Control of a Desktop Mobile Haptic Interface

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Abstract

Most haptic devices share two main limits: they are grounded and they have limited workspace. A possible solution is to create haptic interfaces by combining mobile robots and standard grounded force-feedback devices, the so called Mobile Haptic Interfaces (MHIs). However, MHIs are characterized by dynamical limitations due to performance of the employed devices.
This paper focuses on basic design issues and presents a novel (prototype) Mobile Haptics Platform that employs the coordination of numerically controlled wheel torques to render forces to a user handle placed on the top of the device. The interface, consisting in a small omni-directional robot, is link-less, fully portable and it has been designed to support home-rehabilitation exercises.
In the present paper we shall review relevant choices concerning the functional aspects and the control design. In particular a specific embedded sensor fusion was implemented to allow the device to move on a desk without drifting. The sensor fusion algorithm has been optimized to provide users with a quality force feedback while ensuring accurate position tracking. The two requirements are in contrast each other and a specific variant of the Extended Kalman Filter (EKF) was required to allow the device working.

Index Terms: I.2.9 [Artificial Intelligence]: Robotics—Autonomous vehicles; I.2.9 [Artificial Intelligence]: Robotics—Kinematics and dynamics; I.2.9 [Artificial Intelligence]: Robotics—Commercial robots and applications; I.2.9 [Artificial Intelligence]: Robotics—Sensors; H.5.2 [Information Interface and Presentation]: User Interfaces—Haptic I/O; C.5.3 [Computer system implementation]: Microcomputers—Portable devices C.3 [Special-Purpose and Application-based System]: Real-time and embedded systems—

1 Introduction

One of the major limitations in the use of haptic displays is their portability due to both energy and accuracy issues. These devices make use of motors to recreate the sense of perception with a large consumption of power. Further, they require precise positioning and force control systems to cope with the complexity of the perceptual system.
Portable haptics may belong to two different categories: wearable [3] or mobile [13]. With wearable haptic devices, typically hand or arm exoskeletons, the user moves the device sustaining its weight. On the other hand in Mobile Haptic Interfaces (MHIs), the traditional approach is to support a fixed haptic interface with a mobile robot that serves as power supply and transport platform [13]. This approach allows to extend the workspace, but on the other side typically it reduces the portability due to the whole system encumbrance. In addition, since the mobile robots are generally characterized by slow dynamics, delays may be present when attempting to track the human operator motion. As a consequence the operator may feel the boundary on the haptic device workspace, which in turn can create spurious forces and ultimately cause the total loss of transparency for the MHI [5]. Although these limitations, Barbagli [2] benchmarked these type of devices through a set of qualification procedures that demonstrated the adequacy of these systems to deploy a good force feedback.

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Figure 1: MOTORE: Portable Haptic Interface. The photo shows the device with main body in blue, the handle in black, that is grasped by the user and the forearm support in grey.

Several control improvements in these two research lines are currently under development. For wearable interfaces new control algorithms propose to improve the force feedback which is limited by the non-linearities introduced by transmission and joint stiffness [12]; for MHI, instead, the research focuses mainly on a better coordination of the mobile platform with the onboard haptics [10].
Anyway, the effective portability of the haptic display is limited due to the complexity of the wearing procedure in the first case, and the difficulties in carrying the device in the other one. These issues severely compromise the devices usability when portability is the primary requirement. In addition, MHIs generally have very high "'weight-force feedback"' ratio.
In the present paper we investigate the design and control of a mobile haptic interface which uses its own wheels to directly generate the force feedback. A target application of rehabilitation robotics has been addressed in order to derive a correct set of application requirements.
The paper will describe how major specification issues have been devised, and how major issues to control and properly coordinate the force feedback in the space have been solved. In particular a novel kind of sensor fusion between proprioceptive data and CCD based tracking has been employed. This technique, using only onboard electronics, allows a "potentially" unlimited workspace. Overall performance on basic haptic tests will be shown.

The remainder of this paper is organized as follows. Section 2 points out the design requirements of the proposed device. Section 3 introduces and briefly describes our haptic interface. The kinematic model as well as how the localization problem has been addressed will be shown. The section ends with an overview of the control algorithm that was implemented to dealt with rehabilitation tasks. Control aspects, i.e. control algorithm together with localization, represent the main contribution of this paper. The presentation of an experimental setup together with the discussion of experimental results follows in Section 4. Finally, in Section 5, conclusions are drawn and future work is outlined.

2 Design Requirements

The investigation in this paper has been carried out on the MOTORE system [15], a mobile haptic interface designed to be employed in home rehabilitation exercises. Due to the different weight factor and force generation concept, in what follows we will refer to MOTORE as a Portable Haptic Device (PHD) instead of a MHI. Given the target application we have challenging requirements to satisfy. First of all we want to obtain a lightweight device with reduced encumbrance on the working desktop. Secondly, the system has to be completely autonomous to assure a full portability feature. Moreover, the system has also to be easy to use in such a way that in order to work with the device, the user should only place it on the desktop and switch on the power button, as he would do with his notebook. These requirements bring to the design of a system with a reduced (or absent) calibration procedure. From the control point of view we need to design a control algorithm able to guarantee good position tracking as well as a very precise haptic feedback. In addition, the platform should have omni-directional kinematics to change the direction of motion without extensive maneuvering.
In upper limb robotic rehabilitation the MIT-MANUS [6] is currently being employed in most of the protocols and it has been proved effective for neuro-rehabilitation [9]. The specifications of MIT-MANUS, in terms of maximum velocity and maximum force, have provided to MOTORE a relevant basis to define rehabilitation protocols requirements. In addition, with respect to MIT-MANUS, MOTORE performance is not affected by the handle position and orientation, allowing to deploy easily large workspace span exercise, differently from base haptic platforms.
We considered a target maximum operational force of 35N sufficient to deploy most rehabilitation exercises. In particular, given the actuation group design, such a force is only limited by the weight of the device and the friction coefficient of the moving wheels.
The reference workspace, where the mobile robot can freely move, has been fixed to 1080x720 mm, only for convenience. As we will see later, this size has no potential limitation and it is only bounded by the support sheet that is being employed to host the device.
The system has to assure an entire rehabilitation session, thus we set a minimum operational autonomy requirement of 75 minutes.
The device inertia needs to be actively compensated through the introduction of a force sensor that cuts-off part of the device weight allowing to perceive a reduced reflected inertia. Moreover, the load cell must be capable to measure the interaction forces between the robot's handle and the user's hand.
Finally, the absolute position and orientation of the platform, with respect to a fixed coordinate system, is required by the type of exercises that are involved in rehabilitation, like trajectory training or pick and place operations.

3 System Description

The general system requirements, as stated in the preceding section, pose a number of interesting aspects which will be analyzed below.

Figure 1 shows the MOTORE prototype. The interface can be operated directly by the user. An handle on the top of the device (the black knob in the Fig.1) facilitates the grasping for unpaired

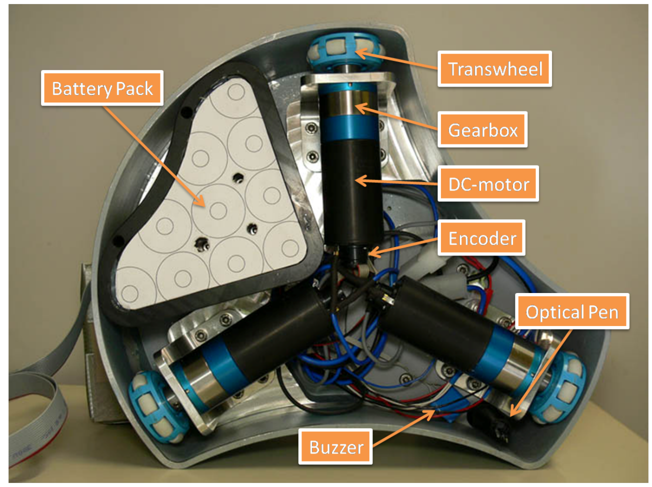

Figure 2: Bottom view of the interface. The setup with single Transwheels (Kornylak, OH, USA) has been replaced with couple of Transwheels in order to cancel vertical chattering which was introduced by the lack of regularity in the Transwheels profiles

people, while passive rotational support for the forearm improves device usability. A particular care has been given to the design of the forearm support in order to minimize the amount of forces exchanged on the motion plane between the support and the forearm. Bearings and backlashes inside the support allow a reduced planar motion with minimal interaction forces.
The proposed PHD consists in a omni-directional mobile cart actuated by three omni-wheels which are driven by three independent DC-motors[1]. Given the motor features, a reduction box was introduced to match the force/torque requirements at the user's handle. In addition, each motor is sensorized with an incremental optical encoder (512 CPR) which provides an extremely high accuracy of the wheel movements tracking (it is 20um/count). The overall MOTORE weight is about 10 Kg (batteries included).
The peculiarity of the system consists in the force feedback which is provided on the user's handle directly by the robot wheels. Moreover the system is completely link-less concerning both the mechanical and electrical parts. The power supply is provided by a battery pack that was designed to support a continuous supply of 12 Volts for a minimum amount of time of 75 minutes. This autonomy is being estimated as average power consumption when the device is moving continuously at the nominal velocity of about 1.15 m/s. Such a velocity was considered adequate for the considered rehabilitation protocol.
Table 1 summarizes major design information, while Fig. 2 reports a bottom view of the device which shows almost all the above mentioned system components.

On board there is a three axes accelerometer introduced both for critical condition detection and for performance improvement. In particular at the startup of the device the accelerometer is used to detect if the robot is being operated on a non-horizontal surface, whereas during the working session it is used to detect dangerous conditions, like too high acceleration on a specific direction. The accelerometer is also used for improving performance by dynamically compensate the robot's inertia.
As mentioned above a two axes force sensor was embedded in the robot handle to detect and measure the interaction forces with the user's hand.
Finally, the system embeds an optical pen to solve the localization

[1] The motor currents are provided by three H-bridges which are controlled through PWM technique.

Table 1: Major system parameters

Simbol	Value	Unit	Name
M	10	kg	Mass
I_z	0.1086825	$kg\ m^2$	Inertia
B	25.4	mm	Wheel radius
L	145	mm	Robot radius
N	676/49	/	Motor reduction gear
CPR	512	/	Motor encoder resolution
K	16.8	mNm/A	Motor constant
I_{mot}	1.18 10^{-5}	$kg\ m^2$	Reduced inertia of actuation group
$[X_p, Y_p]$	[124, -40]	mm	Anoto Pen position

problem, and later we will see the importance of this element in the sensing.
The system intelligence is provided by a DSP and the control architecture is organized exploiting a new DSP micro evaluation board commercialized by Texas Instruments$^{\text{TM}}$in 2009. The evaluation board integrates a TMS320F28335 running at 150MHz, with a 32bits floating point unit, a fast and accurate PWM control, an integrated encoder acquisition units, 68/512Kbyte RAM/Flash for memory and programming, low pass filters for analog acquisition, all in a DIMM size card. This DSP board allows handling the whole control scheme at a base rate up to 10kHz.
Concerning the communication, a Bluetooth module has been interfaced to the DSP in order to establish a bidirectional communication with the remote console PC. The MOTORE system receives by Bluetooth the command inputs and updates concerning the absolute position information, while at the same time, it sends back scoring information suitable for the rehabilitation exercise, like for example the interaction forces.

3.0.1 Safety

The risk to patients is minimized by an active supervising control and an easy to access shutdown button. To ensure safety, the device physically limits the range of motion in absolute coordinates. In addiction force feedback is limited both by a software saturation and the intrinsically nature of the actuation system due to the upper limit in the wheel friction coefficient. Our assertion on the safety is based on the fact that the upper bound limited force feedback and speed of motion, determined by careful test and evaluation of patient data, will not impose excessive force on a patient if a muscle spasm prevents the arm from reaching the desired position.

3.1 Kinematic Model

MOTORE kinematics is based on the "Killough's mobile robot platform" [14]. Three-couples of Transwheels [16] are placed on the circumference contour with their axes oriented at 120 degrees and incident in the device center.

The three contact points given by Killough's kinematics ensure that the contact with the support plane is always isostatic. It is the simplest layout with no kinematic redundancy, in addiction such a layout is able to compensate for the small planar deflections which can be present in the supporting surface.

The derivation of the kinematics model is well known in literature and some references can be found in [4] and [11]. Here we only summarize the convention employed in the following and the major relevant formulas.
Figure 3 reports the two reference systems used to identify world coordinates (O-frame) and local coordinates (B-frame). In particular, the B-frame is fixed with the mobile device and has its origin in the geometrical center of the robot. Wheel 1 is the one below

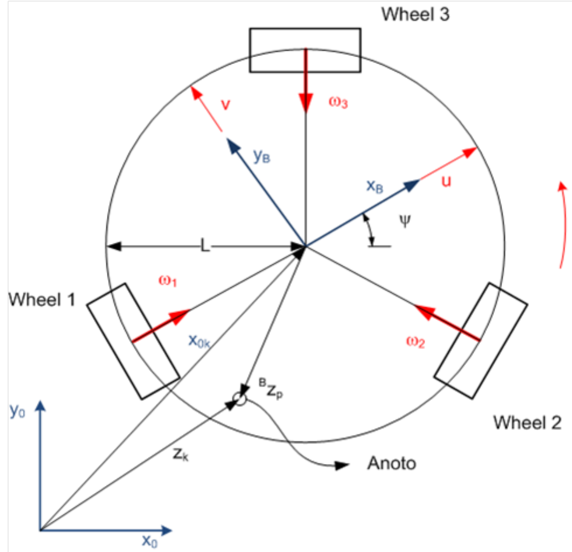

Figure 3: Kinematic diagram of the system depicting the three wheels, the absolute position tracking device, and the reference systems.

the user handle, other wheels are numbered in counter-clockwise order (from top view). $[x, y, \psi]$ describes the posture of the robot with respect to the absolute fixed frame, while $[u, v, r]$ describes local linear and angular velocities in accordance with the body frame. Finally, $[\omega_1, \omega_2, \omega_3]$ describes the angular velocities of the three motors. It has to be noted that the optical pen position is placed at coordinate $[X_p, Y_p]$ with respect to the body frame. With the above convention and the parameters defined in Table 1, the local Jacobian can be expressed as in the eq.(1):

$$\begin{bmatrix} u \\ v \\ r \end{bmatrix} = \frac{B}{3NL} \begin{bmatrix} 0 & L\sqrt{3} & -L\sqrt{3} \\ -2L & L & L \\ 1 & 1 & 1 \end{bmatrix} \begin{bmatrix} \omega_1 \\ \omega_2 \\ \omega_3 \end{bmatrix} \tag{1}$$

Given the system simmetry, we can easily obtain the absolute Jacobian as in eq.(2):

$$\begin{bmatrix} \dot{x} \\ \dot{y} \\ \dot{\psi} \end{bmatrix} = \frac{2B}{3N} \begin{bmatrix} \sin(\psi) & \sin\left(\psi + \frac{2\pi}{3}\right) & \sin\left(\psi - \frac{2\pi}{3}\right) \\ -\cos(\psi) & -\cos\left(\psi + \frac{2\pi}{3}\right) & -\cos\left(\psi - \frac{2\pi}{3}\right) \\ \frac{1}{2L} & \frac{1}{2L} & \frac{1}{2L} \end{bmatrix} \begin{bmatrix} \omega_1 \\ \omega_2 \\ \omega_3 \end{bmatrix} \tag{2}$$

3.2 Localization problem

A mobile system like MOTORE requires an absolute and precise localization in the environment for supporting the proposed rehabilitation exercises. These two properties of localization are difficult to achieve in combination, in particular with the additional constraint of no calibration phase. For this reason, precise system localization cannot be solved by means of a single measuring method. Therefore, several measuring systems as well as methods for sensory data fusion have to be applied. Odometry and dynamic system models provide the desired relative accuracy together with sufficient bandwidth. On the other hand, a localization method based on optical or acoustic sensor systems provides the desired absolute accuracy.

The use of Kalman filtering is well known to solve the localization problem of mobile robots. Standard and multi-rate, extended and unscented Kalman filters have proven to achieve increasing levels of accuracy [1], [8]. However, the user's interaction required in a haptic force generation poses additional constraints to the position reconstruction provided by Kalman filtering. Achieving the best interaction here means to be able to mediate between the position accuracy (reconstruction achieved by Kalman filtering) and the quality of force feedback.

In our case to track the robot absolute position and orientation on the sliding surface, we combine the information coming from proprioception with the one provided by a sensing pen placed inside the device. To this purpose, we have re-engineered a commercial digital pen developed by Anoto®[7] that enables accurate tracking of pen position on a bi-dimensional bar-coded sheet. The Anoto pen incorporates a CCD, that computes in real-time the position from patterns, and a Bluetooth transmitter that sends such an information to a remote system. This solution, under the assumption that the desk is the workspace and the sheet is correctly placed in front of the user, has an advantage respect to camera based optical or acoustic tracking because there is no need for external cameras or calibrations. Anyway the proposed fusion technique is almost independent of the absolute tracking system and it is quite general. The algorithm adopted here is an Extended Kalman Filter which will be modified during the experimentation to take into the account the specific application requirements who the system is focused too and mainly the interaction with the human being.
In state estimation, the system dynamics is represented by a nonlinear discrete-time equation referred to as the process model:

$$x(k) = f(x(k-1), u(k-1)) + w(k-1) \tag{3}$$

where $x(k)$ is the state at time k, $u(k)$ is the control input, and $w(k)$ is a zero-mean, white, Gaussian process noise with covariance matrix $Q(k)$. $f(\cdot)$ represents the relation between variables in $(k-1)$-th step and $x(k)$, i.e. it is the state prediction based on the actual state and inputs. The state prediction equation is obtained integrating eq.(2):

$$\begin{bmatrix} x_{0k} \\ y_{0k} \\ \psi_k \end{bmatrix} = \begin{bmatrix} x_{0k-1} \\ y_{0k-1} \\ \psi_{k-1} \end{bmatrix} + \frac{2B}{3N} \begin{bmatrix} \sin(\psi_{k-1}) & \sin\left(\psi_{k-1} + \frac{2\pi}{3}\right) & \sin\left(\psi_{k-1} - \frac{2\pi}{3}\right) \\ -\cos(\psi_{k-1}) & -\cos\left(\psi_{k-1} + \frac{2\pi}{3}\right) & -\cos\left(\psi_{k-1} - \frac{2\pi}{3}\right) \\ \frac{1}{2L} & \frac{1}{2L} & \frac{1}{2L} \end{bmatrix} \left(\begin{bmatrix} \Delta\theta_{1k-1} \\ \Delta\theta_{2k-1} \\ \Delta\theta_{3k-1} \end{bmatrix} + \begin{bmatrix} w_{1k-1} \\ w_{2k-1} \\ w_{3k-1} \end{bmatrix} \right) \tag{4}$$

The measurement equation is:

$$z(k) = g(x(k), u(k)) + v(k) \tag{5}$$

where $z(k)$ is the measured value by means of the Anoto Pen, $w(k)$ is a zero-mean, white, Gaussian measurement noise with covariance matrix $R(k)$, and $g(\cdot)$ is the relation between the state and the measured value, which is:

$$\begin{bmatrix} z_{1k} \\ z_{2k} \end{bmatrix} = \begin{bmatrix} x_{0k} \\ y_{0k} \end{bmatrix} + \begin{bmatrix} \cos(\psi_k), -\sin(\psi_k) \\ \sin(\psi_k), \cos(\psi_k) \end{bmatrix} \begin{bmatrix} X_p \\ Y_p \end{bmatrix} + \begin{bmatrix} v_{1k} \\ v_{2k} \end{bmatrix} \tag{6}$$

With our setup the prediction phase is executed at 1 kHz whereas the correction at 50 Hz, due to the limitations in the optical pen. Every time a correction step is applied the estimated position shows a step behavior which due to the force feedback law[2] results in a force impulse on the robot handle. Since we want to maximize the quality of haptic feedback, we modify the standard EKF in order to apply the estimate correction in a smoother way than applying the position correction immediately. The innovation is applied gradually between two consecutive update steps of the Kalman algorithm.

[2]Like, for example, the classical virtual coupling that consists in a force proportional to measured displacements of the haptic interface end-effector.

3.3 Control loops

Rehabilitation therapy requires high fidelity force feedback and smooth movements. The most developed paradigm for rehabilitation is the assistive one. Assistive controllers help participants to move their weakened limbs in desired patterns during grasping, reaching, or walking, following a strategy similar to "active assist" exercises performed by rehabilitation therapists.
Thus, the control algorithm has to provide force and position control and it has to respond to fail safe criteria. Because we are interested in a quantitative evaluation of task performance, the implemented algorithm has to provide accurate position and force measurements.
The embedded control unit runs both the low level control (sensors measurement and motors control) and the high level control (logic control for phase detection, safety checks and etc.). We now shortly present the low level control architecture which is schematically represented in Fig.4
The basic function of the low level control is to reconstruct the robot position (using the EKF previously introduced) and to provide the force feedback to the user handle given the device displacement from a target position.

The control algorithm is mainly composed by three closed loops and by two open loops, i.e. inertia compensator and torsion compensator. The inertia compensator based on the measured acceleration and the dynamic system parameters gives the required wheel torques to compensate the inertia of the MOTORE system. Similarly, the torsion compensator, based on the interface model, provides the required wheel torques to compensate the disturbance generated by the user interacting with the robot handle which is misaligned from the center of mass.
From the faster operation frequency to the slower one the closed loops are the current controller, the velocity controller and the position update sub-system.
The current controller is the fastest loop in the system, it runs at 5 kHz. Such a controller consists in an integrator plus a feed forward component. It aims to regulate the motor current in order to provide the desired force feedback.
The velocity controller consists in a proportional-integral controller that runs at 1kHz. It generates the wheel torques to track the desired velocity provided by the "feedback generator" sub-system. The position update sub-system is the slowest loops and it runs at about 50Hz as introduced in the previous section.
The "feedback generator" implements the suitable algorithms to cope with the rehabilitation therapy. The system has the capability to allow both impedance and admittance controllers, and changing the feedback generator allows to personalize the device for the application in other fields.
Anyway in our case, given the measured interaction force, the actual device posture and the commanded exercise modality, the algorithm provides the desired velocity to be tracked. In particular, we implement an admittance control law along the desired direction and an impedance control law along the orthogonal one. For the sake of simplicity consider a straight line. If a user push the robot handle along the line, the measured interacting force passing through an admittance filter generates a desired velocity which will be passed to the velocity controller. On the other hand if the user push the robot handle along a different direction, only the projection, i.e. the amount of force along the straight line will be passed to the admittance filter. Depending on the patient injury, the orthogonal amount of force can be neglected or attenuated before entering

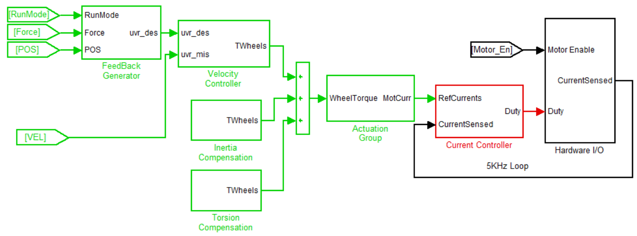

Figure 4: Scheme of the MOTORE Control Loops. The control architecture consists in three different operating frequencies. In red the current loop running at 5kHz, in green the velocity loop, the feedback generator, the inertia and the torsion compensator running at 1khz. POS signal comes from the EKF algorithm, VEL and Force are respectively the velocity and the force measurements. RunMode is the signal received from the Bluetooth communication which controls the system behavior.

the admittance filter.
Moreover, a speed booster based on the position error between the actual device position and the desired one is given. In addiction, the minimum driving force was set to 0.15 N by a digital limitation introduced in the control loops to cope with user requirements.

4 Trajectory Tracking Example

To prove the system effectiveness and to setup a preliminary usability test with patients we setup a user-friendly Graphical User Interface to easily cope with the MOTORE system.
The exercises available with this setup consist in training trajectories. During each exercise the patient has to follow a path shown on the screen in front of him. The robot, accordingly to force interaction and current position on the desk, can help the patient to start the movement or to avoid going in the wrong direction, just like a human therapist would do. Figure 5 shows a snapshot taken during the clinical trails.
Some preliminary trial results of a good health user are shown in Figure 6. In particular the user had to follow the desired trajectory (the red one in the picture), while the robot tried to compensate the wrong movements. It has to be noted the excellent repeatability of the user's trajectory (the blue one in the picture) and the lack of drift in the robot position estimation. More in details we can distinguish the exercise starting phase where the robot being in a random position moves toward the exercise starting point [0;0.15] (the "vertical" movement that crosses the upper half of the "8"). The trajectory deviation near the point [-0.07;-0.02] occurred when the user tried to move the his/her arm in the wrong direction. The system, according to the level of compensation set, exerted the force feedback to guide the user toward the reference path. During the experimentation we observed that the robot is able to provide the force feedback up to 30N.

5 Conclusion

In this paper we have discussed the concept and related design issues of a novel portable haptic device, which is completely autonomous for both actuation and control aspects. The presented device consists in a small omni-directional mobile robot able to follow the user movements exerted on the robot handle by its resting hand in a large workspace[3]. The interface can also drives or correct user's trajectory through a force feedback which is generated directly by its wheels.

[3] Ideally infinite workspace.

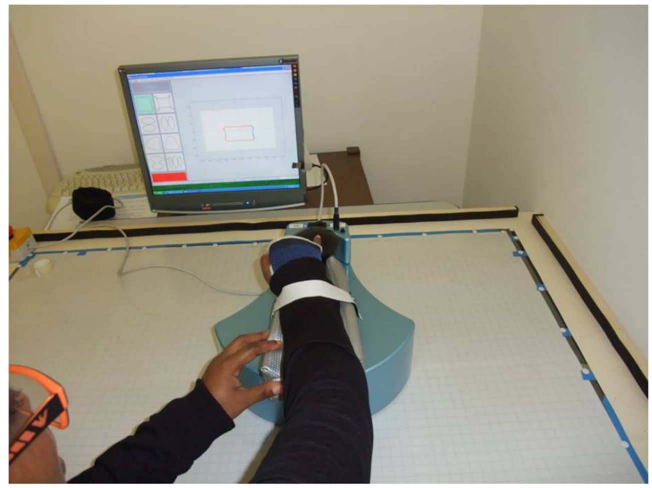

Figure 5: Upper limb rehabilitation using the MOTORE system. The training trajectories can be selected by the GUI which on the other side shows in real-time both system status and useful exercise information, like the actual position, the desired one, the interaction force and so on.

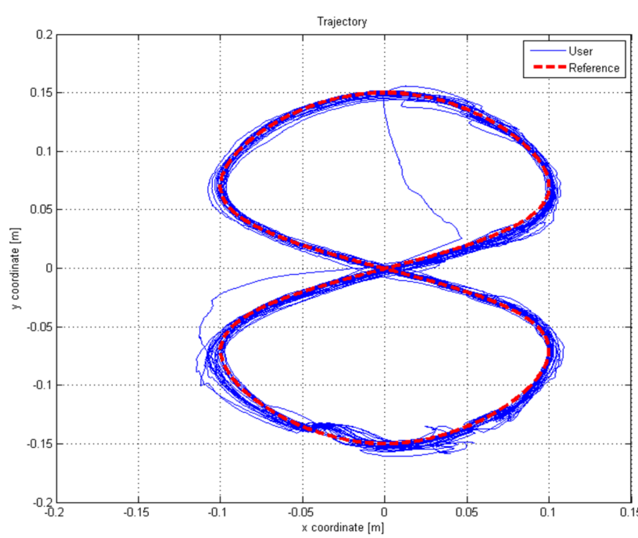

Figure 6: Trajectory tracking example. The user try to move along the reference path (the red line) while to robot compensates the position error with an amount of force feedback.

Our system has been oriented toward robotics rehabilitation thus during the design phase we paid particular attention in order to provide a low cost, safe and easy-to-use, robotic-device that assists the patient and therapist in order to achieve more systematic therapy. Force feedback, audio and visual feedback are used to increase the patient motivation. In addition, given the system characteristics, it could be used for home rehabilitation where the patient performs by himself the rehabilitation protocol and sends to the therapist the exercise results. Such a situation could improve the domestic rehabilitation since generally it has not a direct medical control. New rehabilitation modality like tele-rehabilitation or therapist tele-supervised rehabilitation begins to be attractive.

During the exercise, the patient is driven along the correct path of the rehabilitation protocol, but what is interesting is that the patient is actively involved into the exercise because the system requires a minimum amount of force in order to be activated (around 0.15 N[4]). Active voluntary movements seem to be useful for upper limb motor recovery.

The system can be indifferently used with the right arm or the left one without any reconfiguration procedure (both hardware and software). More specifically, to work with the system it is only required to put it on the support plane and to switch on the power button. Since there is not particular procedure to be accomplished before starting the exercise (like for example calibration procedures or setting up time) the interaction with the system is very user-friendly.

We are currently conducting a feasibility pilot study in order to point out if there are some usability limits. After that a wider experimentation for a full evaluation of effectiveness will be performed.

For the future we would like to improve the estimation procedure in order to cope with the time delay between the position measure and the odometry information, different data acquisition frequencies and control robustness. To address the delay problem, the idea is to implement a more sophisticate EKF that correlates the absolute position measure not with the current position estimation but with the estimation referred by the pen.

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