral deviation. Meta-analysis revealed moderate-quality evidence that exercise interventions improve the quality of life. *Conclusions*. A supervised exercise program was superior to controls in reducing spinal deformities and improving the quality of life in patients with AIS.

1. Introduction

Adolescent idiopathic scoliosis (AIS) is a structural deformity of the spine with 3-dimensional deformation, including lateral shift and vertebral rotation affecting children at puberty [1, 2]. The predisposing factors are genetic predisposition; connective tissue abnormalities; and skeletal, muscular, and neurological disturbances during growth. However, the exact cause remains unknown [3]. In the general population, the prevalence of AIS is about 2.5% with a Cobb angle of >10 degrees [2, 3]. A variety of risk factors may result in higher curve progression, including female gender, age of 10–12 years, absence of menarche, presence of thoracic curves, curve size at presentation >25 degrees, Risser sign of 0-1, and residual growth potential [2–5].

The primary goal of rehabilitation for AIS is to reduce the progression of curves thereby decreasing the risk of secondary impairment, including back pain, breathing problems, and cosmetic deformities, and improve the quality of life [3, 6]. Exercise plays a vital role in reducing curve progression and improving quality of life in AIS. Patients with thoracic curves ≤25 degrees and thoracolumbar or lumbar curves ≤20 degrees can effectively manage with exercise alone, whereas patients with thoracic curves of 25–50 degrees and thoracolumbar or lumbar curves of 20–40 degrees require bracing along with exercise [3, 7–9].

In a previous review, Negrini et al. (2008) reported that exercise had beneficial effects on the rate of progression and Cobb angle. They also found positive effects of exercise in reducing brace prescriptions [10]. More recently, Negrini et al. [11] reviewed the best available evidence regarding the rehabilitation approach to AIS and reported that the Society on Scoliosis Orthopaedic and Rehabilitation Treatment (SOSORT) had the best evidence-based guidelines. Low evidence was found for conservative treatment because out of 65 recommendations none had strong evidence (level I), 2 had moderate evidence (level II), and the remainder had weak evidence. Recently, Romano et al. conducted a Cochrane systematic review to evaluate the efficacy of scoliosis-specific exercise (SSE) in adolescent patients with AIS [12] and reported a lack of high-quality evidence to recommend the use of SSE for AIS. They identified one low-quality study that found the use of exercise more effective than electrostimulation, traction, and postural training to avoid scoliosis progression and one very low-quality study that found the use of SSE more effective than traditional physiotherapy for reducing brace prescriptions. Similarly, in a systematic review, Negrini et al. [13] reported lack of solid evidence for or against the effectiveness of physical exercise for reducing curve progression in AIS. In addition, Mordecai and Dabke reported low-quality evidence for the effect of exercise in the treatment of AIS [8].

To date, no systematic review has examined the effects of exercise on quality of life in patients with AIS. Therefore, more evidence is required regarding the effects of exercise on curve reduction and improvement in quality of life in AIS is required. The aim of this systematic review was to evaluate the effects of an exercise program on spinal deformities and quality of life.

2. Methods

2.1. Data Sources. Electronic databases, including Pubmed, CINAHL, Embase, Scopus, Cochrane Register of Controlled Trials, PEDro, and Web of Science, were searched for published studies using the keywords "postural correction," "postural curve," "Cobb angle," "quality of life," and "spinal deformities" combined with the Medical Subject Heading "scoliosis" and "exercise." The bibliographical search was restricted to randomized and nonrandomized controlled trials published in English language from the earliest available dates up to May 31, 2015 (see Search Strategy). Original authors were contacted and asked to provide the full text of potential papers that were not accessible. Two independent reviewers (Shahnawaz Anwer and Md. Abu Shaphe) selected the studies based on titles and abstracts, excluding those articles not related to the objectives of this review. Consensus between the reviewers was obtained through discussion.

Search Strategy

(1) Pubmed:

#1 Scoliosis [MeSH Terms],
#2 Exercise [MeSH Terms],
#3 Postural correction [MeSH Terms],
#4 Cobb angle [MeSH Terms],
#5 Postural curve [MeSH Terms],
#6 Quality of life [MeSH Terms],
#7 [#1 AND (#2 OR #3)],
#8 [#1 AND (#4 OR #5)],
#9 [#1 AND (#2 OR #6)],
Limits: Comparative study, randomized controlled trial;

(2) Cochrane Register of Controlled Trials:

#1 Scoliosis [MeSH Terms],
#2 Exercise [MeSH Terms],
#3 Postural correction [MeSH Terms],
#4 Cobb angle [MeSH Terms],
#5 Postural curve [MeSH Terms],
#6 Quality of life [MeSH Terms],
#7 [#1 AND (#2 OR #3)],
#8 [#1 AND (#4 OR #5)],
#9 [#1 AND (#2 OR #6)],
Limits: Trials;

(3) PEDro:

∗ Advance Search,
Title or abstract: Scoliosis, exercise, Cobb angle, Quality of life,
Method: Clinical trial;

(4) Web of Science:

∗ Advance Search,
#1 Scoliosis [MeSH Terms],
#2 Exercise [MeSH Terms],
#3 Cobb angle [MeSH Terms],
#4 Quality of life [MeSH Terms],
#5 (#1 AND #2),
#6 (#1 AND #3),
#7 (#1 AND #2 AND #4),
Limits: language (English), Document type (Article).

2.2. Study Selection. Studies were included on the basis of the following criteria: randomized and nonrandomized controlled methodology; subjects with AIS in the age group of 10–19 years; studies comparison of exercise with other interventions or controls; and outcome measures of radiological deformity (i.e., Cobb angle), surface deformities (including trunk rotation, thoracic kyphosis, lumbar lordosis, and average lateral deviation), and quality of life. Studies were excluded if subjects were >19 years of age, interventions did not include exercise or compare exercise with a control, or published results were in abstract form only. Final study selection was achieved through discussion and consensus between the two reviewers.

2.3. Data Extraction. The selected studies were screened by 2 independent reviewers (Shahnawaz Anwer and Md. Abu Shaphe). One of the reviewers had prior experience using the extraction form, systematic review methodology, and quality appraisal tools, including the PEDro and Cochrane databases, and had published two systematic reviews and a meta-analysis. The other reviewer was trained beforehand in the use of the extraction form, systematic review methodology, and quality appraisal tools, including the PEDro and Cochrane databases. The following items were extracted: author/year, design of the study, subject's characteristics, age, sex, sample size, details of exercise program (type, duration, dose, and frequency), outcomes, and conclusions. The two reviewers discussed the data with each other or consulted with a third reviewer (Ahmad Alghadir) to reach consensus. Agreement between the two reviewers was obtained using unweighted kappa (κ) statistics. Mean and standard deviation of the baseline and final end point scores for the Cobb angle, trunk rotation, thoracic kyphosis, lumbar lordosis, lateral deviation, and function were extracted from included studies. The mean change score (final score minus baseline score) for each outcome measure was calculated for each intervention. The standardized mean difference (SMD) for all the outcomes was computed [14].

Cohen's categories were used to define the magnitude of the effect size with values of <0.5 as a small effect size; ≥0.5 and ≤0.8, as a medium effect size; and >0.8, as a large effect size [15]. A fixed-effects meta-analysis was conducted to determine the overall effect size of exercise. The z statistic was used to test the significance of an overall effect. Cochran's Q statistic and Higgins' I^2 statistic were used to determine statistical heterogeneity between studies [14]. All statistics were computed using Comprehensive Meta-Analysis software [16].

2.4. Assessment of Methodological Quality. The 11-item PEDro scale was used to assess the quality of included studies by two independent reviewers (Shahnawaz Anwer and Md. Abu Shaphe) [17]. A study with a score ≥6 was considered high-quality as reported previously [18]. In addition, the Cochrane Collaboration's tool was used to assess the risk of bias. Sequence generation, allocation concealment, blinding, completeness of outcome data, and absence of selective outcome reporting were also assessed. Risk of bias was classified as low, unclear, or high in each domain [19]. Agreement between the two reviewers in regard to the PEDro and Cochrane tools was made using unweighted kappa (κ) statistics.

The quality of evidence was determined using the Grading of Recommendations Assessment, Development, and Evaluation System (GRADE) for each meta-analysis [20]. This method involves downgrading evidence from high-quality to moderate-quality to low-quality and to very low-quality using some factors. If the majority of studies (more than 50%) in the meta-analysis had a PEDro score <6 or had more than low levels of statistical heterogeneity between the studies ($I^2 > 25\%$) [21] or if the studies had large confidence intervals suggestive of a small number of subjects in the studies, then the evidence would be downgraded, for example, from high- to moderate-quality. In the presence of serious methodological flaws, for example, if all studies in the meta-analysis had low PEDro scores (<6) with no allocation concealment and blinding, the evidence would be double downgraded (e.g., from high- to low-quality). The criteria for the grade applied to each meta-analysis are explained as a footnote.

3. Results

3.1. Identified Studies. The abstracts of 30 studies were assessed for eligibility. Twenty-one studies [22–42] were eliminated because they did not match the inclusion criteria or were not available in full text (Figure 1). A total of 9 studies were included in the quality assessment phase [43–51].

3.2. Quality Assessment of Study. The 9 included studies had an average PEDro score of 5.7/10, as illustrated in Table 1. Agreement between reviewers was good (unweighted κ = 0.79). However, multiple sources of bias in these studies may have skewed the results. The most common shortcomings were lack of randomization [46–49, 51], lack of concealed allocation [46–49, 51], and lack of blinding (patient, therapist, or assessor) [43–51]. The most adhered ones to items on the PEDro scale were baseline comparability, follow-up, intention-to-treat analysis, measurements of variability, and between-group comparisons, which were evident in almost all the trials.

Agreement between the reviewers was excellent (unweighted κ = 0.87) in assessing risk of bias across studies. Details of the risk of bias assessment of included studies are given in Table 2. The overall risk of bias assessment indicated that the risk of bias was low in 1 study [43], high in 5 studies [46–49, 51], and unclear in 3 studies [44, 45, 50]. The most common shortcomings were lack of blinding [47–49, 51], lack of concealment [46–49, 51], and inadequate random sequence generation [46–49, 51].

3.3. Characteristics of Study Populations. Table 3 details participant characteristics. The sample size for whole study groups ranged from 30 to 252, with the mean age varying from 12 to 15 years. In most of the studies, the majority of participants with AIS were female [43, 44, 46–49, 51]. Most of the studies used the Cobb angle and Risser sign as inclusion criteria for participants with AIS [43–49, 51].

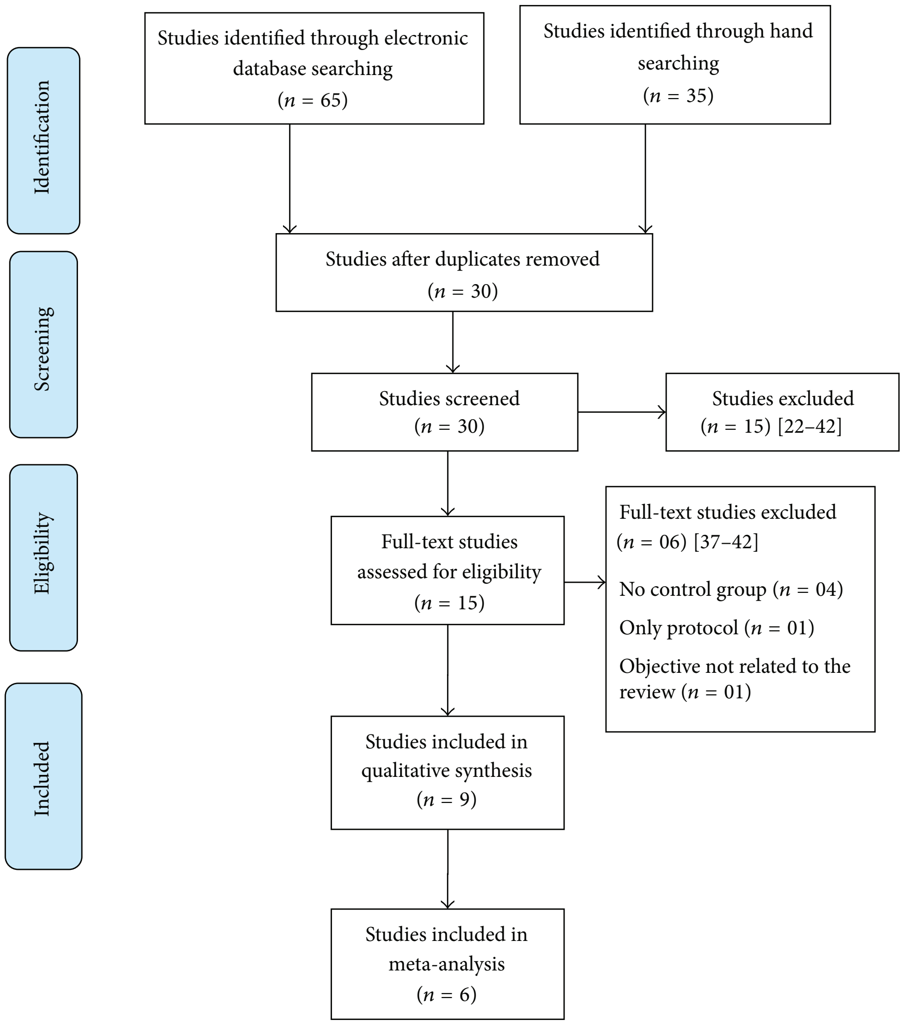


Figure 1: Flow diagram of the study procedure.

Table 1: Methodological classification assessed by PEDro scale.

Criteria	Monticone et al. (2014) [43]	Kuru et al. (2015) [44]	Diab (2012) [45]	Noh et al. (2014) [46]	Negrini et al. (2006) [47]	Negrini et al. (2006) [48]	Weiss and Klein (2006) [49]	Weiss et al. (2002) [50]	Negrini et al. (2008) [51]	Cumulative score*
Random allocation?	Yes	Yes	Yes	No	No	No	No	Yes	No	4
Concealed allocation?	Yes	Yes	Yes	No	No	No	No	Yes	No	4
Baseline comparability?	Yes	Yes	Yes	Yes	Yes	Yes	Yes	No	Yes	8
Blind participants?	Yes	No	No	No	No	No	No	No	No	1
Blind therapists?	No	No	No	No	No	No	No	No	No	0
Blind assessors?	Yes	No	No	Yes	No	No	No	No	No	2
Follow-up?	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	9
Intention-to-treat analysis?	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	9
Group comparisons?	Yes	Yes	Yes	Yes	Yes	Yes	No	No	Yes	7
Point and variability measures?	Yes	Yes	Yes	Yes	Yes	Yes	Yes	No	No	7
Cumulative score	9	7	7	6	5	5	4	4	4	**5.7**†

*Out of the 10 total studies.
†Maximum score of 10.

Table 2: Risk of bias of included studies (yes, low risk of bias; no, high risk of bias).

Citations	Adequate sequence generation?	Allocation concealment?	Blinding?	Incomplete outcome data addressed?	Free of selective reporting?	Conclusions
Monticone et al. (2014) [43]	Yes	Yes	Yes	Yes	Yes	Low risk of bias
Kuru et al. (2015) [44]	Yes	Yes	Unclear	Yes	Yes	Unclear risk of bias
Diab (2012) [45]	Yes	Yes	Unclear	Yes	Yes	Unclear risk of bias
Noh et al. (2014) [46]	No	No	Yes	Yes	Yes	High risk of bias
Negrini et al. (2006) [47]	No	No	No	Yes	Yes	High risk of bias
Negrini et al. (2006) [48]	No	No	No	Yes	Yes	High risk of bias
Weiss and Klein (2006) [49]	No	No	No	Yes	Yes	High risk of bias
Weiss et al. (2002) [50]	Yes	Yes	Unclear	Yes	Yes	Unclear risk of bias
Negrini et al. (2008) [51]	No	No	No	Yes	Yes	High risk of bias

3.4. Training Protocol. Table 3 summarizes the training protocol. Three studies compared the Scientific Exercise Approach to Scoliosis (SEAS.02) exercises with controls [47, 48, 51], 1 study compared active self-correction and task-oriented exercises with controls [43], 1 study compared Schroth exercises with controls [44], 1 study compared forward head correction and traditional exercise with controls [45], 1 study compared the 3D corrective spinal technique with controls [46], 1 study compared physiologic exercise program and scoliosis intensive rehabilitation (SIR) with control [49], and 1 study compared passive transverse force and SIR with a control group [50]. In all included studies, the control group received usual care or performed a traditional exercise program. Only one study [43] had a report of an adverse effect which was a minor temporary worsening of pain.

3.5. Outcome Measures. Six studies used the Cobb angle [43, 44, 46–48, 51], 5 studies used the angle of trunk rotation [43, 44, 47, 48, 51], 3 studies used the thoracic kyphosis angle [45, 46, 49], 2 studies used the lumbar lordosis angle [45, 46], and 3 studies used the average lateral deviation [45, 49, 50] to measure various spinal deformities. Radiographic methods were used to measure the Cobb angle in all six included studies, and a Scoliometer was used to measure the angle of trunk rotation in the 5 included studies. Two studies used a Formetric device to measure thoracic kyphosis [45, 49], and 1 study used a radiographic method for this measurement [46]. One study used a Formetric device to measure lumbar lordosis [45], whereas the other study used a radiographic method for this measurement [46]. Average lateral deviation was measured with a Formetric device in all 3 included studies. Two studies used the Scoliosis Research Society-22 patient questionnaire (SRS-22) [43, 46], 1 study used SRS-23 [44], and another used the Functional Rating Index to measure health related quality of life [45]. The Functional Rating Index is a patient-rated scale composed of 10 items including 4 subscales: pain, sleep, work, and daily activity [52]. The subscales include 3 domains of the World Health Organization International Classification of Functioning, Disability, and Health (WHO-ICF) such as activity limitations with 6 items (personal care, travel, recreation, lifting, walking, and standing), impairment with 3 items (pain frequency, pain intensity, and sleep), and participation restriction with 1 item (work). Each item was scored on a 5-point scale ranging from 0 (no pain or maximum ability) to 4 (maximum pain or disability). The possible score ranges from 0% (no disability) to 100% (severe disability).

3.6. Effect of Exercise on Spinal Deformities. Table 4 gives details of the results of the exercise and control group in included studies. Data syntheses of included studies are given in Table 5 and Figures 2–7. Meta-analysis of 4 studies [43, 44, 46, 47] provided moderate-quality evidence with a significant effect size point estimate across the 4 included studies (p = 0.000), with an overall medium effect size point estimate of 0.65 (95% CI, −0.89 to −0.40) based on a fixed-effects model that favored exercise compared with controls in reducing the Cobb angle (Figure 2).

Meta-analysis of 2 studies [43, 44] provided moderate-quality evidence with a significant effect (p = 0.000), with an overall medium effect size point estimate of 0.73 (95% CI, −1.07 to −0.39) based on a fixed-effects model that favored exercise compared with controls in reducing the angle of trunk rotation (Figure 3).

Meta-analysis of 3 studies [45, 46, 49] provided moderate-quality evidence with a significant effect size point estimate across the 3 included studies (p = 0.001), with an overall medium effect size point estimate of 0.55 (95% CI, −0.89 to −0.22) based on a fixed-effects model that favored exercise compared with controls in reducing the thoracic kyphosis angle (Figure 4).

Meta-analysis of 2 studies [45, 46] provided moderate-quality evidence with a significant overall effect (p = 0.003), with an overall medium effect size point estimate of 0.57 (95% CI, −0.96 to −0.19) based on a fixed-effects model that favored exercise compared with controls in reducing lumbar lordosis (Figure 5).

Meta-analysis of 2 studies [45, 50] provided low-quality evidence with a significant overall effect (p = 0.005), with an overall medium effect size point estimate of 0.54 (95% CI, −0.92 to −0.16) based on a fixed-effects model that favored

Table 3: Overview of selected studies in adolescent idiopathic scoliosis.

Study	Subjects	Mean age, years (male/female, %)	Design	Group	Duration	Adverse effects	Conclusions
Monticone et al. (2014) [43]	AIS Cobb angle: 10–25 degrees Risser sign: <2	Group 1: 12.5 (29/71) Group 2: 12.4 (25/75)	RCT	1: active self-correction and task-oriented exercise (n = 55) 2: control: traditional spinal exercise (n = 55)	60-minute outpatient sessions once a week and 30-minute home exercise sessions twice a week for 2 weeks Follow-up: 12 months	Minor temporary pain worsening (n = 11 exp. group; n = 14 control group)	The active self-correction and task-oriented exercise was superior to traditional spinal exercises in reducing spinal deformities
Kuru et al. (2015) [44]	AIS Cobb angle: 10–60 degrees Risser sign: 0–3	Group 1: 12.9 (7/93) Group 2: 12.8 (13/87)	RCT	1: Schroth exercises (n = 15) 2: control (n = 15)	90-minute sessions thrice a week for 6 weeks Follow-up: 18 weeks	Not reported	Supervised Schroth exercise was superior to control group in reducing spinal deformities.
Diab (2012) [45]	AIS Cobb angle: 10–30 degrees Risser sign: 0–2	Group 1: 13.2 (53/47) Group 2: 14.5 (55/45)	RCT	1: forward head correction and traditional exercise (n = 38) 2: control: traditional exercise (n = 38)	3 sessions a week for 10 weeks Follow-up: 3 months	Not reported	A regime of forward head corrective exercise in addition to traditional exercises improved scoliotic posture and functional status.
Noh et al. (2014) [46]	AIS Risser sign: 0–4	Group 1: 13.8 (25/75) Group 2: 14.9 (13/87)	Retrospective Nonrandomized	1: 3D corrective spinal technique (n = 16) 2: control: traditional exercise (n = 16)	60-minute sessions 2-3 times per week for 3.5 to 4 months Follow-up: no	Not reported	A regime of corrective spinal technique was superior to traditional exercise in reducing most of the spinal deformities and improved quality of life.
Negrini et al. (2006) [47]	AIS Cobb angle: >15 degrees Risser sign: 0–3	Group 1: 13.3 (17/83) Group 2: 13.6 (13/87)	Prospective nonrandomized controlled study	1: SEAS.02 (n = 40) 2: control (n = 70)	1: 1.5-hour sessions every 2-3 months with prosecution in a facility near home for 40 minutes twice a week and 1 exercise daily for 5 minutes 2: performing exercises 2-3 times a week for 45 to 90 minutes Follow-up: no	Not reported	SEAS.02 exercises were superior to control group for reducing spinal deformities.
Negrini et al. (2006) [48]	AIS Cobb angle: >15 degrees Risser sign: 0–3	Group 1: 12.7 (22/78) Group 2: 12.1 (24/76)	Prospective nonrandomized controlled study	1: SEAS.02 (n = 23) 2: control (n = 25)	1: 1.5-hour sessions every 2-3 months with prosecution in a facility near home for 40 minutes twice a week and 1 exercise daily for 5 minutes 2: performing exercises 2-3 times a week for 45 to 90 minutes Follow-up: no	Not reported	SEAS.02 exercises were superior to control group for reducing spinal deformities.

Table 3: Continued.

Study	Subjects	Mean age, years (male/female, %)	Design	Group	Duration	Adverse effects	Conclusions
Weiss and Klein (2006) [49]	AIS Cobb angle: >20 degrees	Group 1: 15.3 (0/100) Group 2: 14.7 (0/100)	Prospective controlled study	1: SIR and physiologic exercise (n = 18) 2: control: SIR (n = 18)	1: 5 days a week (2 hours in the morning and evening each) for 4 weeks and additionally 90 minutes of physiologic exercise for 5 days a week on second or third week 2: 5 days a week (2 hours in the morning and evening each) for 4 weeks	Not reported	Physiologic exercise program in addition to SIR was superior to SIR alone for correcting lateral deviation.
Weiss et al. (2002) [50]	AIS Age group: 12 to 18 years	Group 1: 14.8 (NR) Group 2: 15.2 (NR)	RCT	1: SIR and PTF treatment (n = 126) 2: control: SIR (n = 126)	1: having 5-6 hours of in-patient intensive program and additionally 4–6 PTF treatment of 20 minutes for 4–6 weeks 2: having only 5-6 hours of in-patient intensive program Follow-up: no	Not reported	In-patient rehabilitation with PTF was superior to in-patient rehabilitation alone to correct scoliotic posture.
Negrini et al. (2008) [51]	AIS Cobb angle: >15 degrees Risser sign: 0–3	Group 1: 12.7 (29/71) Group 2: 12.1 (31/69)	Prospective controlled cohort study	1: SEAS.02 (n = 35) 2: control: usual physiotherapy (n = 39)	1: 1.5-hour sessions every 2-3 months with prosecution in a facility near home for 40 minutes twice a week and 1 exercise daily for 5 minutes 2: performing exercises 2-3 times a week for 45 to 90 minutes Follow-up: no	Not reported	SEAS.02 exercises were superior to control group for reducing progression of scoliosis.

AIS, adolescent idiopathic scoliosis; RCT, randomized controlled trial; SEAS.02, Scientific Exercise Approach to Scoliosis; SIR, scoliosis intensive rehabilitation; PTF, passive transverse force; SRS-22, Scoliosis Research Society-22.

TABLE 4: Results of exercise and control group in included studies.

(a)

Citation	Outcomes	Exercise group Pretest	Posttest	Follow-up	^{t}D	^{f}D	Control group Pretest	Posttest	Follow-up	^{t}D	^{f}D	p values*
Monticone et al. (2014) [43]	Cobb angle (degree)	19.3 (3.9)	14.0 (2.4)	14.3 (2.3)	−5.3 (0.6)	−4.9 (0.4)	19.2 (2.5)	20.9 (2.2)	22.0 (1.6)	1.7 (0.3)	2.8 (0.4)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.861^{\#}$ $<0.001^{\#\#}$
	Angle of trunk rotation (degree)	7.1 (1.4)	3.6 (1.1)	3.3 (1.1)	−3.5 (0.2)	3.7 (0.2)	6.9 (1.3)	6.6 (1.2)	6.5 (1.1)	−0.2 (0.1)	−0.4 (0.1)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.403^{\#}$ $<0.001^{\#\#}$
	SRS-22											
	Function (0–5)	3.8 (0.5)	4.7 (0.2)	4.8 (0.2)	0.89 (0.07)	1.0 (0.07)	3.9 (0.5)	4.0 (0.4)	3.9 (0.4)	0.09 (0.03)	0.01 (0.04)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.404^{\#}$ $<0.001^{\#\#}$
	Pain (0–5)	3.8 (0.4)	4.6 (0.3)	4.7 (0.2)	0.82 (0.05)	0.89 (0.06)	3.9 (0.5)	4.3 (0.3)	4.2 (0.4)	0.45 (0.06)	0.33 (0.06)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.383^{\#}$ $<0.001^{\#\#}$
	Image (0–5)	3.6 (0.6)	4.4 (0.3)	4.6 (0.3)	0.82 (0.07)	1.0 (0.08)	3.4 (0.6)	3.7 (0.5)	3.6 (0.4)	0.30 (0.03)	0.21 (0.04)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.094^{\#}$ $<0.001^{\#\#}$
	Mental health (0–5)	3.8 (0.6)	4.5 (0.3)	4.7 (0.2)	0.75 (0.07)	0.95 (0.08)	3.9 (0.6)	4.01 (0.5)	3.8 (0.4)	0.11 (0.03)	−0.1 (0.04)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.433^{\#}$ $<0.001^{\#\#}$
	Satisfaction	NR	4.8 (0.3)	4.9 (0.3)	NR	NR	NR	4.0 (0.5)	4.2 (0.5)	NR	NR	$<0.001^{\#\#}$
Kuru et al. (2015) [44]	Cobb angle (degree)	33.4 (8.9)	NR	30.87 (8.9)	NR	−2.53	30.3 (6.6)	NR	33.3 (6.6)	NR	3.13	$0.397^{\#}$ $0.006^{\#\#\#}$
	Angle of trunk rotation (degree)	11.9 (5.2)	NR	7.66 (5.24)	NR	−4.23 (4.78)	8.4 (2.9)	NR	10.5 (4.21)	NR	2.06 (2.09)	$0.106^{\#}$ $0.000^{\#\#\#}$
	SRS-23	3.9 (0.6)	NR	4.23 (0.7)	NR	0.33 (0.34)	4.1 (0.4)	NR	4.07 (0.4)	NR	−0.03 (0.23)	$0.452^{\#}$ $0.131^{\#\#\#}$

(a) Continued.

Citation	Outcomes	Exercise group Pretest	Exercise group Posttest	Exercise group Follow-up	^{t}D	^{f}D	Control group Pretest	Control group Posttest	Control group Follow-up	^{t}D	^{f}D	p values*
Diab (2012) [45]	Thoracic kyphosis	8.9 (0.9)	10.4 (1.1)	10 (1.05)	NR	NR	8.8 (1.5)	9 (1.8)	8.9 (1.7)	NR	NR	$0.001^{\#\#}$ $0.004^{\#\#\#}$
	Lumbar lordosis	18.6 (5.4)	21.6 (1.8)	20.9 (1.9)	NR	NR	18.1 (5.5)	20 (1.7)	19.2 (1.6)	NR	NR	$0.01^{\#\#}$ $0.017^{\#\#\#}$
	Lateral deviation	16.8 (2.3)	14.3 (2.3)	14.7 (2.4)	NR	NR	15.1 (1.8)	14.5 (1.6)	15.5 (1.7)	NR	NR	$0.001^{\#\#}$ $0.002^{\#\#\#}$
	Functional index	13.9 (1.7)	10.7 (0.9)	10 (0.9)	NR	NR	16.1 (1.7)	11.9 (0.8)	13.8 (1.9)	NR	NR	$0.8^{\#\#}$ $0.001^{\#\#\#}$

(b)

Citation	Outcomes	Exercise group Pretest	Exercise group Posttest	^{t}D	Control group Pretest	Control group Posttest	^{t}D	p values*
Noh et al. (2014) [46]	Cobb angle (degree)	21.6 (10.1)	13.5 (12)	8.1 (4.5)	19 (7)	14.7 (7.2)	4.3 (2.1)	$<0.001^{\dagger}$ $<0.001^{\ddagger}$ $0.003^{\#\#}$
	Thoracic kyphosis	26.7 (12.6)	25.5 (9.3)	1.2 (9.9)	24.3 (8.1)	24.5 (7.5)	−0.2 (7)	$0.611^{\dagger}$ $0.904^{\ddagger}$ $0.625^{\#\#}$
	Lumbar lordosis	52.8 (17.8)	47.7 (6.7)	5 (14.2)	45.6 (12.8)	49 (7.4)	−3.3 (13.4)	$0.176^{\dagger}$ $0.332^{\ddagger}$ $0.095^{\#\#}$
	SRS22							
	Function (0–5)	4.1 (2)	4.7 (1)	NR	4.4 (0.8)	4.6 (1)	NR	$0.027^{\dagger}$ $0.083^{\ddagger}$ $0.931^{\#}$ $0.216^{\#\#}$
	Pain (0–5)	4.5 (2.4)	4.9 (1)	NR	3.8 (1.6)	4.6 (2.4)	NR	$0.026^{\dagger}$ $0.066^{\ddagger}$ $0.140^{\#}$ $0.190^{\#\#}$
	Image (0–5)	3.3 (1.2)	4.2 (1)	NR	2.9 (0.8)	3.4 (1)	NR	$0.011^{\dagger}$ $0.102^{\ddagger}$ $0.343^{\#}$ $0.026^{\#\#}$
	Mental health (0–5)	4 (3)	4.6 (1.4)	NR	3 (1.4)	4 (1.2)	NR	$0.026^{\dagger}$ $0.066^{\ddagger}$ $0.228^{\#}$ $0.121^{\#\#}$
	Satisfaction	NR	5 (1)	NR	NR	4 (1)	NR	$0.039^{\#\#}$
	Total	3.8 (1.8)	4.5 (0.4)	NR	3.5 (1.1)	4.1 (1.4)	NR	$0.012^{\dagger}$ $0.066^{\ddagger}$ $0.306^{\#}$ $0.041^{\#\#}$

(b) Continued.

Citation	Outcomes	Exercise group Pretest	Exercise group Posttest	${}^{t}D$	Control group Pretest	Control group Posttest	${}^{t}D$	p values*
Negrini et al. (2006) [47]	Cobb angle (degree)	30.6 (10.8)	NR	−5.7 (5.6)	31.3 (11.3)	NR	−3.4 (11.3)	<0.05¶
Negrini et al. (2006) [48]	Cobb angle (degree)	15.3 (5.4)	NR	−3.2 (6.2)	14.9 (6)	NR	NR	<0.05¶
Weiss and Klein (2006) [49]	Lateral deviation (mm)	15.4 (5.1)	13.1 (5)	2.31 (5.6)	13.5 (6.8)	13.1 (6.2)	0.32 (2.5)	0.1† 0.6‡
	Thoracic kyphosis angle (degree)	46.5 (8)	45.8 (7.7)	NR	48.8 (11)	46.8 (10.2)	NR	>0.05† >0.05‡
Weiss et al. (2002) [50]	Lateral deviation (mm)	NR	NR	1.50	NR	NR	0.94	0.030† 0.103‡
Negrini et al. (2008) [51]	Cobb angle (degree)	NR	NR	−0.67	NR	NR	1.38	<0.05† <0.05‡
	Angle of trunk rotation (degree)	NR	NR	0.12	NR	NR	0.52	>0.05† >0.05‡

†Pretest versus posttest in exercise group; ‡pretest versus posttest in control group; #between-group comparison at baseline; ##between-group comparison at posttest; ###between-group comparison at follow-up; ${}^{t}D$, difference between pretest and posttest; ${}^{f}D$, difference between pretest and follow-up; ¶comparison of mean difference between two groups; *significant at $p < 0.05$; NR, not reported.

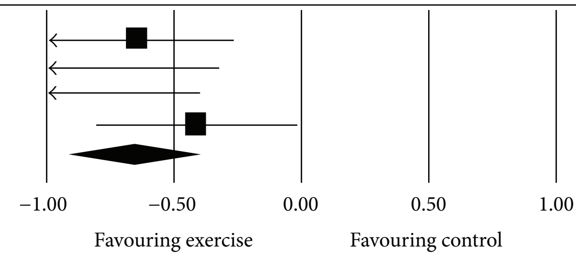
Table 5: Meta-analyses of effect of exercise program.

Outcomes	Number of studies	Ratio of studies (PEDro <6)	Number of subjects	SMD [95% CI]	I^2	Quality of evidence (GRADE)
Cobb angle	4	25%	282	0.65 [−0.89, −0.40]	30.53%	Moderate†
Angle of trunk rotation	2	0%	140	0.73 [−1.07, −0.39]	1.49%	Moderate‡
Thoracic kyphosis angle	3	33%	144	0.55 [−0.89, −.22]	0%	Moderate‡
Lumbar lordosis angle	2	0%	108	0.57 [−0.96, −0.19]	0%	Moderate‡
Average lateral deviation	2	50%	112	0.54 [−0.92, −0.16]	46%	Low¶
Quality of life	3	0%	138	0.73 [−1.07, −0.38]	0%	Moderate‡

GRADE, GRADE working group grades of evidence.
†Statistical heterogeneity results downgrade (I^2 > 25%). ‡Large confidence interval results downgrade. ¶Large confidence interval, statistical heterogeneity results downgrade.

Study name	Statistics for each study: Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Monticone et al., 2014	−0.645	−1.028	−0.262	−3.298	0.001
Kuru et al., 2015	−1.086	−1.852	−0.319	−2.776	0.006
Noh et al., 2014	−1.142	−1.889	−0.395	−2.995	0.003
Negrini et al., 2006	−0.412	−0.804	−0.020	−2.059	0.040
	−0.653	−0.897	−0.409	−5.241	0.000

Std. diff. in means and 95% CI

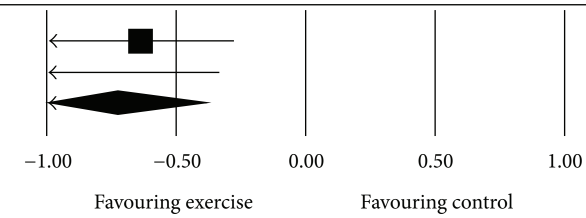


Heterogeneity: Q-value, 4.319 (p = 0.229); I^2, 30.53%

Figure 2: Effect of exercise on the Cobb angle.

Study name	Statistics for each study: Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Monticone et al., 2014	−0.645	−1.028	−0.262	−3.298	0.001
Kuru et al., 2015	−1.086	−1.852	−0.319	−2.776	0.006
	−0.733	−1.076	−0.390	−4.191	0.000

Std. diff. in means and 95% CI

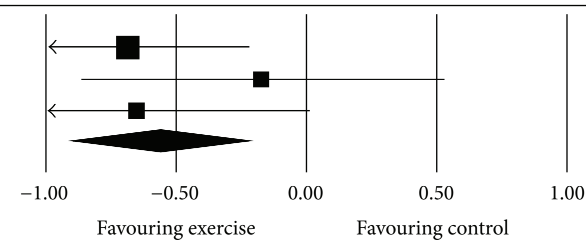


Heterogeneity: Q-value, 1.015 (p = 0.314); I^2, 1.49%

Figure 3: Effect of exercise on the angle of trunk rotation.

Study name	Statistics for each study: Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Diab, 2012	−0.682	−1.144	−0.219	−2.888	0.004
Noh et al., 2014	−0.175	−0.869	0.520	−0.493	0.622
Weiss and Klein, 2006	−0.649	−1.319	0.022	−1.896	0.058
	−0.556	−0.890	−0.222	−3.266	0.001

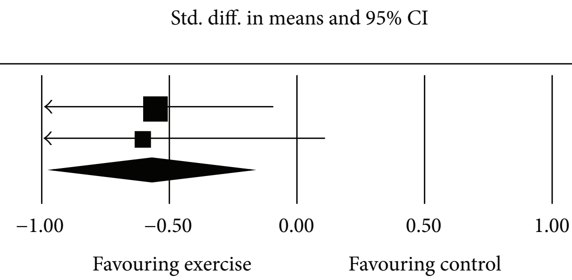
Std. diff. in means and 95% CI



Heterogeneity: Q-value, 1.516 (p = 0.469); I^2, 0%

Figure 4: Effect of exercise on the thoracic kyphosis angle.

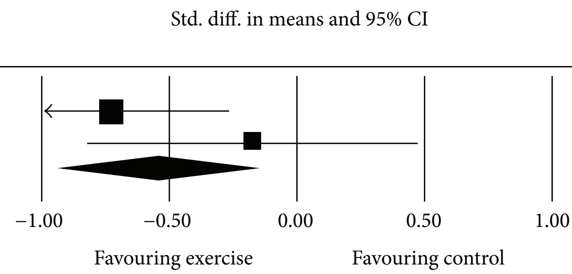
Study name	Statistics for each study				
	Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Diab, 2012	−0.560	−1.019	−0.102	−2.395	0.017
Noh et al., 2014	−0.610	−1.318	0.099	−1.685	0.092
	−0.575	−0.960	−0.190	−2.927	0.003



Heterogeneity: Q-value, 0.013 ($p = 0.909$); I^2, 0%

Figure 5: Effect of exercise on the lumbar lordosis angle.

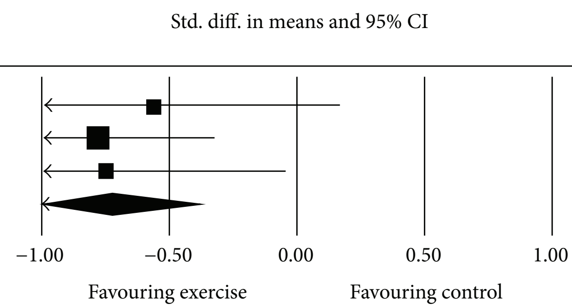
Study name	Statistics for each study				
	Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Diab, 2012	−0.735	−1.200	−0.270	−3.101	0.002
Weiss and Klein, 2006	−0.176	−0.831	0.478	−0.528	0.597
	−0.548	−0.927	−0.169	−2.835	0.005



Heterogeneity: Q-value, 1.860 ($p = 0.173$); I^2, 46%

Figure 6: Effect of exercise on the average lateral deviation.

Study name	Statistics for each study				
	Std. diff. in means	Lower limit	Upper limit	Z-value	p value
Kuru et al., 2015	−0.568	−1.298	0.162	−1.525	0.127
Diab, 2012	−0.786	−1.253	−0.319	−3.302	0.001
Noh et al., 2014	−0.755	−1.472	−0.038	−2.063	0.039
	−0.730	−1.075	−0.386	−4.152	0.000



Heterogeneity: Q-value, 0.249 ($p = 0.883$); I^2, 0%

Figure 7: Effect of exercise on the quality of life.

exercise compared with controls in reducing average lateral deviation (Figure 6).

3.7. Effect of Exercise on Quality of Life. Meta-analysis of 3 studies [44–46] provided moderate-quality evidence with a significant effect size point estimate across the 3 included studies ($p = 0.000$), with an overall medium effect size point estimate of 0.73 (95% CI, −1.07 to −0.03) based on a fixed-effects model that favored exercise compared with controls in improving the quality of life (Table 5 and Figure 7).

4. Discussion

This systematic review investigated current available evidence on the effects of an exercise program on spinal deformities and quality of life in patients with AIS. The review evaluated 9 studies, including a total of 768 participants.

Among the 9 studies evaluated using the PEDro scale [17], 6 were considered of high methodological quality [43–48]. The overall risk of bias assessment showed that 5 studies had a high risk of bias [46–49, 51], and 1 study had a low risk of bias [43], while others had an unclear risk of bias [44, 45, 50]. More than half of the studies failed to perform blinding and (Table 2).

The results of the present systematic review provide moderate-quality evidence for exercise intervention with a medium effect size for reducing the Cobb angle, angle of trunk rotation, thoracic kyphosis angle, and lumbar lordosis angle and improving the quality of life in patients with AIS. Similarly, a systematic review conducted by Fusco et al. [53] reported improvement in the Cobb angle following a regime of exercise. In another review, Negrini et al. [10] confirmed the efficacy of exercises in reducing the progression of deformity and Cobb angles in patients with AIS. In contrast, Mordecai and Dabke [8] reported poor quality evidence supporting the use of an exercise program in the management of AIS, and a Cochrane review conducted by Romano et al. [12] revealed a lack of high-quality evidence to recommend the use of scoliosis-specific exercises to reduce the progression of AIS.

All previous reviews were focused on the effects of exercise on the Cobb angle and brace prescriptions in patients with AIS [8, 10, 53]. However, in the present review, in addition to the Cobb angle, other surface spinal deformities such as trunk rotation, thoracic kyphosis, lumbar lordosis, average lateral deviation, and quality of life were measured. Moreover, in previous reviews, only Romano et al. [12] performed a meta-analysis of the effects of scoliosis-specific exercises to reduce the progression of AIS.

In the present review, 3 studies compared SEAS.02 exercise with a control group and reported that SEAS.02 exercises were superior to control conditions for reducing spinal deformities and the progression of scoliosis [47, 48, 51]. Another 6 studies included in the present review compared 6 different exercise protocols with traditional spinal exercises. All these studies reported significant reduction of spinal deformities and improvement in quality of life as compared with traditional spinal exercise [43–46, 49, 50].

This review had several limitations. Inclusion criteria were not well defined in the included studies, and the majority of the included studies were nonrandomized. Additionally, lack of blinding, lack of concealed allocation, and variations in exercise protocols are significant limitations in the current published literature. Moreover, different types of exercise have different intensities and may induce different effects, and the presence of heterogeneity in exercise protocols prevents conclusive results. For example, the total intervention duration varied between 2 weeks [43] and 4 months [46] and sample size in the included studies varied from 30 [44] to 252 [50]. Another limitation of the present review was the inclusion of only studies published in English, which might have created some selection bias. In addition, most of the included studies did not clarify what types of exercises are found in the usual care.

5. Conclusions

Moderate-quality evidence suggests that an exercise program is superior to controls in reducing the Cobb angle, angle of trunk rotation, thoracic kyphosis angle, and lumbar lordosis angle and improving the quality of life in patients with AIS; and the low-quality evidence suggests that an exercise program is superior to controls in reducing average lateral deviation in patients with AIS. However, the presence of heterogeneity in exercise protocols and poor methodological quality limit the validity of these results.

Conflict of Interests

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

Acknowledgment

The project was full financially supported by King Saud University, through Vice Deanship of Research Chairs, Rehabilitation Research Chair.

References

[1] M. C. Hawes and J. P. O'Brien, "The transformation of spinal curvature into spinal deformity: pathological processes and implications for treatment," *Scoliosis*, vol. 1, no. 1, article 3, 2006.

[2] M. A. Asher and D. C. Burton, "Adolescent idiopathic scoliosis: natural history and long term treatment effects," *Scoliosis*, vol. 1, article 2, 2006.

[3] S. L. Weinstein, L. A. Dolan, J. C. Cheng, A. Danielsson, and J. A. Morcuende, "Adolescent idiopathic scoliosis," *The Lancet*, vol. 371, no. 9623, pp. 1527–1537, 2008.

[4] S. L. Weinstein and I. V. Ponseti, "Curve progression in idiopathic scoliosis," *The Journal of Bone & Joint Surgery Series A*, vol. 65, no. 4, pp. 447–455, 1983.

[5] E. Ascani, P. Bartolozzi, C. A. Logroscino et al., "Natural history of untreated idiopathic scoliosis after skeletal maturity," *Spine*, vol. 11, no. 8, pp. 784–789, 1986.

[6] M. S. Goldberg, N. E. Mayo, B. Poitras, S. Scott, and J. Hanley, "The Ste-Justine adolescent idiopathic scoliosis cohort study. Part II. Perception of health, self and body image, and participation in physical activities," *Spine*, vol. 19, no. 14, pp. 1562–1572, 1994.

[7] M. S. Wong, J. C. Y. Cheng, T. P. Lam et al., "The effect of rigid versus flexible spinal orthosis on the clinical efficacy and acceptance of the patients with adolescent idiopathic scoliosis," *Spine*, vol. 33, no. 12, pp. 1360–1365, 2008.

[8] S. C. Mordecai and H. V. Dabke, "Efficacy of exercise therapy for the treatment of adolescent idiopathic scoliosis: a review of the literature," *European Spine Journal*, vol. 21, no. 3, pp. 382–389, 2012.

[9] S. L. Weinstein, L. A. Dolan, J. G. Wright, and M. B. Dobbs, "Effects of bracing in adolescents with idiopathic scoliosis," *The New England Journal of Medicine*, vol. 369, no. 16, pp. 1512–1521, 2013.

[10] S. Negrini, C. Fusco, S. Minozzi, S. Atanasio, F. Zaina, and M. Romano, "Exercises reduce the progression rate of adolescent idiopathic scoliosis: results of a comprehensive systematic review of the literature," *Disability and Rehabilitation*, vol. 30, no. 10, pp. 772–785, 2008.

[11] S. Negrini, J.-C. De Mauroy, T. B. Grivas et al., "Actual evidence in the medical approach to adolescents with idiopathic scoliosis," *European Journal of Physical and Rehabilitation Medicine*, vol. 50, no. 1, pp. 87–92, 2014.

[12] M. Romano, S. Minozzi, F. Zaina et al., "Exercises for adolescent idiopathic scoliosis," *Spine*, vol. 38, no. 14, pp. E883–E893, 2013.

[13] S. Negrini, G. Antonini, R. Carabalona, and S. Minozzi, "Physical exercises as a treatment for adolescent idiopathic scoliosis. A systematic review," *Pediatric Rehabilitation*, vol. 6, no. 3-4, pp. 227–235, 2003.

[14] J. Higgins and S. Green, *Cochrane Handbook for Systematic Reviews of Interventions Version 5.0.0*, The Cochrane Collaboration, Chichester, UK, 2008.

[15] J. Cohen, "Quantitative methods in psychology," *Psychological Bulletin*, vol. 112, pp. 155–159, 1992.

[16] Biostat, *Comprehensive Meta-Analysis Version 2.0*, [Computer software], Biostat, Englewood, NJ, USA, 2014.

[17] C. G. Maher, C. Sherrington, R. D. Herbert, A. M. Moseley, and M. Elkins, "Reliability of the PEDro scale for rating quality of randomized controlled trials," *Physical Therapy*, vol. 83, no. 8, pp. 713–721, 2003.

[18] J. A. Wallis and N. F. Taylor, "Pre-operative interventions (non-surgical and non-pharmacological) for patients with hip or knee osteoarthritis awaiting joint replacement surgery—a systematic review and meta-analysis," *Osteoarthritis and Cartilage*, vol. 19, no. 12, pp. 1381–1395, 2011.

[19] J. P. T. Higgins and D. G. Altman, "Assessing risk of bias in included studies," in *Cochrane Handbook for Systematic Reviews of Interventions Version 5.0.1*, J. P. T. Higgins and S. Green, Eds., The Cochrane Collaboration, Oxford, UK, 2008.

[20] D. Atkins, D. Best, P. A. Briss et al., "Grading quality of evidence and strength of recommendations," *The British Medical Journal*, vol. 328, no. 7454, p. 1490, 2004.

[21] J. P. T. Higgins, S. G. Thompson, J. J. Deeks, and D. G. Altman, "Measuring inconsistency in meta-analyses," *British Medical Journal*, vol. 327, no. 7414, pp. 557–560, 2003.

[22] B. Falk, W. A. Rigby, and N. Akseer, "Adolescent idiopathic scoliosis: the possible harm of bracing and the likely benefit of exercise," *Spine Journal*, vol. 15, no. 1, pp. 209–210, 2015.

[23] M. Płaszewski, I. Cieśliński, P. Kowalski, A. Truszczyńska, and R. Nowobilski, "Does scoliosis-specific exercise treatment in adolescence alter adult quality of life?" *The Scientific World Journal*, vol. 2014, Article ID 539671, 10 pages, 2014.

[24] K. Fabian and K. Rożek-Piechura, "Exercise tolerance and selected motor skills in young females with idiopathic scoliosis treated with different physiotherapeutic methods," *Ortopedia Traumatologia Rehabilitacja*, vol. 16, no. 5, pp. 507–522, 2014.

[25] V. L. dos Santos Alves, R. J. Alves da Silva, and O. Avanzi, "Effect of a preoperative protocol of aerobic physical therapy on the quality of life of patients with adolescent idiopathic scoliosis: a randomized clinical study," *The American Journal of Orthopedics (Belle Mead, NJ)*, vol. 43, pp. E112–E116, 2014.

[26] N. Pugacheva, "Corrective exercises in multimodality therapy of idiopathic scoliosis in children—analysis of six weeks efficiency—pilot study," *Studies in Health Technology and Informatics*, vol. 176, pp. 365–371, 2012.

[27] D. Czaprowski, T. Kotwicki, R. Biernat, J. Urniaz, and A. Ronikier, "Physical capacity of girls with mild and moderate idiopathic scoliosis: influence of the size, length and number of curvatures," *European Spine Journal*, vol. 21, no. 6, pp. 1099–1105, 2012.

[28] M. W. Morningstar, "Outcomes for adult scoliosis patients receiving chiropractic rehabilitation: a 24-month retrospective analysis," *Journal of Chiropractic Medicine*, vol. 10, no. 3, pp. 179–184, 2011.

[29] F. Canavese and A. Kaelin, "Adolescent idiopathic scoliosis: indications and efficacy of nonoperative treatment," *Indian Journal of Orthopaedics*, vol. 45, no. 1, pp. 7–14, 2011.

[30] Y. T. Tsai, C.-P. Leong, Y.-C. Huang et al., "The electromyographic responses of paraspinal muscles during isokinetic exercise in adolescents with idiopathic scoliosis with a Cobb's angle less than fifty degrees," *Chang Gung Medical Journal*, vol. 33, no. 5, pp. 540–550, 2010.

[31] S. D. Glassman, L. Y. Carreon, C. I. Shaffrey et al., "The costs and benefits of nonoperative management for adult scoliosis," *Spine*, vol. 35, no. 5, pp. 578–582, 2010.

[32] M.-L. B. Lenssinck, A. C. Frijlirik, M. Y. Berger, S. M. A. Bierma-Zeinstra, K. Verkerk, and A. P. Verhagen, "Effect of bracing and other conservative interventions in the treatment of idiopathic scoliosis in adolescents: a systematic review of clinical trials," *Physical Therapy*, vol. 85, no. 12, pp. 1329–1339, 2005.

[33] K. Ferrari, P. Goti, A. Sanna et al., "Short-term effects of bracing on exercise performance in mild idiopathic thoracic scoliosis," *Lung*, vol. 175, no. 5, pp. 299–310, 1997.

[34] H. R. Weiss, "The effect of an exercise program on vital capacity and rib mobility in patients with idiopathic scoliosis," *Spine*, vol. 16, no. 1, pp. 88–93, 1991.

[35] J. M. Shneerson and R. Madgwick, "the effect of physical training on exercise ability in adolescent idiopathic scoliosis," *Acta Orthopaedica*, vol. 50, no. 3, pp. 303–306, 1979.

[36] A. L. Nachemson, J. C. Bjure, L. G. Grimby, and M. Lindh, "Physical fitness in young women with idiopathic scoliosis before and after an exercise program," *Archives of Physical Medicine and Rehabilitation*, vol. 51, no. 2, pp. 95–98, 1970.

[37] S. Schreiber, E. C. Parent, D. M. Hedden, M. Moreau, D. Hill, and E. Lou, "Effect of Schroth exercises on curve characteristics and clinical outcomes in adolescent idiopathic scoliosis: protocol for a multicentre randomised controlled trial," *Journal of Physiotherapy*, vol. 60, no. 4, p. 234, 2014.

[38] H. R. Weiss and G. Weiss, "Curvature progression in patients treated with scoliosis in-patient rehabilitation—a sex and age matched controlled study," *Studies in Health Technology and Informatics*, vol. 91, pp. 352–356, 2002.

[39] S. Negrini, S. Donzelli, M. Lusini, S. Minnella, and F. Zaina, "The effectiveness of combined bracing and exercise in adolescent idiopathic scoliosis based on SRS and SOSORT criteria: a prospective study," *BMC Musculoskeletal Disorders*, vol. 15, article 263, 2014.

[40] C. Lewis, R. Diaz, G. Lopez, N. Marki, and B. Olivio, "A preliminary study to evaluate postural improvement in subjects with scoliosis: active therapeutic movement version 2 device and home exercises using the Mulligan's mobilization-with-movement concept," *Journal of Manipulative and Physiological Therapeutics*, vol. 37, pp. 502–509, 2014.

[41] L. Rivett, A. Stewart, and J. Potterton, "The effect of compliance to a Rigo System Cheneau brace and a specific exercise programme on idiopathic scoliosis curvature: a comparative study: SOSORT 2014 award winner," *Scoliosis*, vol. 9, article 5, 2014.

[42] H. Hundozi-Hysenaj, I. B. Dallku, A. Murtezani, and S. Rrecaj, "Treatment of the idiopathic scoliosis with brace and physiotherapy," *Nigerian Journal of Medicine*, vol. 18, no. 3, pp. 256–259, 2009.

[43] M. Monticone, E. Ambrosini, D. Cazzaniga, B. Rocca, and S. Ferrante, "Active self-correction and task-oriented exercises reduce spinal deformity and improve quality of life in subjects with mild adolescent idiopathic scoliosis. Results of a randomised controlled trial," *European Spine Journal*, vol. 23, no. 6, pp. 1204–1214, 2014.

[44] T. Kuru, İ. Yeldan, E. E. Dereli, A. R. Özdinçler, F. Dikici, and İ. Colak, "The efficacy of three-dimensional Schroth exercises in adolescent idiopathic scoliosis: a randomised controlled clinical trial," *Clinical Rehabilitation*, 2015.

[45] A. A. Diab, "The role of forward head correction in management of adolescent idiopathic scoliotic patients: a randomized controlled trial," *Clinical Rehabilitation*, vol. 26, no. 12, pp. 1123–1132, 2012.

[46] D. K. Noh, J. S. You, J. H. Koh et al., "Effects of novel corrective spinal technique on adolescent idiopathic scoliosis as assessed by radiographic imaging," *Journal of Back and Musculoskeletal Rehabilitation*, vol. 27, pp. 331–338, 2014.

[47] S. Negrini, A. Negrini, M. Romano, N. Verzini, A. Negrini, and S. Parzini, "A controlled prospective study on the efficacy of SEAS.02 exercises in preparation to bracing for idiopathic scoliosis," *Studies in Health Technology and Informatics*, vol. 123, pp. 519–522, 2006.

[48] S. Negrini, A. Negrini, M. Romano, N. Verzini, A. Negrini, and S. Parzini, "A controlled prospective study on the efficacy of SEAS.02 exercises in preventing progression and bracing in mild idiopathic scoliosis," *Studies in Health Technology and Informatics*, vol. 123, pp. 523–526, 2006.

[49] H.-R. Weiss and R. Klein, "Improving excellence in scoliosis rehabilitation: a controlled study of matched pairs," *Pediatric Rehabilitation*, vol. 9, no. 3, pp. 190–200, 2006.

[50] H. R. Weiss, I. Heckel, and C. Stephan, "Application of passive transverse forces in the rehabilitation of spinal deformities: a randomized controlled study," *Studies in Health Technology and Informatics*, vol. 88, pp. 304–308, 2002.

[51] S. Negrini, F. Zaina, M. Romano, A. Negrini, and S. Parzini, "Specific exercises reduce brace prescription in adolescent idiopathic scoliosis: a prospective controlled cohort study with worst-case analysis," *Journal of Rehabilitation Medicine*, vol. 40, no. 6, pp. 451–455, 2008.

[52] R. J. Feise and J. M. Menke, "Functional rating index: literature review," *Medical Science Monitor*, vol. 16, no. 2, pp. RA25–RA36, 2010.

[53] C. Fusco, F. Zaina, S. Atanasio, M. Romano, A. Negrini, and S. Negrini, "Physical exercises in the treatment of adolescent idiopathic scoliosis: an updated systematic review," *Physiotherapy Theory and Practice*, vol. 27, no. 1, pp. 80–114, 2011.