MOTORE: a Mobile Haptic Interface for Neuro-Rehabilitation

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***Abstract*—The present paper introduces a novel type of haptic interface which is fully portable and employs only onboard sensors and electronics to solve accurate localization and force feedback generation. The device offers 2DOF force control while sliding on a plane and maintaining its orientation comfortable for the user. The device generates force feedback information without any intermediate link between the motion wheels and the grasping handle. The device has been designed for application in neuro-rehabilitation protocols and it adopts specific mechanical, electrical and control solutions in order to cope with patient requirements. The paper describes the mechanical and electronic solutions as well as the most relevant features of control implementation issues that were addressed in the system design.**

I. Introduction

Rehabilitation robotics is a growing practice in hospitals to personalize therapies with goal directed tasks. These devices increase the amount of rehabilitation practice and possibly they improve the quality of medium term rehabilitation [1-4]. MIT-Manus [5,6] has been the first device to be massively experimented for robotic based rehabilitation. To date most robotic rehabilitation devices are bound to be employed in hospital but they require the preset of several environmental conditions. These devices have large power consumption due to motor and long linkages, and they require a solid grounding to transmit the haptic feedback to the ground. Usually such systems use proprioception as an accurate localization mean, or alternatively they use external localization tools, which are embedded in the environment, to have an accurate feedback on the device position.

With these premises, the use of such tools as instruments for long self-rehabilitation exercises is difficult. In addition, the cost of patent's home personalization for hosting the tools as well as the safety issues connected to the proper and autonomous use of the device, compromise the rapid spread of them [7].

A natural solution to these issues seems to make such devices portable, removing rigid connection to the environment and self powering the motors with batteries. However making these devices mobile introduces several issues to be handled.

The most known limitation is in the accuracy of localization. Wheeled devices have severe slip issues [8,9], which propagate odometry errors on localization through integral of simple differential motions. Usually, in order to attenuate these errors, Kalman Filters (KF), Extended Kalman Filters (EKF) or Unscented Kalman Filtering (UKF) are adopted [10-11]. These algorithms allow to obtain a better estimation of the robot location by improving trajectory control of these units. However, to be effective such algorithms require a "complementary sensor", i.e. another source that is not affected by the same type of differential error, usually providing absolute position information.

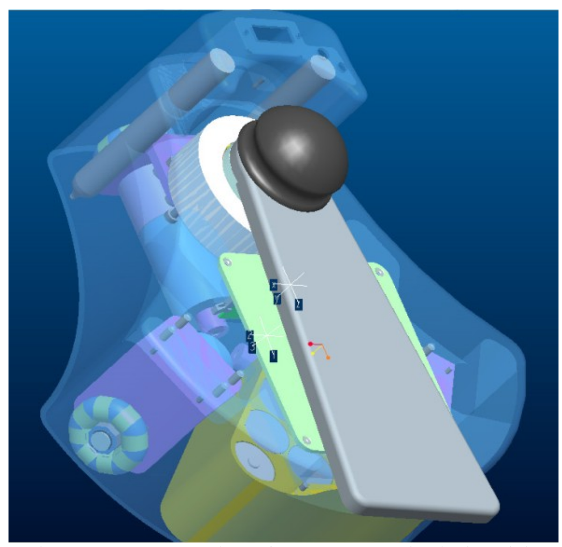

Fig. 1. A transparent view of MOTORE mechanical model.

In addition, small positional errors of even few millimeters produced during haptic feedback are able to cause high force disturbances when typical stiff walls are simulated (2÷5N/mm).

A common approach used in the field of the Mobile Haptic Interfaces consists to decouple the force generation through the introduction of intermediate links that provide a force feedback which is unconstrained to the device motion control [12,13]. However, as a counter effect, the increased mechanical complexity introduces limitations on the overall performance.

In the following there will be introduced the design guidelines of the MOTORE system. MOTORE stands for

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MObile roboT for upper limb neurOrtho REhabilitation and it is a novel link-less haptic device that employs its case to interact with the user and its wheels to generate force feedback. The device is specifically conceived for tele-operation and robotic rehabilitation applications. With respect to existing systems, such as the MIT-Manus, the portability of MOTORE allows to extend the therapy periods at home without any specific setup. MOTORE has been conceived to be transported and employed as-is, and the therapy operation has been embedded by coding it in the therapy sheets employed for rehabilitation. MOTORE includes an onboard computing unit, an odometry system (based on encoders) and a specifically designed global localization system that recognizes patterns on the sliding surface.

The present work is organized as follows. Section II covers the specific requirements which should be added to an haptic system to achieve fully mobile capabilities. Section III reviewes the major design choices that allow the implementation of the introduced functionalities, while preserving standard requirements of classic haptic devices. Section IV introduces some kinematic elements that are required for the task generation which is shown in section V. Finally, in section VI a brief overview of the force/motion control is given (more details can be found in [18]).

II. MOBILE HAPTIC USER REQUIREMENTS

Notwithstanding the large amount of rehabilitation devices existing in literature, a set of device specifications is still unknown when a fully mobile robotic device is taken into consideration. The force and velocity specifications common to rehabilitation haptic design are expanded with additional elements developed in cooperation with the Auxilium Vitae Center in Volterra-Pisa.

The device has been designed having in mind users with severe neurological handicaps (lesions to CNS, cerebellar lesions,…), in whose cases the rehabilitation protocol is highly expensive in terms of follow up time and frequency. This type of patient does not benefit of isokinetic rehabilitation systems but requires goal directed rehabilitation which could be adapted to specific per-patient requirements. The device should therefore be able to generate a force feedback that changes its behavior according to the underlying task, its phase and the interpretation of the cognitive state of the subject. A smart onboard electronic system is hence required to perform task related analyses in parallel to the control computing.

At a first stage, for the rehabilitation of patients with neurological impairments, is preferable to employ planar devices instead of those exploiting three dimensional workspace. Devices operating on a plane reducing the muscular fatigue will allow to improve the concentration on motion and coordination tasks.

To extend the amount of available rehabilitation time for the patient, the device should be portable, able to operate in any place and on any table, provided that the slope of the sliding surface is close to horizontal plane. To prevent potential risks for the patient, internal sensors and a self-diagnostic should allow the device to analyze the sliding surface properties and to prevent the use on improper surfaces.

To assure the portability, the motion should be self powered through a battery set placed within the case. The device should be self chargeable with a direct connection to a proper AC adapter and monitoring LEDs should facilitate the recharge operation. This feature allows the relatives to assist the patient in multiple rehabilitation cycles without needing for specialized personnel and/or the risk of damaging the device. In addition, the device should monitor its own charge to avoid the misuse in critical battery conditions. Considering a continuous use of the device, the autonomy for a single full charge should be greater than 70 minutes, which is close to the rehabilitation tasks operated in clinic. The device may be recharged while the user is resting. The overall weight of the device, including the batteries, should be within 10 kilograms for transportability reason. The mass during the rehabilitation exercises should be compensated actively by the control in order to decrease the patient fatigue. A particular attention to the maximum acceleration allowed to the device as to be considered. The peak acceleration has to be contained around 1 m/s^2 to limit spasms in patients.

Finally from the portability and the usability point of view, the system should be also autonomous for the computing aspect. The adoption of an embedded control unit is required and the requirement of an external PC should be minimized. Wireless connections between the system and an external PC used to visualize a virtual environment as well as to provide a visual feedback to the patient, should be useful to provide a total link-less system.

Concerning the force feedback to guide the patient performing the correct movements, an average maximum force of about 25÷30 Newton is required during rehabilitation protocols. To avoid that slow exercises can result annoying for the patients, a maximum average velocity of 0.5 m/s is required. It is also important to take into consideration patient needs and spasms, and to make the system operating only when the user hand is grasping the device. Hence the system should be provided with a contact sensor to detect the user grasp. Finally, a support tool limiting the operational workspace to a safe region is relevant for home rehabilitation.

Concerning the localization problem, the device should be provided with an absolute localization mechanism which prevents drifts and rotation during pure odometry motion reconstruction. The potentially unlimited workspace should allow the therapist to design motion trajectories which are more user-oriented than device limited. On the other hand, the system should provide a simple programming mode that enables the therapist to personalize rehabilitation tasks with different protocols.

The therapy protocols may include active and passive trajectory following operation, as well as more complex tasks, cognitive associations, labyrinths and computer interacting tasks where the device is used as a smart force feedback mouse.

Such protocols should be easily interchangeable and, most important, they should not require any programming skill from the patient or his relatives. This could be achieved by embedding the protocols in portable elements (like sheets or memory cards) that mask the burden of device programming.

For a better assessment, the activities carried at home should be monitored and recorded for future analyses and progress monitoring. Using a force/torque sensor placed within the device handle, the user forces should be recorded during the whole rehabilitation therapy. Thus the system should provide an internal memory to store the suitable variables describing the user performance in the rehabilitation training. Force data as well as position error, can be retrieved and analyzed to improve parameters of the rehabilitation protocol.

III. ELEMENTS OF DESIGN

This section presents the design elements of the MOTORE interface in terms of mechanics, electronics, sensing capabilities and dimensioning of motors. Figure 1 presents a transparent view of the mechanical model, detailed by the schematic of Fig. 2. The basic kinematics is provided with three omnidirectional wheels (Kornylak Corporation Transwheel), that are arranged in a 120 degrees configuration having their axes converging in the center of the device following the Killough type kinematics [14]. Each Omniwheel is composed by a double row of roller to avoid hopping during translational motions.

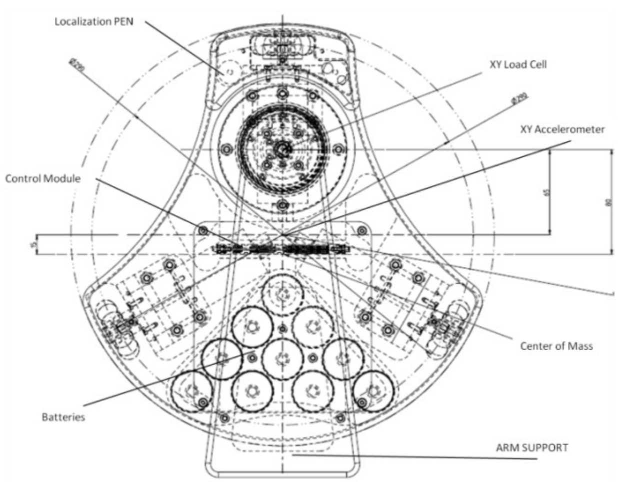

Fig 2. Vertical view of the device design.

The user handle is delocalized toward front, in order to provide a comfortable support to the user forearm and to achieve a better weight distribution, see also Fig.7 that shows a real usage of the device. The handle is supported by a flange that is constrained to rotate using cylindrical bearings around the vertical handle axis. This solution allows a frictionless rotation of the forearm support that improves the user comfort while operating the device. Moreover, when the user leans on the flange, the weight of his hand and the weight of his forearm are used to increase the total weight of the device and so, to improve the amount of force that may be transmitted by the wheels through pure friction with the sliding surface.

A two axes load cell (Pre-amplified with 0.2% linearity on full scale) has been placed in the device handle in order to measure the planar forces interchanged with the user hand. The cell gains and the computing unit have been calibrated to achieve 0.01 Newton accuracy on a full scale of 40 Newton. This load cell represents the control input provided to the user to drive the system.

The choice of the computing unit was particularly critical in order to ensure that the onboard electronics was low-power consumption but at the same time powerful enough to handle typical application of robotic rehabilitation. The micro control has been selected to cope with several requirements: 6 ADC/12bit for force sensor, battery level and motor currents monitor; 4 digital inputs; 3 PWM control modules with related control lines; 3 LEDs; 3 encoder inputs; 1 serial port for Bluetooth communication and 1 SPI port for accelerometer reading. In addition, large onboard flash memory and RAM were preferable to allow prototyping and delivery of control firmware on the fly. Finally, 32-bit architecture and hardware floating point unit were required to perform the sensor fusion and the feedback control at the minimum frequency of 1kHz.

Only two micro controller classes were found to comply with almost all the requirements: the novel generation of MPC5X5X, and the novel control DSP F28335 from Texas Instrument. The latter was chosen for the side advantages offered by the Delfino™ control stick which implements in a single low cost module additional features such as: JTAG interface port for debug and programming, Matlab/Simulink development environment, on-board low-pass filters for ADC inputs, USB-to-Serial channels for programming and debug, and finally high operational frequency (150MHz). Detailed specifications can be found in [21].

The localization of the system has been achieved by the adoption of a CCD sensor below the case of the device. The sensor operates with a proprietary technology from Anoto® that ensures accurate localization in a very large displacement [20]. Anoto sensors are primarily employed for creating digital pens that precisely track and record the writing pattern by detecting the position of the pen over a purposely printed sheet of paper. In order to be used in our system, an Anoto pen has been re-engineered as a positional real-time sensor. One of these pen sensors is placed in a well defined position inside the MOTORE case and it is oriented along the vertical axis in order to observe perpendicularly the sliding sheet. An illumination LED has also been added to ensure that an adequate illumination is present even below the interface case. In such a way the system uniquely recognize the XY position of the CCD with respect to the observed pattern. This technology has been verified being reliable and noise tolerant enough to be employed in combination with the odometry for an accurate and robust localization of the device.

The motion of the device is powered by three identical motors with a maximum power of 90W and a maximum continuous torque of 110 Nm. This choice allows to achieve, for each wheel, a maximum traction force of $F_{max} = N_{max}\eta_{el}\eta_{mecc}N/B = 0.11 \cdot 0.85 \cdot 0.8 \cdot 14/0.025 = 41.88N$, where have been considered the mechanical and

electrical efficiency, the relative reduction ratios of the gear (N) and the wheel radius (B). An additional reduction ratio of $cos(\pi/6)$ between the peripheral wheel velocities and the corresponding linear device velocities (XY) is introduced by the cart Jacobian. This results to a nominal maximum force of $F_{mot} = 36N$.
A similar result is achieved when comparing the expected weight of the system (device weight plus human forearm) with the friction coefficient. Assuming a nominal forearm/hand weight of 1.5kg and a friction coefficient of $\mu = 0.36$ [15], obtaining $F_{frict} = 0.36 \cdot 9.81 \cdot 11.75 = 41.5N$.
Concerning the power supply, ten batteries with a voltage level of 1.2V and the capacity of 3.0Ah, provide 36Wh of energy that allows the device to operate in a wide autonomy. This energy is used to power the motors and all onboard electronics like the DPS, the drivers and the sensors. The design autonomy is 75 minutes, estimated by the use of motors while generating in sinusoidal force patterns having the peak force as amplitude, and velocity of 1 m/s. This is assumed to be the worst condition while preliminary evaluations have shown that autonomy during experimentation goes typically beyond the 2.5 hours. The following table reports the main characteristics of the interface:

M = 10.25 kg	Device overall mass
$\begin{bmatrix} 10.00067 \\ 32.78035 \\ 12,16308 \end{bmatrix}$	XYZ center of mass location in (mm) with respect the mobile reference frame (B)
$\begin{bmatrix} 0.027 \\ 0.0042 \\ 0.0875 \end{bmatrix}$	Principal inertia moment frame orientation expressed as RPY angles (RAD). The frame is almost parallel to the main frame
Iz= 0.1086825 Kg m^2	Device inertia on the vertical axis
B = 25.4mm (1inch)	Omniwheels equivalent radius
L=140.5cm	Device radius (from center to wheel planes)
N = 676/49	Motor gears reduction ratio
2 * Delta= (π/3)	Angle between Wheels axes
K = 16.8mNm/A	EMF constant of motors
$I_{mot} \approx$ 1.18e-5 Kg m^2	Motor/gears/wheel equivalent inertia

IV. Kinematics

A fixed reference system {0} is used to model motion on the sliding sheet, and a local reference system {B} is placed on the device. Figure 3 summarizes most of the conventions fixed for the mobile device and introduces the position and orientation of reference frames used during the device modeling. The explanation of these symbols is given in the following table:

F_1, F_2, F_3	Wheels traction forces
$\omega_1, \omega_2, \omega_3$	Motor angular velocities
X_p, Y_p	Pen position in the mobile reference frame (B)
Q_1, Q_2, Q_3	Motor torques
u,v,r	Velocities in the mobile reference frame (B)
X_B, Y_B, ψ	Mobile reference frame on the device
X_o, Y_o, ψ	Absolute position and orientation of the device
C_x, C_y	Contact force (torque is zero)

The fixed reference frame has been defined originating in the center of the sliding sheet. It is assumed that XY-axes of such a frame are oriented in the transverse plane of the patient, X axis laying in the intersection with the coronal plane (positive direction toward right) and Y axis laying in the intersection with the sagittal plane (positive direction toward front). The motion kinematics of the robot is derived in a discrete form based on the sampling time of the controller (1kHz). Sensors redundancy is provided by the encoder measurements and the optical pen. Both of them have a mathematical relationship with the device state according the following equations [16]:

$$\begin{bmatrix} x_{0k} \\ y_{0k} \\ \psi_k \end{bmatrix} = \begin{bmatrix} x_{0k-1} \\ y_{0k-1} \\ \psi_{k-1} \end{bmatrix} + J \begin{pmatrix} \Delta\theta_{1,k-1} \\ \Delta\theta_{2,k-1} \\ \Delta\theta_{3,k-1} \end{pmatrix}$$

$$z_k = \begin{bmatrix} \cos(\psi_k) & -\sin(\psi_k) & x_{0k} \\ \sin(\psi_k) & \cos(\psi_k) & y_{0k} \end{bmatrix} \begin{pmatrix} x_p \\ y_p \\ 1 \end{pmatrix}$$

where J is the motion Jacobian matrix that is defined as:

$$J = \frac{BT}{3NL} \begin{bmatrix} \cos(\psi_{k-1}) & -\sin(\psi_{k-1}) & 0 \\ \sin(\psi_{k-1}) & \cos(\psi_{k-1}) & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} 0 & L\sqrt{3} & -L\sqrt{3} \\ -2L & \mathrm{L} & L \\ 1 & 1 & 1 \end{bmatrix}$$

Here is omitted the influence of the sensor noises that provides indetermination on the localization. Such a indetermination is corrected adopting the Extended Kalman Filtering [17]. Tests performed on the EKF, at a constant speed of 0.2m/s, demonstrate the ability of the filtering procedure to maintain the average error below 2mm.

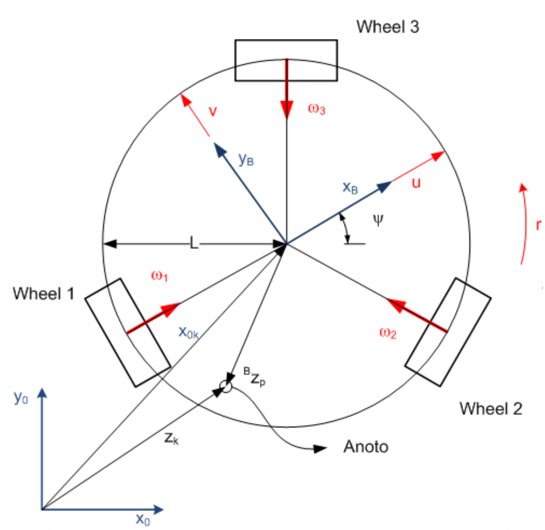

Fig. 3. Basic kinematics and reference frames.

V. TASK GENERATION

The robot behavior, according to the operation phase, is handled by the behavior generation module that is embedded onboard into the control unit. The various behaviors are implemented as Simulink/Stateflow charts or, for more complex features, as S-functions. In the experimental system, a basic trajectory constraint with pre-planned force/velocities guidance is implemented. The geometric constraints can be programmed through point based 2D-splines. The inner levels of control automatically reconstruct these trajectories computing the cubic C3-splines [19]. A 2D Newton-Raphson method is implemented on such splines to track the closest proxy point to serve as feedback generator. The specific choice of the spline model avoids discontinuities in the acceleration information while tracking the whole trajectory. Point-based trajectories also allow to scale, translate and rotate the geometries with low computational cost. Hence, an option in the controller allows to scale up and down the rehabilitation trajectories in order to fit with all user sizes.

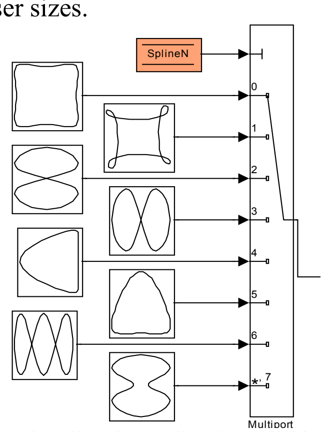

Fig. 4. Sample spline shapes directly available on-board

More in details, eight types of splines have been preloaded in the controller as shown in Fig.4. Each of the generated trajectory has been designed in order to limit the minimum curvature all over the spline. Such a condition has been considered essential for the proper stability of the proxy algorithm as well as to limit the inertia effect during constant velocity control. Average size of the splines ranges from 30 to 70 cm accordingly to the chosen scale factor.

VI. DEVICE CONTROL

Device control is based on several nested loops. At the inner level the current feedback control determines the PWM duty cycle to control the current flowing within the motors. This loop is arranged at a frequency of 5kHz and makes use of the current monitors available from the H-Bridges, see Fig.5.

The current regulation is closed on a local velocity controller that enables the platform to follow specific reference velocities in the local coordinate system.

According to this structure of control, device velocities can be accurately estimated using only the proprioception system and, in particular, without requiring that the Kalman correction is operating. The choice of this configuration

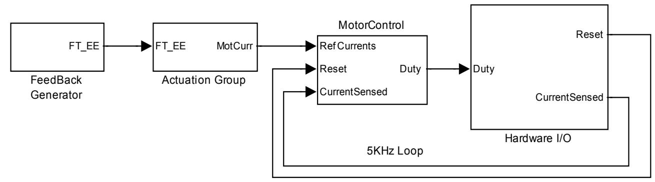

Fig. 5. Control loop structure: the internal loop is running at 5kHz and monitors the current flowing within the motors to generate a proper Duty cycle for the H-Bridge. To avoid aliasing, a LP first order filter is placed to reduce sensed current bandwidth to 4.2 kHz.

allows to implement a safe behavior in the absence of pen information (due for example to large Bluetooth's lags) and in presence of accelerometers alarms (like the ones caused by inclined sliding slopes). This architecture is particularly indicated for haptic devices which interacts with human body, since it does not require any external control in order to prevent accidental falls.

The outer loop of the control interacts with the generated splines and with the low level controller. The outer loop takes into the account some open loop compensations, for example the compensation of the device inertia. Figure 6 depicts all the subsystems involved in the generation of the wheel torques.

At the core of the outer loop there is a velocity generator. This module decides how to propagate the input force generated by the patient to the velocity controller governing the main unit. The module operates in different ways according to the phase of operation of the haptic device, in particular: zero velocity at startup, free motion during spline selection, position follower during spline initialization, or constrained admittance tracker during the spline follow up.

Each mode of operation is subject to the proper operation of localization module, communication and accelerometer and a state machine surveys that each mode will be properly activated only if all the other information are. This state machine surveys the status of the battery, the status of the handle grasp, the average velocities, the sliding plane inclination, the location on the exercise sheet and the Kalman correction, to ensure that at each step the device is able to properly perform without risks for the user. Additionally, a specific saturation unit provides a smooth force shaping when one of the wheels is operated close to its limits.

The velocity generator operates to make the device appear as a pure spherical mass placed in a carved path, with the effect of being free to move along the path, while being constrained in the normal direction. This is done by computing at each time a virtual impedance which operates in the normal direction with respect to the constraint, and by computing at each step the “virtual acceleration” that needs to be compensated due to the spline curvature. Step-by-step the velocity generator computes the mass-equivalent centripetal force and uses it to compensate the desired velocity of the device.

The unit allows to modulate its operation in order to soften the constraint impedance, simulating a virtual viscosity and/or self generating a minimum kinetic energy to help the patient following the trajectory.

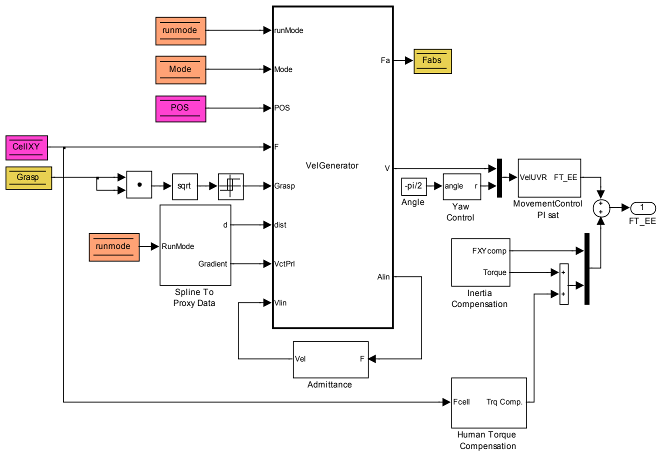

Fig. 6. Outer control loop. CellXY is the force measured by the load cell, Grasp is the information on handle grasping, POS is the Kalman estimate position, spline to proxy data provides the distance to the closest spline point and the relative gradient on the spline. Admittance generates a reference velocity vector to follow a given acceleration. Yaw control stabilizes the ϕ angle, movement control mixes stabilization with force tracking components, Inertia and human torque compensation take into account the forces and their application points generated by the relative components.

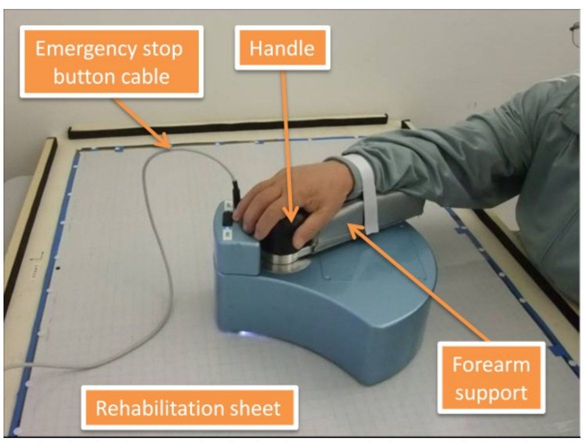

Fig. 7. The device in use. The device is presently in a validation phase in the center 'Auxilium Vitae' of Volterra.

Finally, when the spline control is not yet activated and, in order to prevent accidental falls due to spasm, a 'virtual walls' box prevents the device moving out of the workspace.

VII. Conclusions

The haptic interface presented in this work paves the way for new type of interfaces that combine the principles of mobile robotics with the requirements of haptic interaction. From the discussion above emerges the series of design decisions required to create a portable but effective solution.

This discussion about the interface and force feedback approach will be complemented by two main activities. The first is the detailed analysis of the sensor fusion under the effect of communication and sensing errors. The second is the evaluation of effectiveness and acceptability by subjects.

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