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Virtual environment for lower-extremities training

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Abstract

This study proposed virtual reality (VR) as a modality of lower-extremities training. A kinematic model of a human body and a corresponding virtual figure were developed, in order to visualize the movements of the subject in a real-time virtual environment on a large display, which represented a virtual mirror. An optical system with active markers was used to assess the movements of a training subject. A preliminary investigation was conducted with a group of healthy male subjects, who performed the stepping-in-place test by tracking the movements of the reference virtual figure, which represented a virtual instructor. Both figures were shown in the virtual mirror at the same time from the desired angle of view. Four stepping tasks featuring different cadences and hip angles were performed, with difficulty levels ranging from easy to demanding. The results obtained included basic kinematic and temporal parameters, which provided quantitative measures of a subject's adaptation to the virtual training environment, and thereby justifying the feasibility of the virtual mirror as a useful system in lower-extremities training applications.

Keywords: Virtual reality; Virtual mirror; Visual biofeedback; Lower-extremities training; Virtual rehabilitation

1. Introduction

Virtual reality (VR) is a powerful tool in a rehabilitation environment, providing the patients with repetitive practice, feedback information, and motivation to endure practice. Further advantages of the VR rehabilitative systems include various possibilities of adaptation to the patient's capabilities, reprogrammable virtual tasks, and extended measurability. The number of studies and experimental applications exploiting VR in the rehabilitation environment has been increasing rapidly over the last few years; however, the majority of these endeavors have focused on the upper extremities [1]. The present study investigates the use of the VR as a means of augmenting visual biofeedback information in lower-extremities training. We propose the use of a virtual mirror; this is a large screen in front of which the subject performs the lower-extremity movements (Fig. 1). The subject can see two figures in a three-dimensional (3D) virtual environment in the virtual mirror, from the desired viewing angle. The solid figure represents the training subject, the movements of which correspond to those of the subject in real time. The transparent figure represents the virtual instructor. The two figures are superimposed. The movements of the instructor are preprogrammed, and are obtained through learning trials with a healthy subject. The task of the training subject is to follow the movements of the virtual instructor as accurately as possible, so that both figures are closely overlaid throughout the duration of the lower-extremities training.

We explored the feasibility and applicability of the virtual mirror by performing a stepping-in-place (SIP) experiment. SIP test has a relatively long history in the clinical environment. More than 50 years ago, it was first used for detecting peripheral vestibular dysfunction [2]. During the last decade, the test has also entered the rehabilitation environment. The SIP test has been applied to stroke patients [3], patients with Parkinson disease [4], and amputees. These previous studies have focused either on evaluating the test as an indicator of a disease, or on assessing patients' strategies

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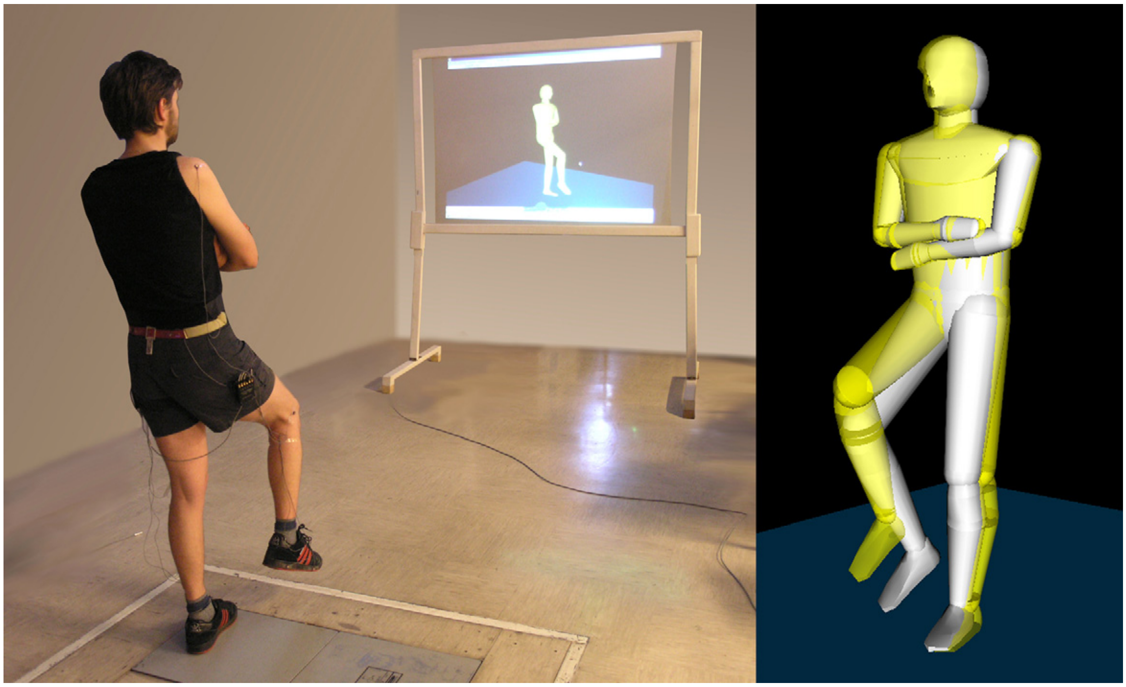

Fig. 1. Virtual mirror: a large screen showing movements of the subject in real time (left), superimposed virtual trainee and virtual instructor enlarged (right).

for overcoming limited movement in their lower extremities. Combined with VR, the SIP test can also be considered as a modality of lower-extremities training during rehabilitation. This test cannot replace gait training and/or analysis; however, similar reciprocal rhythmic movement patterns can be observed during both locomotor activities. The SIP test allows the assessment of basic temporal parameters, such as stance and swing phase, double-stance phase, and step frequency. Symmetry of lower-extremity movements can also be observed. In addition, stability and balance can be studied during the SIP test.

The aim of this preliminary study was to investigate how VR can provide a methodology which allows standardization of the clinical test during lower-extremities rehabilitation by visual biofeedback, and to demonstrate the VR applicability in quantification, recording, and monitoring in the clinical setting. To achieve this, the SIP test was performed in a group of healthy adults at different cadences and different heights of knee-joint lifting, where VR adaptation abilities were evaluated among the subjects. We studied kinematic and temporal adaptation to the virtual instructor by investigating the subjects' consistency (adopting a steady stepping pattern) and accuracy (maintaining a pattern similar to the virtual instructor's).

2. Methods and measurements

2.1. Body kinematics

A simplified kinematic model of the human body was developed using vector parameters [5], in order to visualize the movements of the subject in the virtual mirror. The model comprised eight rigid segments representing parts of the body, which were connected by rotational joints. The head, arms, and torso were represented as a single segment (HAT), which was connected to the pelvis segment by a spherical joint with three degrees of freedom (DOF). The hip joints were also represented as three DOF spherical joints, while the knees and ankles were simplified as one DOF hinge joints, with axes of rotation parallel to each other. In this way, the segments representing the thigh, shank, and foot, all moved in one plane, the normal vector of which had the same direction as the axes of rotation in the knee and ankle joints. The pelvic segment represented the base segment, which moved in space freely and therefore had six DOF.

In order to obtain the values of the joint variables, 11 active markers were placed on the skin over anatomical prominences of the human body. We aimed to keep the number of markers as low as possible, in order to facilitate a quick and efficient experimental procedure. The positions of the markers were measured using the OPTOTRAK (Northern Digital Inc.) system with a 70-Hz sample rate. The pose of the pelvic segment was determined from the three markers placed over the posterior superior iliac spines (PSIS) and lower edge of the sacrum. In addition, the positions of the PSIS and sacral markers were used to calculate the centers of the hip joints [6]. The midpoint between the hip joints was considered as the pelvic center. One marker was placed on the skin at the approximate center of rotation of each metatarsophalangeal (MTP) joint, knee, ankle, and shoulder in order to determine the positions of the underlying joints.

The position of the body was represented by the position coordinates of the pelvic center. It was expressed as a percentage of the subject's body height (BH) in order to allow comparisons among subjects. The joint angles were calculated from the vectors connecting neighboring joints, which represented body segments. Vector cross-product operations were applied to the consecutive body-segment vectors to obtain the segment coordinate systems, and the corresponding axes and angles of joint rotations. The proposed angle-determination method used a reduced number of markers per segment (less than three), which caused computational issues when the joint was in (or near) the singular pose (angle values 0, $\pm\pi$). The vectors representing consecutive segments in the singular pose were collinear, making it impossible to calculate the rotation axis. During SIP training, singular poses occurred relatively often in the knee joint. Hence, the observation mentioned above could not be neglected. The following set of equations overcame the problem by calculating the axis of knee and ankle rotation in three different ways, shown here for the left leg only (Fig. 2).

The direction of the knee rotation axis (y_{LL}) was obtained by performing cross-product operations upon the thigh and shank unit vectors:

$$y_{LL} = z_{LT} \times z_{LL}. \tag{1}$$

According to our simplified model, the direction of the ankle rotation axis was the same as the knee axis, and was obtained from

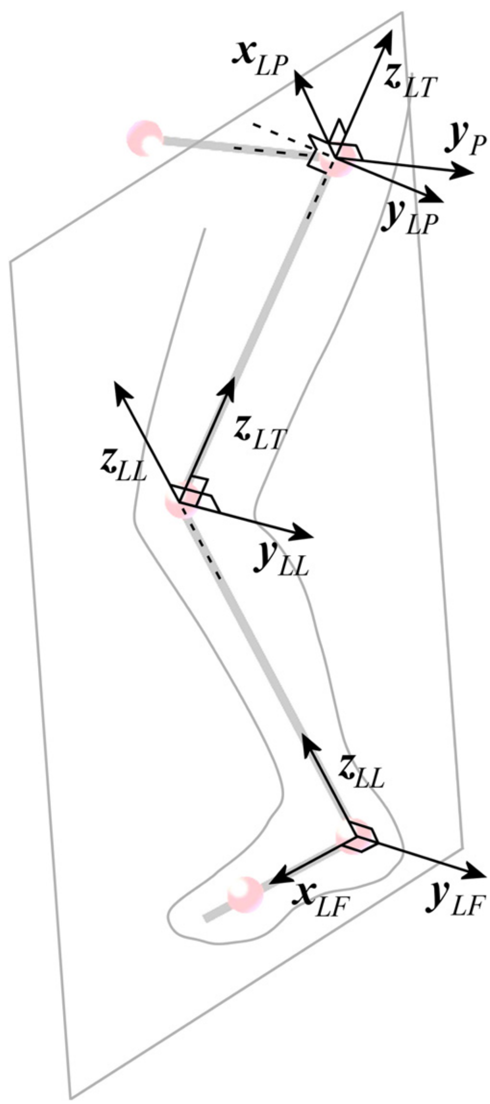

Fig. 2. Kinematic representation of the left leg showing vectors needed for knee and ankle axes computation.

the shank and foot unit vectors:

$$y_{LF} = z_{LL} \times x_{LF}. \tag{2}$$

The third knee-axis calculation assumed that the hip joint had only two DOF, with the longitudinal rotation about the thigh axis omitted (y_P was a unit vector connecting both hip joints):

$$x_{LP} = y_P \times z_{LT}, \tag{3}$$

$$y_{LP} = z_{LT} \times x_{LP}. \tag{4}$$

For each of the three axis vectors, a weighting factor was considered:

$$f_{LL} = |y_{LL}|, \tag{5}$$

$$f_{LF} = \begin{Bmatrix} (1-|y_{LL}|)\cdot|y_{LF}| & ; & (1-|y_{LL}|)>0 \\ 0 & ; & (1-|y_{LL}|)<0 \end{Bmatrix}, \tag{6}$$

$$f_{LP} = \begin{Bmatrix} 1-(f_{LL}+f_{LF}) & ; & (1-(f_{LL}+f_{LF}))>0 \\ 0 & ; & (1-(f_{LL}+f_{LF}))<0 \end{Bmatrix}. \tag{7}$$

Finally, the new axis of knee and ankle rotation was obtained as a weighted sum of unit vectors:

$$y'_{LL} = f_{LL}y_{LL} + f_{LF}y_{LF} + f_{LP}y_{LP}. \tag{8}$$

In poses where the knee rotation axis was well pronounced, the significance of the first weighting factor was prevalent in Eq. (8). Near the singularity of the knee joint, the first weighting factor moved towards zero whereas the second factor increased if the ankle axis was well pronounced. In the worst case, when the ankle was also approaching singularity (in the case of strong plantar flexion atop of the fully extended knee), the third weighting factor gained significance. In this case, the hip joint was considered as having only two DOF, thus preventing any longitudinal rotation about the thigh axis. The described procedure overcame the unpredictable behavior of the knee joints in and near the singular poses, and was found to be effective with regard to the requirements of the virtual-mirror design.

2.2. Virtual mirror

Kinematic data, which were calculated from the OPTOTRAK measurements, were used to animate the motion of the human model in VR. The virtual environment consisted of a simplified human figure placed on a semi-transparent plane. The figure was made up of eight rigid segments (that is, the HAT, pelvis, thighs, shanks, and feet), which matched the described kinematic model, and imitated the shape of the human body [7,8]. The ratios between the segment lengths were based on statistical anthropometry [9]. The movements of the figure corresponded to the movements of the subject at a 35-Hz refresh rate without detectable lag, thereby enabling a convincing perception of the virtual mirror. We used VRML 2.0 (Virtual Reality Modeling Language) which exploits the fast built-in functions of the graphics processor to visualize the movements of the figure. Kinematic data were fed into the VRML model in the form of a four-element vector for each joint; the first three elements represented the x, y, and z components of the rotation axis vector, while the last element was the angle of rotation, expressed in radians, thereby forming a standard axis-angle notation. It was not necessary to compute the direct kinematics for proper visualization of the model, as this could be achieved by the graphics-processing unit when

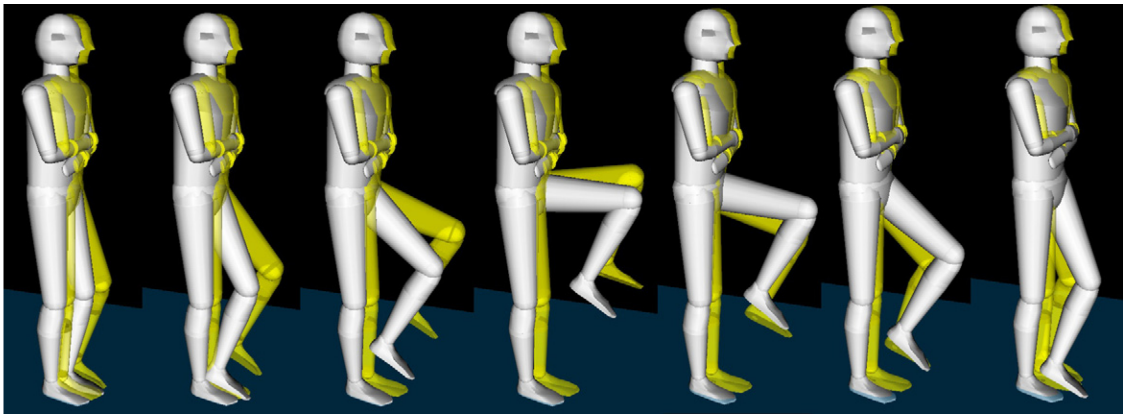

Fig. 3. The two superimposed figures representing the subject (solid) tracking the stepping movements of the virtual instructor (semi-transparent).

provided with an adequate VRML tree structure of the model and axis-angle vectors. The position of the virtual figure corresponded to the subject's pelvic position, which was expressed as the percentage (%) of BH and multiplied by the height of the virtual figure. This enabled the use of the same virtual figure for all subjects.

Prior to the SIP training, a calibration of the virtual figure was performed. This was achieved by instructing the subject, with markers in place, to remain still for 3 s in a quiet stance, with the knees fully extended and the feet oriented in parallel. The median values of the joint angles during the stance were regarded as offsets to the initial pose. All angles were set to 0 in the initial pose of the virtual figure; thus, when using kinematic data to visualize the subject's movements in the virtual mirror, offset-compensated values were assigned to the virtual figure. The median value of the pelvic position during calibration was regarded as the origin point.

During the SIP training, the subject would see another figure besides his own in the virtual mirror. The additional figure, which was semi-transparent and of different color, represented the virtual instructor. The two figures were superimposed. The motion of the virtual instructor was preprogrammed with stepping movements, and presented a reference that the subject was instructed to follow (Fig. 3). Ideally, both figures would be perfectly aligned at all times, thereby indicating that the subject was performing the SIP simultaneously with the virtual instructor. The semi-transparency allowed the subjects to see their figure even when it was behind the virtual instructor. In order to provide the subject with the desired view of his performance, it was possible to set the viewing angle and distance of the virtual camera arbitrarily. It was also possible to switch the image in the virtual mirror between the real and mirror views.

2.3. SIP training tasks

Assessment of the subjects' ability to follow the SIP movements of the virtual instructor was undertaken at different cadences and different heights of knee-joint lifting. The movements of the virtual instructor were obtained by capturing the steps of a healthy male subject (aged 25 years), who was well familiarized with the virtual mirror. After capturing and averaging a series of steps, a single step was isolated, and was adjusted for symmetry and smoothness; slight deviations from the desired reference maximal angle values in both sides were compensated by multiplying all the samples with appropriate constants (values ≈ 1) in order to achieve exact reference values. A continuous stepping motion was achieved by programming the virtual instructor with a number of repetitions of the selected step. Smooth transition between steps was ensured by low-pass filtering the data in the step transition stage. After establishing reasonable physical limits of performance, four tasks were proposed featuring different cadences and hip angles as shown in Table 1. The number of step repetitions was set as 30 for all tasks. All subjects performed the described tasks in the same order. Each subject also completed a single trial of each task before the actual measurements took place, in order to become familiarized with the virtual mirror. The angle of view for the SIP experiment was set to a non-mirror 3D view as seen in Fig. 3, based on the optimal visibility of the lower-extremities movements and was the same for all subjects.

During the test, the following parameters were recorded: the rotation axes vectors and corresponding angles of the HAT, pelvis, thighs, shanks, and feet, and the position of the pelvis. These data were sufficient to replay the SIP performance of the subject later on, with or without the virtual instructor being included in the replay. Additionally, all marker positions were stored for subsequent comparison and verification of the model behavior.

A test group for the SIP experiment consisted of 10 healthy male subjects (aged 23–39 years; mean value (MV) = 28.5 years; standard deviation (S.D.) = 4.7 years). None of the subjects had a medical history of significant lower limb injuries or any other medical condition that would impair movement. All subjects gave informed consent to participate in the experiment.

Table 1
Hip angles in degrees (°) and cadences in beats per minute (BPM) featured in all SIP tasks

	SIP task			
	1	2	3	4
Hip angle (°)	45	90	90	45
Cadence (BPM)	60	60	90	120

3. Results

Several basic SIP parameters were chosen from the measured data for statistical analysis and for quantitative comparison among the subjects in the test group. These included two kinematic parameters; the maximal angles for the hip and knee joints achieved in each step, and two temporal parameters, the swing-phase duration and SIP period duration (time between two consecutive MTP joint rises).

Fig. 4(a and b) shows the kinematic parameters for all tasks and all subjects. The solid horizontal line represents the reference angle of the virtual instructor. The boxes indicate the 25th percentile, median value, and the 75th percentile in step, while the error bars show the maximal and minimal angles measured in the group of 10 subjects. Grey bands indicate $\pm 10^{\circ}$ deviation. One-way analysis of variance was used to compare the steps within each task;

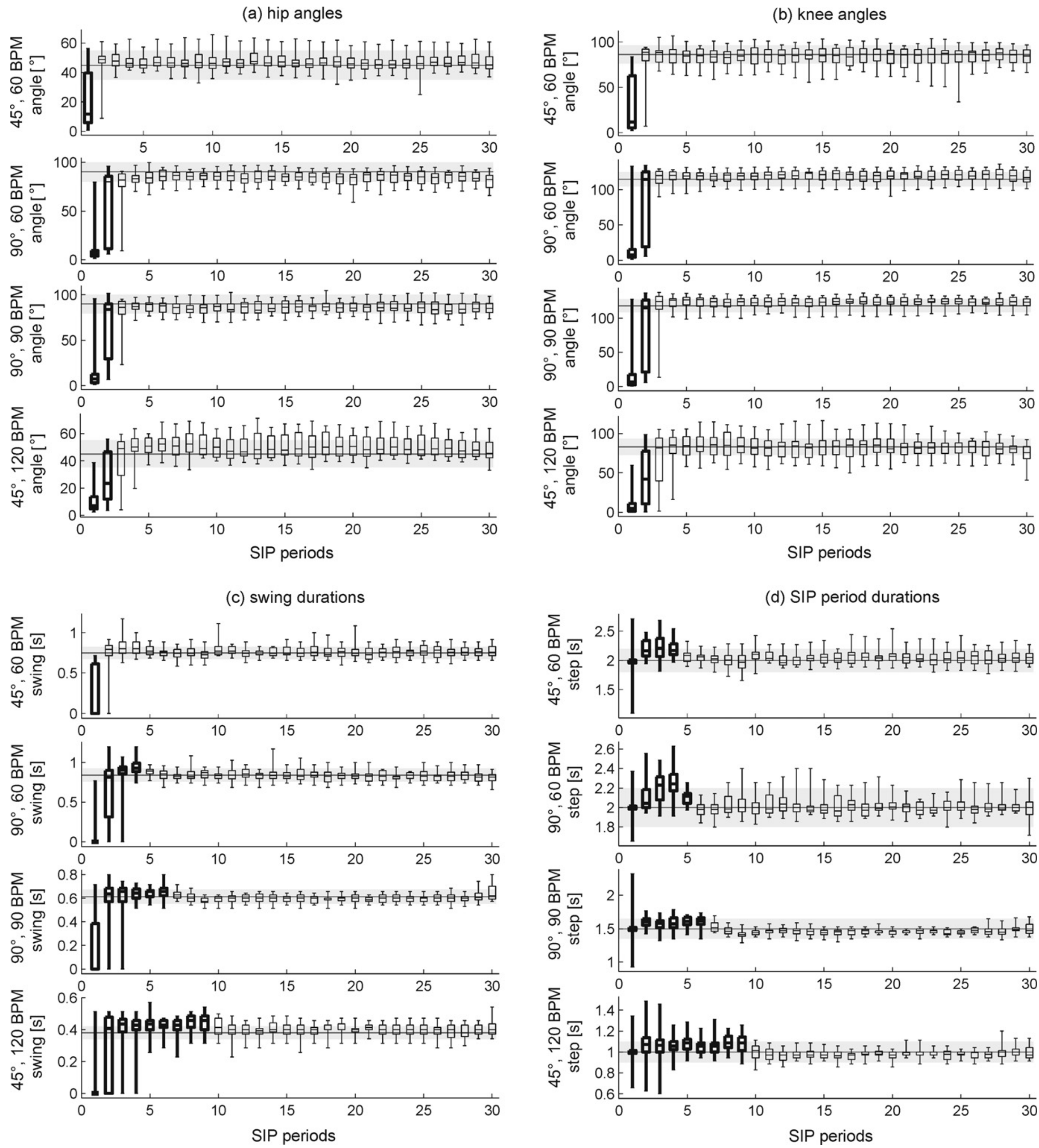

Fig. 4. SIP parameters (a: hip angle, b: knee angle, c: swing time, d: SIP period time) for SIP periods 1–30, showing 25th percentile, median value, 75th percentile, maximal, and minimal value in each SIP period.

bold lines indicate the steps that are significantly different from the steps that follow ($p < 0.001$ for all tasks). We considered these steps as adaptation time, needed by the subjects to become consistent. The number of steps needed to adapt to the virtual instructor increased with faster cadences and greater hip angle values. The accuracy of kinematic adaptation was evaluated first visually by replaying the subjects' performance and focusing on a particular parameter; we observed that mean values of both parameters were within $\pm 10^{\circ}$ band. This was considered as accurate kinematic adaptation to the virtual instructor. MVs in all steps following the adaptation time were within this range; however, it was observed that, while remaining within this acceptable range, the group's response tended to overshoot the hip angle reference in the first task, and especially in the fourth task, but did not reach that of the reference in the second and third tasks ($p < 0.001$ for both observations). Knee angles were significantly greater than the reference in the second and third tasks ($p < 0.001$).

Fig. 4(c and d) shows the temporal parameters (significance levels for different swing durations: $p < 0.001$ for the first three tasks, $p = 0.015$ for the last task; SIP period durations: $p = 0.002$ for the first task, $p = 0.035$ for the second task, and $p < 0.001$ for the last two tasks). A $\pm 10\%$ range of the reference swing and SIP period durations (marked as grey bands) was considered as accurate temporal adaptation. Again, MVs in all steps that followed the adaptation time were within this range. While remaining within this acceptable range, the subjects exhibited significantly shorter swing and SIP period durations than the reference in the third task ($p < 0.001$). In the fourth task, swing durations were longer and SIP period durations were shorter than the reference durations ($p < 0.001$). In order to distinguish between the swing and stance phase in each SIP period, vertical position data from the markers placed over the MTP joints were considered. Again, an increase in the number of steps needed to adapt was noticed at faster cadences and larger hip angles. Swing-phase durations of 0 s, when interpreted together with the hip angles, indicated that some of the subjects made attempts to catch up with the virtual instructor during the first few steps; however, their movements did not manifest as articulated steps, as their feet were still touching the ground. Fig. 4 also shows that all subjects were able to track the movements of the virtual instructor without missing any steps after the adaptation time during any task.

4. Discussion

4.1. Interpretation of results

The four tasks for which results are given were established according to the subjects' expected physical abilities, such that the first task was easy to perform for any subject, while the last task could be described as demanding. This was proven to be accurate by the test group. Less demanding tasks than those proposed in our study should be introduced in clinical practice.

The adaptation of the subjects to the virtual instructor was evaluated by determining the time needed to achieve consistency, and accuracy of kinematic and temporal adaptation. According to Fig. 4 the slower cadences and smaller angles presented easier tasks for the subjects, although applying the same order of task performance, starting with the easiest and finishing with the most difficult task might have suggested some improvement by learning. This only consolidated the impression that healthy subjects could adapt to the virtual mirror quickly; however, kinematic adaptation was generally achieved sooner than temporal adaptation, especially in the more demanding tasks. Whether or not this is a general or methodology-related phenomenon cannot be concluded from this study. Further experiments should be proposed to address this issue, exploring different tasks and complementary methods such as combining visual and audio biofeedback.

The results indicated that healthy subjects were able to perform a rather complex task in the virtual environment which included coordinated balanced motion of the whole body, while adapting to the reference movements presented in a form of a virtual mirror and virtual instructor.

4.2. Model performance

The kinematic model used for the VR visualization had 19 DOF. The kinematics of the upper body during the SIP test were of little interest in our study, and were therefore simplified to a single HAT segment, thus reducing the number of active markers required. Another major simplification was made by representing the ankles as one DOF joints, although this is not uncommon in human modeling [10,11]. This was made possible by adopting the pelvis as the base segment of the stepping figure. In this way, the position error occurring at the ankle joints affected only the feet of the virtual figure. As the foot segments were relatively short, the overall position error was within the acceptable margins for VR visualization. The knees also exhibited motion with at least three DOF; however, flexion/extension in the sagittal plane was predominant, making a one DOF hinge joint a satisfactory representation of the knee joint [11,12]. Together with the three DOF hip joints and the pelvis–HAT linkage, the motion of the virtual figure proved to be a convincing representation of the actual lower body movements of the subjects. The computational issues caused by the singular poses in the knee joints were handled effectively by the procedure described in Section 2, resulting in smooth and natural appearing motion. Without exploiting this simple procedure, the model exhibited sudden knee-axis shifts and inadequate poses in an unpredictable manner when approaching singularities. By contrast, considering hips as two DOF joints in all poses resulted in unsatisfactory model behavior in terms of the ability to realize the actual poses of the subjects during the SIP training. The complexity level of the presented model is sufficient to translate into the virtual environment the patient-specific deformities in the lower extremities, such as adduction contractures or excessive internal rotation of the femur. However, the simplification of the upper body could distort postural deformities of the spine and torso.

One of the main concerns in our virtual environment was the use of the same human figure model for all subjects in order to enable comparison. Transferring the angle values from reality to VR was a rapid process; however, this was not the case with the position values. The differences in BH and anthropometric data among subjects made normalization of the position values necessary. By dividing the actual position data from OPTOTRAK (expressed in mm) by the BH and multiplying it by the virtual figure height, the influence of a subject's BH was eliminated. This was achieved conjointly with the calibration procedure through which the pelvic center point was obtained. Calibrating the subject and virtual figure also compensated for imprecisely placed markers, and ensured that the joint angles and the position of the subject were properly assigned to the virtual figure.

Conducting the proposed experiment with a group of healthy subjects presented no major concerns regarding their safety and their ability to tolerate the procedure. However, several issues remain concerning the transfer of virtual training to patients. The time needed to prepare the subject for the test (i.e. to place the active markers, setup the system to avoid marker occlusions and give instructions) is highly disproportionate to the actual duration of the training. While all four tasks, with calibration and breaks included, did not take more than 5–10 min per subject, the average time needed to fit the subjects with non-impeding markers, OPTOTRAK strober units, and wires was about 30 min. During the preparation period, the subjects were asked to stand still for most of the time, to enable proper setup. This might not be appropriate for patients undergoing lower-extremity rehabilitation. Furthermore, the calibration procedure should be altered for patients with joint contractures who cannot stand still. However, human motion-assessment techniques utilizing computer vision are evolving to become promising complements to existing optical measurements [13,14]. A computer-vision approach combined with accelerometers and gyroscopes would be less inconvenient for the subjects, and could render the lower-extremities training in VR more suitable for use with patients. Further modalities, such as robotic devices, passive exoskeletons, or functional electrical stimulation should be considered when using the virtual mirror for lower-extremities training in patients with lesion in the central nervous system [15,16].

5. Conclusion

The current study offered a preliminary insight into using the VR and visual biofeedback in lower-extremities training. Introducing a virtual mirror enabled active inclusion of subjects in the training process. The adaptation to the virtual instructor among a group of 10 healthy persons was evaluated by performing the virtual SIP training. The same investigation could be performed in treadmill walking; however, SIP has several advantages regarding the experimental conditions. It can be performed in a small area as long walkways are not required. This also applies to the measurement of movements. SIP training is safer than treadmill walking in VR, which has been associated with gait instability [17], and does not impose a fixed speed on the patient [18]. We concluded that healthy subjects were able to track the virtual instructor during SIP which suggests further applicability of the virtual mirror to other forms of lower-extremities virtual training. Introducing the virtual mirror in the rehabilitation environment could be potentially beneficial in terms of process quantification, standardization, and VR-related effects [1].

Conflicts of interest statement

The authors declare that no conflict of interest exists.

Acknowledgement

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