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A novel motion tracking system for evaluation of functional rehabilitation of the upper limbs*****

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Research Highlights

(1) In this study, we developed an inertial sensor-based motion tracking system, a tool for evaluation of the functional rehabilitation of upper limbs after central nervous system injury. The motion tracking system enabled us to analyze the complex upper limb and head movements in three dimensions according to nine degrees of freedom data from the kinematic models.
(2) The inertial sensor-based motion tracking system can be used to evaluate the functional recovery of the upper limbs after central nervous system injury accurately and stably.

Abstract

Upper limb function impairment is one of the most common sequelae of central nervous system injury, especially in stroke patients and when spinal cord injury produces tetraplegia. Conventional assessment methods cannot provide objective evaluation of patient performance and the effectiveness of therapies. The most common assessment tools are based on rating scales, which are inefficient when measuring small changes and can yield subjective bias. In this study, we designed an inertial sensor-based monitoring system composed of five sensors to measure and analyze the complex movements of the upper limbs, which are common in activities of daily living. We developed a kinematic model with nine degrees of freedom to analyze upper limb and head movements in three dimensions. This system was then validated using a commercial optoelectronic system. These findings suggest that an inertial sensor-based motion tracking system can be used in patients who have upper limb impairment through data integration with a virtual reality-based neurorehabilitation system.

Key Words

neural regeneration; brain injury; spinal cord injury; kinematic analysis; inertial measurement; motion tracking; upper limb; neurorehabilitation; virtual reality; sensors; grants-supported paper; neuroregeneration

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Conflicts of interest: None declared.

Ethical approval: The guidelines of the *Declaration of Helsinki* were followed and the study design was approved by the Local Ethics Committee of the National Hospital for Spinal Cord Injury, Toledo, Spain.

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INTRODUCTION

Upper limb function impairment is one of the most common sequelae of central nervous system injury[1-2]. Conventionally used assessment methods cannot provide objective evaluations of patient performance and the effectiveness of therapies[3]. The most common assessment tools are based on rating scales, which are inefficient when measuring small changes and can yield subjective bias[4-7]. An objective quantification of patient performance during rehabilitation can be achieved using instruments to capture motion trajectories and specific details of task execution. Various commercial systems use different sensor technologies to accurately track human motion[8-10]. Photogrammetry is based on the analysis of images captured from different positions to estimate the 3D coordinates of active or passive markers. Although this technique is very precise (with errors in the range of ± 1 mm[11]), markers can be occluded during the analysis of complex 3D movements of the upper limb, and its use is limited to a laboratory environment. Electromagnetic motion capture systems have been widely used to track human movements in virtual reality applications. While the problem of marker occlusion does not arise with these systems, the electromagnetic fields they use are subjected to interference and are affected by metallic objects. Inertial measurement units provide another alternative, and these sensors are designed to measure the orientation of an object within a given space. As they provide accurate readings without inherent latency (static accuracy of < 1.0° root mean square and dynamic accuracy of 3° root mean square[11]), these sensors are useful for human motion tracking applications. These devices are robust and several successful examples of inertial measurement unit measurement of upper limb movements have been described[10, 12-13]. However, most inertial measurement unit-based motion capture systems have focused on single-joint tasks and not on complex movements such as activities of daily living, which are required for upper limb rehabilitation.

Virtual reality technology is one of the most innovative and promising therapies for the rehabilitation of patients with motor deficits of the upper limb[14]. This approach can increase patient motivation, while extracting objective and accurate information enables the patient's progress to be monitored remotely. However, it is not yet possible to conduct a full objective kinematic assessment of the entire upper limb while performing the activities required using virtual reality systems and remote treatment monitoring. The ability to capture the actual movement of the patient and transfer it to a virtual environment is one of the strengths of virtual reality systems.

Because inertial measurement units are compact, light, resistant to environmental interference and easy to wear, they can be used as a motion capture system for virtual reality applications. The aim of this study was to develop and validate a motion capture system to analyze complex tasks performed using the upper limbs that are common in activities of daily life. Accordingly, we designed a suitable inertial measurement unit-based motion tracking system, and developed and validated a kinematic model with nine degrees of freedom that allows upper limb and head movements to be appropriately analyzed. These data can then be incorporated into a virtual reality-based rehabilitation device known as "Toyra".

RESULTS

The accuracy of the system

Joint movement was analyzed simultaneously using the Xsens system (the inertial sensor motion capture system using the proposed kinematic model) and the Codamotion system (a commercial human engineering metrological system)[12], inertial sensor-based motion tracking systems. The greatest difference in the range of motion values calculated by each system was found for wrist flexion-extension movement, which differed by 10.18° (129.41° ± 18.69 *vs*. 139.59° ± 6.52 for Xsens and Codamotion systems, respectively) (Table 1).

Table 1 Simultaneous analysis of joint movement using the Xsens and Codamotion systems (joint angle [°])

Joint trajectory	Maximum		Minimum		Range of motion	
	Xsens	Codamotion	Xsens	Codamotion	Xsens	Codamotion
Shoulder						
Flexion	157.76±13.03	159.25±11.69	–4.38±15.57	–4.89±12.20	162.14±19.59	164.14±13.86
Abduction	170.11±3.25	167.03±2.68	–6.23±0.47	2.59±0.37	176.34±2.89	169.62±2.64
Rotation	92.74±10.54	91.35±16.44	–49.39±5.67	–50.66±5.25	142.13±16.10	142.00±21.60
Elbow						
Flexion	153.92±3.40	150.67±1.04	11.81±8.25	10.20±8.61	142.10±4.86	140.47±9.61
Pronation-supination	41.46±2.32	38.87±0.41	–75.96±0.86	–77.75±4.37	117.42±1.46	116.62±3.97
Wrist						
Flexion	54.06±22.05	67.40±4.58	–75.35±3.35	–72.19±1.94	129.41±18.69	139.59±6.52
Radial-ulnar deviation	28.74±5.18	25.05±4.47	–26.07±4.01	–26.93±3.71	54.81±1.17	51.98±0.79
Head						
Flexion	32.70±5.18	35.16±5.31	–57.41±7.82	–57.10±8.59	90.12±12.94	92.27±13.83
Inclination	19.33±7.52	27.94±10.06	–41.59±6.79	–31.61±7.12	60.92±4.54	59.55±4.94

Results were expressed as mean ± SD.

To compare the captured data from both systems, the difference (distance) between the two sets of data was analyzed point by point in each sample. The final measure was the mean of all differences (distances) calculated by means of Student's *t*-test (Table 2).

Table 2 Comparison of the fluctuation of the same data (joint angle [°]) (experiment 1)

Joint trajectory	Joint angles (difference) (mean±SD)	*P*	*r*
Shoulder			
Flexion-extension	0.76±4.04	0.849	0.998
Abduction-adduction	0.69±10.47	0.851	0.991
External-internal rotation	–0.65±5.67	0.820	0.992
Elbow			
Flexion-extension	–0.54±2.63	0.880	0.999
Pronation-supination	–5.16±4.50	0.094	0.991
Wrist			
Flexion-extension	3.47±9.43	0.254	0.974
Radial-ulnar deviation	–2.19±4.64	0.068	0.954
Head			
Flexion-extension	1.58±1.34	0.424	0.999
Inclination	–8.24±2.10	0.000	0.993

Student's *t*-test was applied to analyze the difference (distance) between the numeric data obtained by means of both systems (Codamotion and Xsens technologies). *r* value is Pearson's correlation coefficient.

The *P* value was calculated to see if there were significant differences between these distances. With the exception of head inclination, there were no significant differences observed between the two sets of data (Xsens and Codamotion systems) obtained for any of the magnitudes analyzed.

To compare the increasing or decreasing trend of the captured data across both systems (Xsens and Codamotion), the Pearson's correlation coefficient (*r* value) was applied. Values between 0.95 and 1 were obtained for all the magnitudes measured, even those for which the mean was particularly high (Table 2). Thus, there was a great similarity between captured data in spite of the difference expressed in Table 1.

The system's robustness–drinking task test

Data obtained by the inertial sensor-based motion capture system using the proposed kinematic model for analyzing an activity of daily living (drinking from a cup) were compared with a previous study using Codamotion[12]. Data are shown graphically in Figure 1. Figure 1 shows the range of motion for each degree of freedom using means of the mean and standard deviation of the maximum and minimum values. For example, in shoulder flexion-extension, the maximum, a positive value, is a flexion value, and the minimum is negative, indicating an extension value for the shoulder joint. Figure 1 shows that both technologies possessed the required robustness for the measurement and analysis of human movements.

DISCUSSION

The objective of the present study was to develop and validate a motion capture system to analyze functional movements such as activities of daily living. In this study, we used inertial measurement units to design a suitable motion tracking system, and developed and validated a kinematic model with nine degrees of freedom that enabled complex upper limb and head movements to be analyzed. This system is currently being incorporated into a virtual reality-based rehabilitation device known as "Toyra".

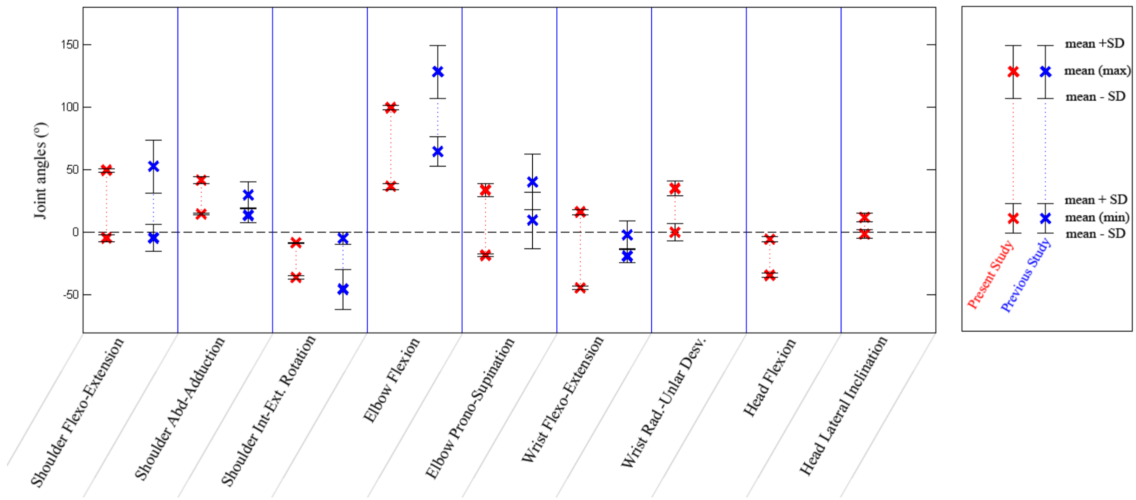

Figure 1 Joint angles in the activity of daily living (drinking task test) obtained by the inertial sensor-based motion capture system compared with a previous study[12].

In this figure, red shows the performance of the inertial sensors, whereas blue shows the performance of Codamotion. The statistical analysis is descriptive, so the figure shows the mean and standard deviation (SD) values for the maximum (max) and minimum (min) angles of each joint analyzed. Flexo: Flexion; Abd: abduction; Int: internal; Ext: external; Prono: pronation; Rad: radial.

The accuracy of our system was tested by measuring single-joint upper limb movements using a photogrammetric system, Codamotion, and the inertial measurement units, revealing similar results for both systems. The robustness of our system was also assessed by measuring a drinking activity using only the inertial measurement units, which produced comparable results to those reported previously using Codamotion[9]. Unlike previous studies[3, 8, 12], nine degrees of freedom (two in the head, three in the shoulder, two in the elbow and two in the wrist) were involved in the use of the inertial measurement unit-based system for the analysis of head movements in this study. We also developed a new kinematic model by modifying a method proposed previously[15]. The findings from the present study demonstrate the accuracy of the proposed system and the associated biomechanical model as well as their suitability for clinical use.

The measurement of complex movements performed by the upper limbs using the inertial measurement unit-based measurement system described here provided results similar to those previously obtained using other measurement systems based on kinematic models[16-21]. Simultaneous recordings of movements using the Xsens and Codamotion systems revealed that range of motion values for shoulder, elbow and wrist were comparable to those of previous studies[16-21]. Several considerations should be taken into account when interpreting these results. First, the results obtained at the shoulder vary greatly from one study to the next due to the complexity of this joint, and they are strongly influenced by the particular model applied. Thus, while the range of motion values for shoulder flexion and abduction closely matched previously reported values[9], they were higher than those reported in other studies[16-17]. Moreover, to account for the displacement of the scapula, we did not model the shoulder as a single joint, unlike previous studies [16-17]. This difference in the experimental approach may have further contributed to the divergent findings. The range of motion values obtained for shoulder rotation were similar to those previously reported using sensor-based measurement systems[17-18] but they were lower than those reported in goniometry studies[19-20]. Range of motion values for elbow pronation-supination in the present study were also lower than previously reported values (117.42° versus 160–180°)[17-21].

For head movements, the flexion-extension range of motion values was comparable to those reported previously[20, 22], although the lateral inclination range of motion values was lower in magnitude, possibly because subjects were requested not to reach the maximum point of their trajectory when performing these movements in order to avoid occlusion of the Codamotion markers. Significant differences in head inclination were found between the curves generated by the Codamotion and Xsens systems, which may have resulted from misalignment of the local coordinates for the inertial sensors and the Codamotion markers. Thus, despite obtaining a mean error of –8.24° and detecting significant differences in magnitude only, we obtained a correlation of 0.9932.

Strikingly, the range of motion values for shoulder rotation and pronation-supination were lower than the joint's anatomical range of motion[20], which may be because of a displacement of the sensors and markers in relation to the bone structure. Although our model assumes an invariable shape and size of each body segment, muscle and skin displacement in relation to the bone does occur[23].

The mean errors obtained using the Codamotion system were lower than those previously reported[12, 24], which may reflect methodological differences. In one study[24], gait was analyzed using sensors in which foot contact with the ground resulted in inertial acceleration peaks and a subsequent loss of accuracy[25]. By contrast, in the other study, photogrammetry markers were not placed in the same positions as inertial sensors, creating an additional source of error due to the relative displacement between sensors and markers[12]. Interestingly, the mean errors obtained for the shoulder joint were lower than those for the wrist, possibly because sensors in more distal positions are subjected to greater linear acceleration, making the Kalman filter less precise[25].

It should be noted that the activity of daily life drinking task was not recorded simultaneously with photogrammetric and inertial sensor-based systems. Due to the complexity of this movement and the localization of the sensors and markers at the same sites, some of the markers were hidden for the majority of the drinking task cycle, precluding simultaneous analysis. Thus, the results of the drinking task were compared with those of the control groups in two previous studies, one of which was conducted by our group[9, 26]. These results (range of motion and errors) allowed us to evaluate the accuracy of the system, a comparison that was designed to assess the robustness of the kinematic model used, with the modifications proposed, when analyzing a complex human movement.

While the results obtained for flexion and rotation of the shoulder joint were similar to previous findings[9, 26-27], some differences were observed in the maximum amplitude of abduction of this joint with respect to our previous study[9]. This discrepancy may be due to the drinking style of the subject who performed the task: some subjects kept their elbow close to the body while others moved it away from the body when drinking[26-27]. The maximum and minimum values for pronation-supination, and flexion and extension of the elbow, were lower than those obtained previously[13]. However, the greatest differences were observed for wrist flexion-extension, possibly because the participants in the previous study[9] began this task with the wrist in a neutral pronation-supination and flexion-extension position, while those in the present study could freely adopt the starting position of their choice.

The sources of error that might affect the particular system used should be considered when analyzing the results. In addition to the errors inherent to a system of this kind, including measurement errors or misalignment between the local coordinates of the sensors and the real coordinates of the joints, the relative displacement of the sensors in relation to the bone also affects the final results. This error mainly affects the measurement of the amplitude of shoulder rotation and the elbow pronation-supination. As sensors cannot be attached to bony prominences, this error can be minimized using a calibration process, as described previously[12]. This calibration involves assessing full shoulder rotation with markers placed on bony prominences, thereby minimizing the effect of displacement relative to the bone. The recorded signal is taken as the calibration signal. To correct for the effect of displacement of the inertial measurement units, the same movement (full shoulder rotation) is performed and the results are compared with the calibration signal to generate a correction function[12]. Another common source of error in inertial sensor-based systems is the drift introduced when calculating orientations using integration methods. In the present study, we used orientations provided by the sensors and those were calculated using a Kalman filter. As such, no drift was observed during the recording process.

Here we describe an inertial measurement unit-based motion capture system to analyze upper limb movement, for which we have developed a biomechanical model with nine degrees of freedom that provides kinematic data for the cervical spine and upper limb joints. The accuracy of this system was assessed by simultaneously analyzing single-joint upper limb movement using a validated photogrammetry system, which provided comparable results in the analysis of a drinking task. These findings demonstrate the suitability of our system for clinical applications. Moreover, in clinical settings, this system can be used in conjunction with new virtual reality devices (*e.g.*, Toyra) to achieve motor rehabilitation of the upper limbs.

SUBJECTS AND METHODS

Design
A descriptive study.

Time and setting
This study was performed at the Department of Biomechanics and Technical Aids, National Hospital for Spinal Cord Injury, Toledo, Spain in January 2012.

Subjects
A 30-year-old healthy right-handed male volunteer participated in the study after providing informed consent. The man underwent a physical examination to exclude any potentially serious pathology.

Methods
System description
We developed a motion tracking system using commercially available Xsens MTx inertial sensors (Xsens Dynamics Technologies, the Netherlands). These MTx inertial measurement units integrate a tri-axis accelerometer, tri-axis gyroscope, tri-axis magnetometer, and a temperature sensor to correct for temperature dependence. The position and angle of an inertial sensor cannot be correctly determined through integration methods, due to the noise and fluctuation of the offsets. Thus, the orientation of the MTx is computed by means of a Kalman Filter[28]. This filter uses the input from the rate gyroscopes, accelerometers and magnetometers to provide an accurate optimal estimate of the 3D orientation with very little drift[28]. In a homogeneous earth's magnetic field, the MTx system provides an angular resolution of 0.05° root mean square, static accuracy of < 1.0° root mean square and dynamic accuracy of 3° root mean square[11].

We used a set of five interconnected inertial measurement units that were connected wirelessly (Bluetooth) to a computer *via* a digital data bus (Master Xbus), which was responsible for the synchronization, data collection and transmission.

Kinematic model
While the inertial sensors provided information on the orientation of each body segment, a biomechanical model was required to calculate the angular magnitudes of clinical relevance on the basis of each orientation. The kinematic models commonly used to describe human motion are based on the Euler method, and thus the results depended on the sequence of rotations used[29]. By contrast, each magnitude was unequivocally represented in our model to aid the interpretation of the results.

The model proposed here considered only the head and the upper limbs. The upper limb was considered as a chain of three rigid bodies joined by the shoulder, elbow and wrist joints. This representation relies on several assumptions:

1. The head is considered to be a rigid solid object linked to the trunk by a hinged joint with two degrees of freedom, flexion-extension and lateral inclinations.
2. The shoulder joint is modeled as a spherical joint with three degrees of freedom. While the clavicle or scapula should also be included to provide a comprehensive representation of movement of the shoulder complex, these measurements were not performed for the following reasons. First, we sought to develop a simple system using as few sensors as possible, with only five inertial sensing units to monitor the hand, forearm, humerus and head. Secondly, in the case of the clavicle, inertial measurement unit attachment was quite difficult due to the small surface area available. Although inertial measurement units have been successfully attached to the scapula in other studies, demonstrating that scapulohumeral rhythm can be measured with minimal cross-talk[16], placing an inertial measurement unit over the scapula requires that the user's back be unclothed, which increases the set-up time and causes certain discomfort.
3. The forearm was considered to be a rigid body, and thus the pronation-supination movement was reallocated to the elbow as an additional degree of freedom in this joint[30]. The elbow was modeled as a hinged joint with two degrees of freedom, flexion-extension and pronation-supination.
4. The hand was considered to be open and was modeled as a single rigid body. The wrist was modeled as a Cardan joint with two degrees of freedom.
5. Each segment, including bones and soft tissues, had similar rigid body motions. The deformation of soft tissues did not significantly affect the mechanical properties of a segment as a whole[23].

The kinematic chain proposed in this model consists of nine degrees of freedom: two in the head (flexion-extension and lateral inclinations), three in the shoulder joint (flexion-extension, abduction-adduction and external-internal rotation), two in the elbow joint (flexion-extension and pronation-supination) and two in the wrist (palmar-dorsal flexion and radial-ulnar deviation).

As each degree of freedom was defined independently using planes and local coordinate systems in the human body, the angular magnitudes calculated did not depend on the user's position with respect to the global coordinate system.

A total of five MTx inertial measurement units were used to capture movements of the head and the right upper limb. The inertial measurement units were strategically placed on the trunk, the back of the head, the right arm, the forearm and the hand. The sensor in the trunk was mounted on a rigid mobile structure to align the Y axis of the sensor with the spinal cord. The forearm sensor was positioned distally to minimize displacement in relation to the bone (Figure 2).

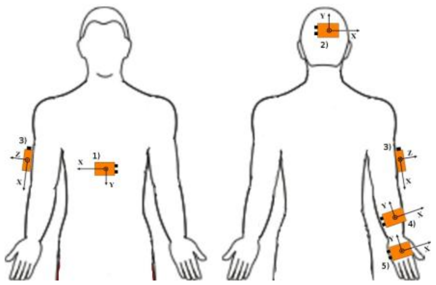

Figure 2 Placement of inertial sensors.

(A) Frontal view; (B) posterior view. The sensors were located on the trunk (1), the back of the head (2), the right arm (3), the forearm (4) and the hand (5).

Computation of joint angles

Each movement was defined independently using the planes and reference axes of the human body. To measure movements of one segment relative to the previous segment in the chain, it was necessary to define a local coordinate system for each segment. This reference system included three unitarian and orthogonal vectors.

As a global reference, we defined a reference system fixed to the trunk (*t1*, *t2* and *t3*: Figure 3), where: vector *t1* follows the straight line from one shoulder to another; vector *t2* follows the frontal axis in the anterior direction; and vector *t3* follows the vertical axis, completing an orthogonal base (*t3* = *t1* x *t2*). This reference system is centered on the base of the trunk (Figure 3).

The local coordinate system of the arm (*h1*, *h2* and *h3*) was established with the arm abducted at 90°, with the palm facing forward (Figure 4). This system is referenced to the center of the shoulder joint, where: vector *h1* follows the longitudinal axis of the arm, fixed to the humerus, from the shoulder to the elbow; vector *h2* follows the antero-posterior axis in the anterior direction; and vector *h3* represents the cross product of vectors *h1* and *h2* (*h3* = *h1* x *h2*).

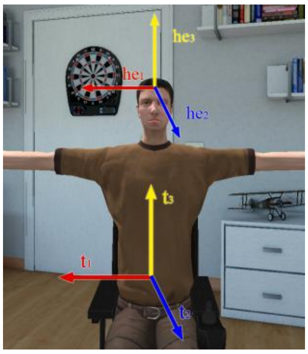

Figure 3 The avatar represents the movement performed by the 30-year-old right-handed male.

The figure shows the local reference system (*he1*, *he2*, *he3*) of the head and the global reference system of the avatar (*t1*, *t2*, *t3*). The head coordinate system (*he1*, *he2*, *he3*) was defined parallel to the trunk reference system and centered on the top of the head. The blue color represents the anterior-posterior axis, the red color the medial-lateral axis, and the yellow color the longitudinal axis.

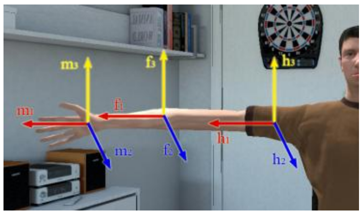

Figure 4 The avatar represents the upper limb's movement performed by the 30-year-old right-handed male.

The figure shows the local reference systems of the arm (*h1*, *h2*, *h3*), forearm (*f1*, *f2*, *f3*) and hand (*m1*, *m2*, *m3*). The blue color represents the anterior-posterior axis, the yellow color the medial-lateral axis, and the red color the longitudinal axis.

In the forearm (Figure 4): vector *f1* follows the longitudinal direction of the forearm from the elbow to the wrist; vector *f3* is perpendicular to *f1* and parallel to the wrist from the ulnar to the radial styloid; and vector *f2* completes the reference system (*f2* = *f3* x *f1*). The neutral position of the forearm is defined in relation to the humerus when the arm is completely extended (*h1* is parallel to *f1*) with the palm of the hand in a medial position.

The local reference system of the hand describes its movements with respect to the forearm and in the proposed model, and the hand is represented as a single rigid body. The hand is considered open, facilitating the definition of the vectors: vector *m1* runs over the palm of the hand, from the center of the wrist joint to the fingers; vector *m2* runs perpendicular to the palm of the hand;

and vector *m3* represents the cross product of vectors *m1* and *m2* (Figure 3).

The model used to calculate the angular magnitudes was based on a model previously proposed[15]. The definition of the angles relative to the shoulder was modified slightly, as the authors of the previous study did not establish an unequivocal relationship between the position of the arm and the value of each magnitude.

(1) Shoulder flexion: this is represented by the angle formed between the upper arm and the coronal plane. When *h1* is below the transverse plane, it can be calculated as π/2 radians minus the angle formed between the *h1* and *t2* vectors, and as π/2 plus the angle formed between *h1* and *t2* when *h1* is over the transverse plane.
(2) Shoulder abduction: this is represented by the angle formed between the upper arm and the sagittal plane. When *h1* is below the transverse plane it can be calculated as π/2 radians minus the angle formed between *h1* and *t1* vectors, and as π/2 plus the angle formed between h1 and t1 when h1 is over the transverse plane.
(3) Shoulder rotation: defined as the angular movement of the humerus over its own longitudinal axis (*i.e*., over vector *h1*).
(4) Elbow flexion: this is defined as the angle between the *f1* and *h1* vectors, according to the neutral position defined when the arm is completely extended.
(5) Forearm pronation: this is defined as the angular movement of the forearm over its own longitudinal axis (*i.e*., over vector *f1*).
(6) Radial-ulnar deviation: this is the angle formed by vector m1 and the plane that includes vectors *f1* and *f2*. This angle can be calculated as π/2 radians minus the angle between *m1* and *f3*.
(7) Palmar flexion of the wrist: this is the angle formed by vector m1 and the plane that includes vectors *f1* and *f3*. This angle can be calculated as π/2 minus the angle between *m1* and *f2*.
(8) Head flexion: this is defined as the angle between vector he3 and the plane that includes vectors *t1* and *t3*. This angle can be calculated as π/2 minus the angle between *he3* and *t2*.
(9) Head inclination: this is the angle formed by the vector *he3* and the plane that includes vectors *t2* and *t3*. This angle can be calculated as π/2 radians minus the angle between *he3* and *t1*.

Validation procedure

The system was validated *in vivo* in two experiments carried out on different days. The first assessed the accuracy of the proposed inertial measurement unit system in measuring upper limb kinematics in a clinical environment, while the second assessed its robustness.

Testing the accuracy of the system

The accuracy of the proposed inertial sensor-based measurement system and the biomechanical model described above was validated using a clinically recognized procedure with kinematic analysis equipment (Codamotion: Charnwood Dynamics Ltd, UK), a photogrammetry system based on active markers. This system has active markers that emit infrared light that could be recorded by scanning units (cx1).

Set-up and procedure: Single-joint upper limb movements were recorded simultaneously in the selected subject using two motion capture systems, Codamotion and inertial measurement units. A set of 15 active markers was used to capture movement with the photogrammetry system on the basis of a previously described model[5]. These markers were distributed on five rigid structures to minimize the potential error resulting from marker displacement over the skin surface, and each was placed on the body segments to be analyzed: trunk, head, arm, forearm and hand. Each structure contained three active markers and one inertial measurement unit, and accordingly, simultaneous measurements were obtained with both motion tracking systems in the same environmental conditions (Figure 5).

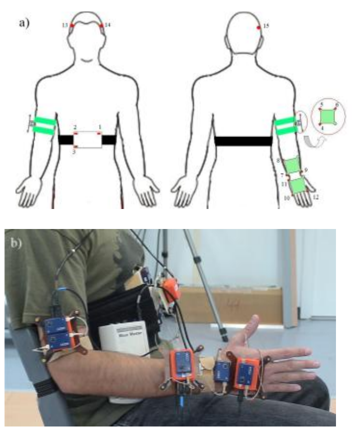

Figure 5 Placement of Codamotion markers and inertial measurement units (IMUs) simultaneously.

(a) Frontal view and posterior view of Codamotion markers (numbers between 1 and 15 represented in the figure).

(b) Codamotion markers and IMUs simultaneous set up.

Two Codamotion scanner units (cx1) were used, one placed in front of the subject, slightly to one side with respect to the midline and contralateral to the instrumented arm of the subject, and the second positioned laterally[9] (Figure 6). The subject was instructed to perform each of the following nine single-joint-angle tasks three times: head flexion-extension and lateral inclinations; shoulder rotations, flexion-extension and abduction-adduction; elbow flexion-extension and pronation-supination; wrist flexion-extension and ulnar-radial deviations. In each repetition, the subject cyclically executed the movement three times.

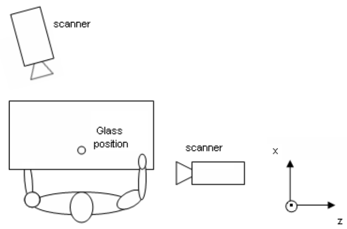

Figure 6 View from above of the set-up for the activity of drinking from a glass.

The *XYZ* coordinate system and cameras position are visible. The subject has the arm at the starting point.

Data analysis: Data were obtained from nine movement cycles for each task and the data from both measurement systems were collected simultaneously. A sampling frequency of 200 Hz was used for Codamotion photogrammetry recordings and of 25 Hz for the MTX inertial sensor systems. The first processing step involved applying a decimation process to the photogrammetry recordings for frequency equalization. Thus, the sampling frequency was set to 25 Hz, the same frequency as that used in the virtual reality-based rehabilitation platform with which we sought to integrate our system.

The orientation matrices of each segment were derived from the position of the photogrammetry markers. These matrices for the inertial measurement units were provided directly, and thus did not need to be calculated. The angular magnitudes of interest were calculated for the recordings obtained with both systems, as indicated in the description of the kinematic model, and the results were converted from radians to degrees. To assess the differences between the systems for each task, the mean and standard deviation (SD) were computed for each variable. Based on previous studies[17-18], the following kinematic variables were included in the present study: maximum value, minimum value, and the difference between maximum and minimum values (*i.e.*, the range of motion). These variables were calculated for the following joint trajectories: flexion-extension, abduction-adduction and external-internal rotation of the shoulder joint; flexion-extension and pronation-supination of the elbow joint; palmar-dorsal flexion and radial-ulnar deviation of the wrist; and flexion-extension and lateral inclination of the head. The data were processed using MATLAB software version R2007b (Mathworks, United States).

Testing the robustness of the system with a drinking task

After demonstrating a high degree of accuracy by comparing our system with the established Codamotion system, we analyzed its performance in measuring complex activities, such as those associated with activities of daily living. Thus, we analyzed its performance in a drinking task performed by the same subject registered in Experiment 1 using only the validated inertial measurement units, comparing the results with those of a previous study in which the same task was analyzed in similar subjects using the same experimental set-up, but with the Codamotion system using the same marker positions as described previously[9].

Set-up and procedure: The five inertial measurement units were attached to the five rigid structures used in the previous experiment, without the active markers of the photogrammetry system. The experimental set-up (subject starting position, seating configuration, subject-to-table distance, glass position) was identical to that used in the previous study (Figure 6). The subject was instructed how to perform the drinking task, which involved reaching out for the glass from the starting position, grasping it, raising the glass to the mouth, drinking, lowering the glass to the pickup point, and returning the hand to the starting position. This activity was practiced twice to establish a comfortable sitting position before the exercise was recorded[9]. Movements were recorded as the subject executed the drinking task at a comfortable, self-selected speed. Three recordings were obtained for analysis and processing.

Data analysis: To assess the differences between the results of this and a previous experiment[9], the mean and SD were computed for each variable. The variables analyzed were the same as those described for the accuracy experiment.

Statistical analysis

The descriptive statistical analysis was performed using the mean and SD. To compare the results obtained using the Codamotion and Xsens MTx inertial sensors, we calculated the mean and the SD (the distance between two samples of data) obtained with both systems for each degree of freedom analyzed, comparing point by point, using Student's *t*-test. The Pearson's correlation coefficient was applied to analyze the trend (fluctuation) between the numeric data from both systems (Codamotion and MTX inertial sensor systems). All statistical analyses were performed using SPSS 12.0 software (SPSS, Chicago, IL, USA) and P < 0.05 was considered statistically significant.

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