Wearable Absolute 6 DOF Exercise Training System for Post Stroke Rehabilitation

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ABSTRACT

This paper introduces a multidisciplinary research effort to design an interactive, real time system to assist rehabilitation of stroke patients. The proposed system aims to help stroke survivors with limited mobility to relearn lost skills and regain maximal functional independence. The system assesses the quality of exercise being performed and provides advisory feedback to the patient in order to enable task specific training and correct motor relearning. The system uses a novel, ambulatory and wearable 6 Degree of Freedom (DOF) motion capture system composed of miniature ultrasonic and inertial sensors. The system captures the 3D kinematic parameters of the subject's movement while performing exercise, enabling telemedicine applications or subsequent review by clinical experts. The paper presents the architecture of the aforementioned system and results from early stages of the research.

1. INTRODUCTION

Stroke, or Cerebrovascular Accident (CVA), is caused by a disruption to the blood supply to certain parts of the brain due to the blockage or rupture of blood vessels. Stroke is the third largest cause of death and the main cause of adult acquired disability worldwide [37]. In Ireland, stroke accounts for 7.4% of mortality. It is estimated that there are over 30,000 stroke survivors in the country, many living with residual disability [11]. Common sequelae post stroke include muscle paralysis, communication problems, fatigue, urinary incontinence and mental and emotional problems. Of stroke survivors, 48% suffer hemiparesis, 22% are unable to walk and 53% require help with activities of daily living (ADLs) [10]. The aim of rehabilitation is to maximize recovery and enable the stroke survivor to achieve maximum functional independence. The rehabilitation program for a stroke survivor may include physical therapy, speech therapy and occupational therapy. Research suggests that while most recovery takes place within the first 30 days, recovery may continue for 6 months to several years, owing to the neuroplastic behavior of brain [29]. A well organized, extended, task specific and repetitive rehabilitation program can enable restoration of motor function and improve the quality of life of the stroke survivor.

A recent national audit of stroke services revealed that there is an insufficient number of dedicated, specialized stroke units and that existing services are overburdened [10]. Pressure on limited resources often reduces the average length of inpatient care, creating a greater need for domicillary intensive rehabilitation. Home-Based rehabilitation systems allow patients to practice rehabilitation exercises in a comfortable environment reducing the burden on specialized stroke care units. There is evidence that supported discharge can reduce long-term dependency and admission to institutional care [24]. A dose dependent effect of therapy has also been demonstrated with increased augmented exercise therapy time after stroke having a favourable effect on ADLs and gait speed in cases of mild to moderate post-Stroke impairment [22]. Depending as it does on the area and extent of brain damage post-Stroke, it is rare for two stroke survivors to have the same clinical presentation. Given the heterogeneity of the stroke population, rehabilitation systems need to be adaptive so that they can be individually tailored to the patients' needs. They must possess an easy-to-use human-computer interface for those with acquired visual or language impairments. In addition, a structured database to enable periodic monitoring by experts tracking the recovery of the patients is needed.

The proposed system is an adaptive system controlled by a remote or local healthcare professional. It employs a novel, real-time, ambulatory 6 Degree of Freedom (DOF) motion capture system composed of inertial measurement units and acoustic sensors. The use of acoustic receivers allows for 6 DOF absolute positioning at low cost in a wearable system. The inertial sensors provide dead reckoning position estimates when the acoustic sensors do not have line of sight to the transmitters. The system derives kinematic parameters from the motion capture system and performs an assessment of the current exercise using a scale-invariant state space mathematical model. The system provides real time corrective biofeedback signals to the user in order to enhance his/her performance. For key functional tasks such as sit-to-stand, the system can be either used in a domicillary setting with remote supervision or as a adjunct to routine therapy in a tertiary unit, reducing the demand for direct supervision by a physiotherapist. In a home exercise programme, the system can be used for motor relearning tasks as prescribed by the treating therapist, with regular expert reviews and appropriate updates to the therapy plan.

The paper is organized as follows. Section 2 presents previous work reported in the area of rehabilitation systems for stroke patients. Section 3 outlines the architecture of the system proposed herein. Section 4 provides insight into the motion tracking unit used in the system, Section 5 presents initial experimental results. Section 6 presents conclusions and future work.

2. RELATED WORK

A survey of existing systems for assisting post stroke rehabilitation was conducted. This section reports key contributions in the field. Section 2.1 provides insight into existing rehabilitation systems. Section 2.2 reports background work on motion tracking systems, which forms a major functional subunit of rehabilitation systems. Section 2.3 describes previous work on automated assessment and biofeedback systems.

2.1 Rehabilitation systems

Recent advances in telemedicine, virtual reality and motion capture systems have contributed to several commercial systems being developed with the aim of assisting home based rehabilitation. Philips reported prototyping and testing a rehabilitation exerciser to increase the efficiency and effectiveness of rehabilitation at home [23]. The reported system consisted of small inertial measurement units containing accelerometers, gyroscopes and magnetometers. The system determines the orientation of the upper limbs by fusing sensor data from the inertial units and magnetometers. The SMART rehabilitation system [27], is a web based telemonitoring system consisting of three components: motion tracking units, base station and web-server. The tracking unit is composed of MT9 sensors attached to a patient's limbs. It captures arm movements during the reach task. Data is sent to the base station where the patient can compare his/her actions against a 3D graphical template. The web server unit enables the therapist to remotely examine the performance of the patient. In both of these systems, the capability of sensor units limits their use in applications requiring accurate position estimates for joints, e.g., tasks (such as eating) involving positioning the limbs at specified points in 3D space. Furthermore, magnetometers are susceptible to indoor electromagnetic fields caused by ferrous materials or magnets (eg. speakers).

Robot assisted rehabilitation devices have been developed for automating training of arm, wrist and finger movements and for gait and posture correction following a stroke. MIT-MANUS [15] can move, guide or perturb the movement of a subject's upper limb. Clinical trials [19], [20] using MANUS illustrate greater functional recovery compared to conventional therapy practice. The Mirror Image Movement Enabler (MIME), is a robotic device for shoulder and elbow rehabilitation. MIME [26] consists of a PUMA 560, which applies forces to the paretic limb during unilateral and bilateral movements in three dimensions. Arm Guide [31] is a diagnostic and therapeutic tool. As a diagnostic tool, it evaluates key motor impairments such as weakness, non-coordination and as a therapeutic tool it provides a means to assist therapy for the arm. The task supported is a reaching task. Although robot assisted rehabilitators improve motor recovery, questions have been raised regarding the suitability and effectiveness of robot assisted rehabilitation systems [35]. Moreover, such systems lack flexibility, and are complex, potentially unsafe and expensive.

Virtual reality aided rehabilitation systems provide an interactive, contextual, meaningful environment for motor training tasks. A rehabilitation system for upper extremity function restoration using virtual reality [18] has been reported. Using a virtual environment-based motor-training system [16],[17], the therapist at a remote location conducts treatment sessions with the patient located at home. The patient uses a computer, a motion capture system and video cameras to emulate exercises played on a screen. The therapist continually monitors the actions performed by the patient and provides augmented feedback. Mechatronic devices and virtual reality were incorporated into a rehabilitative hand training system [9]. The Rutgers arm [21] consists of a low-friction table, 3D tracker, custom forearm support, PC workstation, library of Java 3D virtual reality (VR) exercises, clinical database module, and a tele-rehabilitation extension. An increase in finger dexterity, grip force, and endurance was reported using the system reported in [6]. A machine mediated neuro-rehabilitation system, GENTLE/s [25], involving integration of haptic technologies and virtual environments provided therapies to people with upper limb impairment due to a stroke. The performance of virtual reality based systems is affected by static registration errors, optical distortion and mechanical misalignment [4]. End to end system delay is also an issue that limits their use on slower processing platforms.

A review of the literature shows that a number of rehabilitation systems exist for upper extremity rehabilitation. But lower extremity rehabilitation is relatively unexplored. Two systems [33], [34] robot assisted systems for gait training of stroke survivors have been reported. Preliminary results reported in [3] for automated assessment of a sit-to-stand exercise using a stereo vision based system, include scores consistent with the Berg Balance Scale. Standing up is critical to locomotion and thus functional independence of the stroke survivor. The proposed system is intended for rehabilitation of the sit-to-stand exercise.

2.2 Motion Capture Systems

Existing commercial motion capture systems can be categorized into two broad categories, vision-based systems and non-vision-based systems. Vision based motion capture systems can be either marker based or marker-free systems. The CODA [28] system consists of precalibrated cameras which can measure the location of active markers. VICON [2] uses multiple video cameras with passive markers for motion capture. Qualysis, PeakMotus, ReActor2, ELITE bitomech are other vision-based motion capture systems. Although these systems give remarkable accuracy in a lab setup, but they are not a practical choice for home based rehabilitation system [38]. Vision based systems need a long setup time, a high performance processing platform, a controlled environment (lighting conditions) and suffer from occlusion problems. The requirement for optical cameras renders the system to be non ambulatory and non wearable.

An alternative to vision-based systems are inertial systems, consisting of accelerometers and gyroscopes. Due to their low latencies, high sampling rates and small form factor they are a good choice for ambulatory motion capture systems. These systems don't have line-of-sight restrictions. MT9 (Xsens Motion Technology, Netherlands), G-link [1], GypsyGyro-18 from Animazoo are inertial tracking systems suitable for placement on-body. Magnetic MoCap systems (such as MotionStar from Ascension) use a permanent transmitter (a set of three coils) that induces magnetic fields in the environment. Small receivers attached to the user's body measure these magnetic fields to estimate the user's body position. Inertial systems have inherent drift problems. The aforementioned systems use magnetometers to correct the heading and to minimize drift error. Nev-

ertheless, they are not convenient for home environments because electronic devices and ferrous metals can change the electromagnetic fields induced by the transmitter and thus distort receiver measurements. MotionStar, Intersense, Polhemus are alternative devices [38]. Intersense is a hybrid system which integrates data from inertial measurement units and the Global Positioning System. Researchers have used GPS systems with inertial measurement units to correct drift problems, but GPS is not suitable for indoor use. An ultra-WideBand (UWB) radio positioning system (UBISENSE) has been used in addition to Gypsy-gyro inertial motion capture system, to get accurate position information and track human movement data [7]. However, the UWB data is used to determine the global translation in the position of the human operator, not the individual anatomical landmarks of the human body. IGS-190-Hybrid from Animazoo uses hybrid gryroscopic data and ultrasonic data for 3D motion capture, with receivers at fixed positions in room. The system is accurate but non ambulatory. We propose a wearable, ambulatory motion capture system that fuses accurate position estimates from ultrasonic system with inertial measurement unit data to correct drift errors. Thus the system gains from the absolute accuracy of ultrasonic ranging and tolerance of occlusion provided by inertial sensors.

2.3 Assessment and Biofeedback Systems

Several biofeedback systems for automated assessment of stroke patients have been reported in the literature. Statistically significant improvement in muscle strength after stroke using functional electrosimulation (FES) was reported in [12]. An angular biofeedback device [8] for stroke patients using a mercury switch helped stroke survivors to maintain an erect posture while sitting and to regain balance. A combination of auditory and vibratory stimuli [30] was used on stroke patients, who had the task of maintaining an upright position. This system reduced sagittal torque variance (body sway). An apparatus for learning body motion [5], provides feedback to the user using a musical tune whose note goes off-key depending on the level of discrepancy. The Philips rehab exerciser [23], uses animation feedback to the patient showing the target exercise and the current exercise on a split screen, the patient perceives the discrepancy in movement and adjusts in the next trial accordingly. The system proposed herein employs a real-time audio-visual feedback system, which is distinguished by its capability to identify spatial and temporal frame where the user goes off the desired motion trajectory.

3. SYSTEM OVERVIEW

Herein, we propose a rehabilitation system with the functional units shown in Figure 1. The treating clinician manages the system. Based on assessment findings, the clinician, in partnership with the patient, decides the appropriate functional goals and therapy plan. The therapy programme includes the exercises to be performed, their repetition frequency, difficulty level, etc. Exercises conducted under a clinician's supervision serves as a template for the automated assessment of the patient's performance. The clinician has access to a historical log for the patient to better understand the recovery path to date and updates the therapy plan at regular intervals.

When exercising, the patient wears a number of ambulatory motion tracking units. The patient tries to emulate the exercises shown to him/her in a guidance video. The aim is to score points based on the degree of similarity between the template and the performed action. The patient receives advisory feedback from the system which enables and motivates him to correct the mistakes in the previous trial. Repetitive training sessions with gradually increasing difficulty levels enable the patient to effectively relearn lost motor skills.

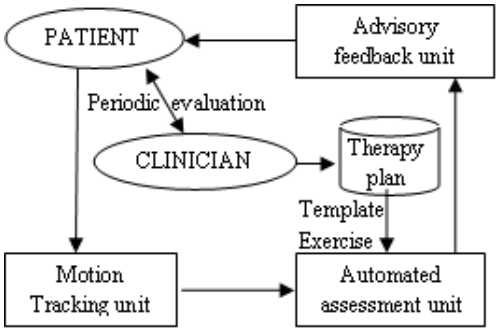

Figure 1: Proposed System Architecture

The motion tracking unit is a wearable, ambulatory, real time system capable of capturing 6 degrees of freedom of human motion, i.e., translation and rotation along 3 perpendicular axes in space. The unit is composed of inertial measurement and acoustic sensors. Details of the motion tracking unit are contained in Section 4.

Automated assessment is an important part of the system. A state space representation of the exercise performed under the clinician's guidance is stored as a part of the therapy plan. This serves as the reference template for use in automated assessment. The aforementioned state space representation is based on spatio temporal features extracted from exercise. The assessment unit takes as input kinematic parameters (joint angles, limb positions) from on-body motion tracking units. It extracts features from the input and compares them with the template movement. Based on the resemblance of the two state space models, a score is generated, which is indicative of the level of similarity between the patient's current attempt and the target template exercise which the patient aims to achieve. For sit-to-stand exercise, it has been shown that the temporal characteristics of hip and knee angular displacements are indicative of improvements in stand-up task [36]. So, kinematic parameters for assessment of sit-to-stand task are hip and knee angular velocity and their velocity curves to infer co-ordination between hip and knee movements. The nature of the state space model and the features to be used in such modeling are under investigation. The system should use a scale invariant, subject independent representation of the template exercise, so that it can be used to train a large number of patients by varying thresholds (such as joint angular velocity) depending on the level and nature of their disability.

The advisory feedback unit provides real-time feedback to the patient based on the resemblance of the current attempt to the template exercise. The advisory feedback unit identifies the specific temporal and spatial frame at which the patient deviates from the original template and provides feedback to the patient in the form of audio-visual

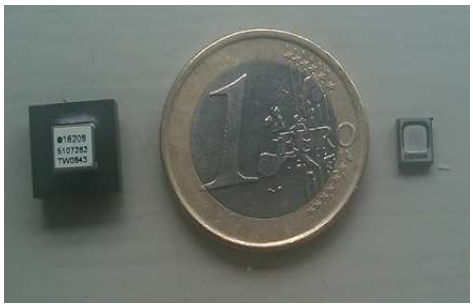

Figure 2: Form factor of IMU (left) and ultrasonic receiver (right) used in the sensor units

signals. The audio signal is in the form of a beep which increases in tone as the patient's limb moves out of the threshold range for a specific exercise and the video is in the form of an animation where the specific part of the body responsible for deviation from the template movement is highlighted. For example, in the case of the sit-to-stand exercise, if the user has poor timing and coordination between hip and knee movements, the audio signal goes off-key as soon as the patient's knee extension is at a higher rate than their hip extension, corresponding areas are highlighted on animation. The patient is required to rectify his/her movement inorder to clear the highlighted areas in animation and to receive a normal audio signal.

4. MOTION TRACKING UNIT

4.1 Overview

The proposed motion tracking system is an ambulatory, six degree of freedom, full body motion capture system. Each sensor unit of the proposed system is composed of an Inertial Measurement Unit (IMU) and an ultrasonic receiver. Figure 2 illustrates the form factor of the IMUs and acoustic receivers used in the sensor units. The ultrasonic subsystem is highly accurate, robust to multi path noise and reverberation [13]. The Sensor placement is illustrated in Figure 3.

The 15 segment Hanavan body model [14] is adopted for placement of the sensors. The base station is composed of 3 ultrasonic transmitters placed around the waist using a belt. As the waist undergoes movement only during whole body translation, it is used as a fixed frame of reference. The belt is also equipped with inertial units whose output is double integrated to determine the whole body translation The following steps are involved in motion capture:

Step 1:- Initial measurement of limb lengths and sensor placement.

Step 2:- Sensor board and limb axis calibration. The calibration procedure is explained in the next section.

Step 3:- The ultrasonic transmitters at the waist transmit Frequency Hop Spread Spectrum (FHSS) signals simultaneously. Frequency Hopping helps reduce the impact of multipath, reverberation and noise problems and allows for separation of the signals at the receiver.

Step 4:- The range data from each sensor unit to the transmitters at the belt is estimated using ultrasonic Time of Flight measurement. The algorithm for accurate range estimation is based on the method employed in [32].

Step 5:- Inertial data (acceleration and 3D rate of turn) is obtained from each sensor board.

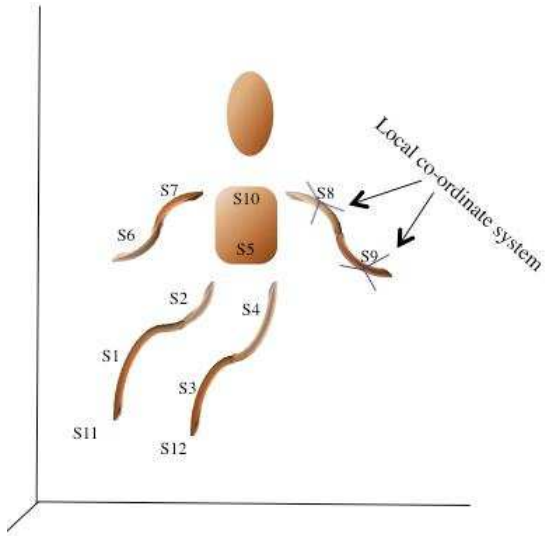

Figure 3: Hanavan body model for sensor placement and motion capture

Step 6:- In the case of clear line of sight from all three transmitters, ultrasonic ranging data is converted into 3D co-ordinate estimates for each sensor unit. Thus the 3D position of each sensor board with respect to fixed frame of reference is known.

Step 7:- Using the known position of the sensor units on the limbs (from step 2), the orientation and position of the limbs is calculated by inverse kinematics. A detailed mathematical description is provided in the next section.

Step 8:- In the case of non line of sight with the ultrasonic receivers, the sensor unit positions on limbs are calculated by dead reckoning. Dead reckoning is effective for brief occlusions, but the errors become large for longer periods of time.

4.2 Limb Tracking

Current work is focused on six degree of freedom arm tracking. Although the system is intended for lower limb tracking, for simplicity and fast prototyping, arm tracking was chosen for the early stages of research. The algorithm proposed here applies equally for lower limbs. Sensor units consisting of acoustic receivers and an IMU are placed on selected anatomical landmarks on the arm. As shown in Figure 4a, the points S, E and W represent shoulder, elbow and wrist respectively. Figure 4b is a zoomed view of the forearm and shows the local axes. Two sensor units, S_1 and S_2 are placed on the forearm and a sensor unit S_3 is placed on the upper arm. The sensors are placed at carefully selected locations on the limbs so that due to human body constraints, there is no rotation of the upper arm around S_3E and no rotation of the forearm around S_1S_2. Limb lengths EW and SE were measured prior to the experiment. Each limb has its own local frame of reference (O_L). The global frame of reference (O_G) is aligned with the sensor unit placed on the waist. Each sensor unit has its own local frame of reference.

Step 1:- Initial Calibration (Sensor/limb axis alignment)

The aim of this step is to find the orientation of vector $\vec{S_1S_2}$with respect to the limb's local frame of reference O_L, i.e., angles α_L, β_L, γ_L. The forearm's local X axis (X_L)

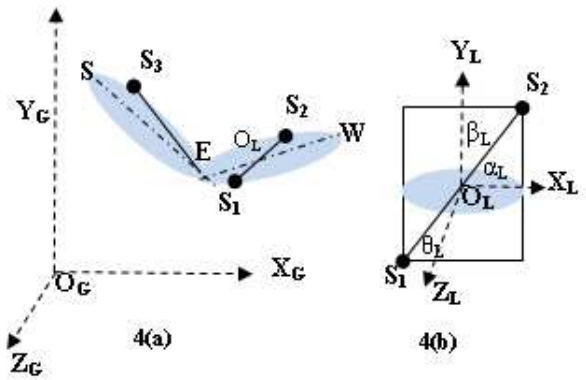

Figure 4: (a) Sensor placement for limb tracking. (b) Orientation of local axes with respect to limb

is parallel to the limb length, the Z axis (Z_L) is normal to the limb and the Y axis (Y_L) is perpendicular to the XZ plane. The subject is asked to keep their elbow fixed on the table and keep their shoulder still. The subject performs a flexion-extension of the arm. This movement corresponds to rotation around the local Y axis of forearm. Data from the sensor units is recorded and the corresponding angles are calculated. As a part of the initial calibration, the distances S_1O_L and S_2O_L are calculated. Let the coordinates of S_1O_L and S_2O_L, with respect to the limb attached frame of reference, be $(x_{L_1}, y_{L_1}, z_{L_1})$ and $(x_{L_2}, y_{L_2}, z_{L_2})$, respectively. These orientation and distances measures are assumed to be fixed until the end of exercise.

Step 2:- Translation of Limb-fixed frame of reference to Global frame of reference.

From acoustic ranging we know $S_1(x_1, y_1, z_1)$, $S_2(x_2, y_2, z_2)$ and $S_3(x_3, y_3, z_3)$ where (x_i, y_i, z_i) are the X, Y and Z co-ordinates of S_1 and S_2 with respect to the fixed frame of reference. In the case of non line of sight, the 3 axis accelerometer data is double integrated to find its current co-ordinates based on last known position. The coordinates of O_L with respect to global frame of reference are given by

$$x_{O_L} = \frac{(\|S_2O_L\|) \times x_1 + (\|S_1O_L\|) \times x_2}{\|\vec{S_1S_2}\|} \tag{1}$$

Similarly, y_{O_L} and z_{O_L} are obtained using Equation (1). Now, the limb's fixed frame of reference with origin, O_L is translated to point O_G. The new coordinates of point S_1, i.e., $S_1^{'}$ $(x_{L_1}^{'}, y_{L_1}^{'}, z_{Z_1}^{'})$ are translated to the new frame of reference using Equation (2).

$$\begin{pmatrix} 1 & 0 & 0 & -x_{O_L} \\ 0 & 1 & 0 & -y_{O_L} \\ 0 & 0 & 1 & -z_{O_L} \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x_1 \\ y_1 \\ z_1 \\ 1 \end{pmatrix} = \begin{pmatrix} x_{L_1}^{'} \\ y_{L_1}^{'} \\ z_{L_1}^{'} \\ 1 \end{pmatrix} \tag{2}$$

Similarly, S_2 is transformed to the translated frame of reference.

Step 3:- Determination of orientation and position of O_L with respect to O_G.

Point $S_1^{'}(x_{L_1}^{'}, y_{L_1}^{'}, z_{L_1}^{'})$ is related to point $S_1(x_{L_1}, y_{L_1}, z_{L_1})$ by the following equation

$$\begin{pmatrix} a_{11} & a_{12} & a_{13} \\ a_{21} & a_{22} & a_{23} \\ a_{31} & a_{32} & a_{33} \end{pmatrix} \begin{pmatrix} x_{L_1}^{'} \\ y_{L_1}^{'} \\ z_{L_1}^{'} \end{pmatrix} = \begin{pmatrix} x_{L_1} \\ y_{L_1} \\ z_{L_1} \end{pmatrix} \tag{3}$$

where, (ψ, θ, ϕ) are the Euler angles of rotation around X, Y and Z axes respectively and,

$a_{11} = cos(\theta)cos(\phi)$
$a_{12} = -cos(\theta)sin(\phi)$
$a_{13} = sin(\theta)$
$a_{21} = sin(\psi)sin(\theta)cos(\phi) + cos(\Psi)sin(\phi)$
$a_{22} = -sin(\psi)sin(\theta)sin(\phi) + cos(\psi)cos(\phi)$
$a_{23} = -sin(\Psi)cos(\theta)$
$a_{31} = -cos(\Psi)sin(\theta)cos(\phi) + sin(\Psi)sin(\phi)$
$a_{32} = cos(\Psi)sin(\theta)sin(\phi) + sin(\Psi)cos(\phi)$
$a_{33} = cos(\Psi)cos(\theta)$

Using Equation (2) and (3), roll, pitch and yaw, i.e.,(ψ, θ, ϕ) are determined. The position of the center of the limb is given by coordinates of O_L derived in step 2.

5. RESULTS

Elbow flexion-extension and forearm pronation-supination were simulated in Matlab. Figure 5 and Figure 6 present the cumulative error of the co-ordinate estimates and orientation estimates in millimeters and degrees respectively, for 1000 different positions and orientations. To access the impact of sensor placement error in step 1, a sensitivity analysis was performed allowing a 10 degree error in pitch, yaw and roll. The error in ultrasonic ranging data is assumed to be less than 2 mm, based on results presented in [32]. It was observed that, with no error in sensor placement, the root mean squared error in 3D coordinate estimates is less than 5 mm in 95% of cases and orientation error is less than 1.5 degrees in 88% cases. With 10 degrees of error in sensor placement, the percentage of cases having error less than 5mm drops by 2%.

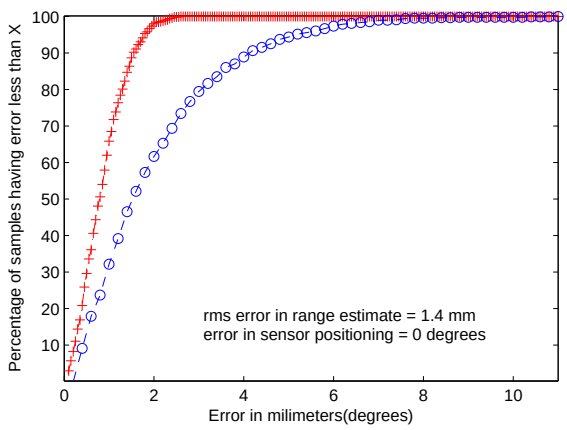

Figure 5: Cumulative error for 1000 trials with no error in sensor placement

6. CONCLUSIONS AND FUTURE WORK

An interactive, real-time rehabilitation system suitable for home-based use with remote supervision or as an adjunct to in-patient therapy has been proposed. The system is flexible

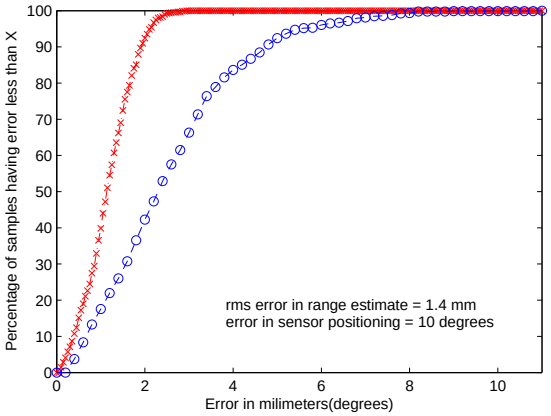

Figure 6: Cumulative error for 1000 trials with 10 degree error in sensor placement

and can adapt to patients with varying levels of disability. The recovery plan is guided by clinical specialists in stroke care. The system will focus on lower extremity rehabilitation exercises, and clinical trials for the system are planned for sit-to-stand task training. Although the authors present the system as one for motor relearning, due to the adaptive nature of the system, it can be used for skills training such as in sports, dance etc.

Work in the near future will focus on validation of the simulation results of limb tracking with actual experiments in the lab. The automated assessment and real-time advisory feedback modules need to be carefully designed so as to ensure the suitability of feedback for the target subject.

7. ACKNOWLEDGEMENT

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