EMU: A Transparent 3D Robotic Manipulandum for Upper-Limb Rehabilitation

Justin Fong, Vincent Crocher, Ying Tan, Denny Oetomo and Iven Mareels

Abstract— **This paper introduces the EMU, a three-dimensional robotic manipulandum for rehabilitation of the upper extremity for patients with neurological injury. The device has been designed to be highly transparent, have a large workspace, and allow the use of the hand for interaction with real-world objects to provide additional contextual cues during exercises. The transparency is achieved through the use of a capstan transmission for the drive joints; a hybrid serial parallel kinematics minimising moving inertia; and lightweight materials. An experimental protocol is reported here which demonstrates the transparency through a comparison to out-of-robot movements, and with an existing rehabilitation robotic device. Additionally, an adjustable gravity compensation method is constructed, which minimises the torque required at the shoulder to carry the subject's arm. These characteristics allow the EMU to serve as a multi-purpose platform for the further development of novel robot assisted rehabilitation strategies.**

I. INTRODUCTION

Motor recovery from neurological injuries is driven by intensive therapy involving repetitions of goal-orientated movements. To assist this, a number of robotic devices designed for use in the rehabilitation of the upper extremity for neurologically impaired patients have been developed over the past 20 years [1]. Such devices mechanically interact with the patient whilst they attempt to perform motor actions, either assisting or challenging the patients in a structured way, aimed at accelerating and/or furthering their recovery.

The specific purpose and mode of interaction with the human user dictate a set of design criteria for an ideal upper limb rehabilitation robot, such as reviewed in [2]. Principal among the desired combination of characteristics are the transparency of the device, ease of setup for each patient, large workspace and sufficient static load. The trade off between transparency and the static load capability is also influenced by the inertial bandwidth of the mechanism. However, it is generally recognised that the required motions in rehabilitation exercises are of low to medium velocity, thus allowing the trade off to be made in an otherwise highly stringent (and expensive) set of design requirements.

Transparency is an important characteristic in the design of an upper limb rehabilitation robot. It allows forces exerted by the subject to affect the motion of the robot, thus, in a subject with some motor functionality of the arm, transparency allows a passive method of detection and interaction with the human. This allows the device to be used as an assessment device. Additionally, a rehabilitation robot sensitive to the forces exerted by the subject's arm is also able to regulate the safe amount of force applied in the exercises. Transparency can be achieved through an active force control approach (admittance) or through mechanical backdriveability and impedance control yielding to a solution which has a lower computational and economic cost.

Existing physically assistive devices are conventionally classified into two types — robotic manipulanda and exoskeletons. Manipulanda interact with the subject at only a single point (usually by a handle or a support piece strapped to the wrist or the forearm). They include devices such as the MIT Manus [3] and the MIME [4]. Exoskeletons are devices whose kinematics are designed to conform with that of the skeletal system of the upper limb, and thus should include a matching degree of freedom for each modeled physiological degree of freedom. Examples of exoskeletons include the ARMin [5], the ArmeoPower (Hocoma, Switzerland) and the ABLE platform [6].

Due to their single-point-of-contact design, existing manipulanda do not fully regulate the posture of the user's arm, which may lead to situations where for pathological synergies [7] are not accounted for in subjects' movements. Furthermore, the majority of existing manipulanda today are of a planar (two-dimensional) design, which does not allow non-planar movements during exercises — movements that occur often in activities of daily living.

Exoskeleton devices have been utilised to produce 3D (spatial) arm motion in rehabilitation. However, this comes at a cost to other aspects of the device. Most important is the difficulty in providing a good match between the kinematics of the robot and the human users. When the axes of movements of the robot do not perfectly align with that of the user, it creates mechanical constraints that hamper movement. Furthermore, due the person to person variation in arm and body shapes, a more complex set up is required for each patient as the lengths of the robotic links must be adjusted each time. Furthermore, due to the serial kinematics of the exoskeleton, conforming to the human upper limb, mechanical inertia introduced by the drive motors and various rigid linkages are commonly distributed along the serial arm, reducing the dynamic transparency of the robot. This is further magnified by the need for a sizeable joint torque, resulting in significant motor inertia involved in the moving parts of the robot. The typical solution is to introduce a high gear ratio to the motor, which compromises the backdriveability of the robot. Finally, due to their more complicated arrangement, exoskeletons are often of higher cost, compared with manipulanda.

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In this paper, a solution is introduced to provide assistance for the rehabilitation of the upper limb that overcomes disadvantages associated with the approaches discussed above. The capabilities are realised within a prototype robotic platform, dubbed EMU. The design follows the manipulandum approach with measures to incorporate some of the important advantages of exoskeletons. The design decisions were made with the specific applications of upper limb rehabilitation of patients with neurological motor impairment in mind. As such, it was designed to provide the advantages of (1) having a large workspace in 3D, (2) being easy to set up for each patient, and most importantly, (3) having high transparency.

The paradigm is similar to that of the HapticMaster [8] device used for the ADLER project [9]. The main conceptual difference lying in the system design — the HapticMaster has a Cartesian, serial, kinematic structure and uses a force sensor coupled with an admittance control to achieve transparency, whereas the EMU takes advantage of the backdrivability obtained from a semi-parallel mechanism.

The remainder of this paper includes an overview of the design of the EMU, followed by two investigations demonstrating the characteristics and potential of this robot. The first explores the transparency of the EMU through an experimental procedure utilising healthy participants. The second discusses a controller for gravity compensation of the upper-limb.

II. SPECIFICATIONS AND DESIGN

The design of EMU to address the aforementioned objectives is discussed in this section. This discussion is divided into two segments — the mechanical design, with the objective of covering a workspace appropriate for a large range of rehabilitation activities, but with dynamic properties which allow mechanical transparency; and the electrical and software design, structured to allow flexibility in controller and interface implementation whilst still maintaining safety and deterministic timing.

A. Mechanical Design

1) Kinematics: The EMU has 6 DOF end-effector movement, where only the first 3 DOF (associated with the translation of the end-effector) are actuated. The first axis is rotational about a vertical axis. The second and the third actuate a 4-bar linkage arrangement, which provides movement in a vertical plane positioned by the first axis — see Figures 1 and 2. This allows most of the motor inertia to be located at the base of the manipulandum, reducing the effective moving inertia of the robot.

An unactuated spherical wrist is constructed for the end-effector. The subject's wrist is connected to the system utilising a splint such that the center of the wrist corresponds to the robot end-effector point and center of rotation of the passive ball-joint, similar to the one proposed in [10]. The spherical joint and splint are designed such that the subject's hand is left free, allowing direct interaction with physical objects — motivated by studies also indicating the importance of context in effective rehabilitation exercises [11]. The rotations of the passive joints are measured, providing the patient's forearm pose (*i.e.* wrist position and forearm orientation). This unactuated spherical joint means that the general posture of the user arm is not physically regulated. This is not necessarily a disadvantage, as clinical practitioners encourage active and conscious participation (of the subject) in the correction of movement postures, and physical restraints can increase the risk of injury.

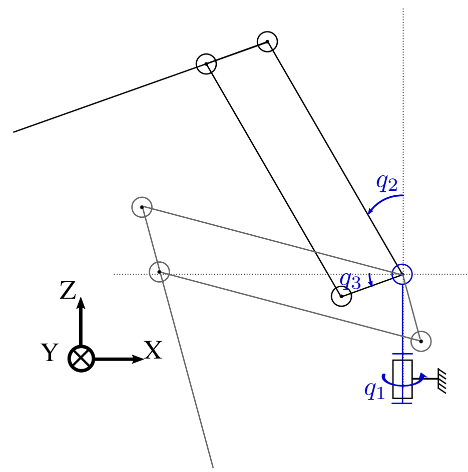

Fig. 1. EMU schematic displaying the kinematic structure and the three actuated axis and extreme configurations.

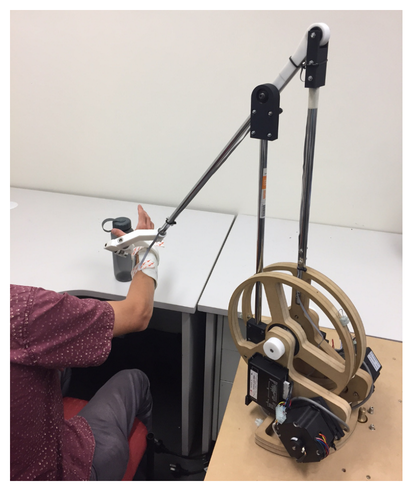

Fig. 2. EMU prototype with a subject.

The lengths of the links are designed to allow for access to a workspace volume of $0.8m \times 0.8m \times 1m$ covering a large portion of the human wrist workspace. Figure 3 shows the intersection of the robot workspace and the wrist workspace of a subject with limb lengths of $0.34m$ and $0.27m$ [12].

2) Actuation and Transmission: The EMU was designed to achieve a high torque capability while preserving backdrivability. This is achieved in a number of ways. First, all three actuated axes are driven through a capstan transmission, directly by a DC motor (without a gearhead). The capstan arrangement provides a 23:1 gear ratio through sizing of the capstan wheel and a bushing mounted on the motor shaft.

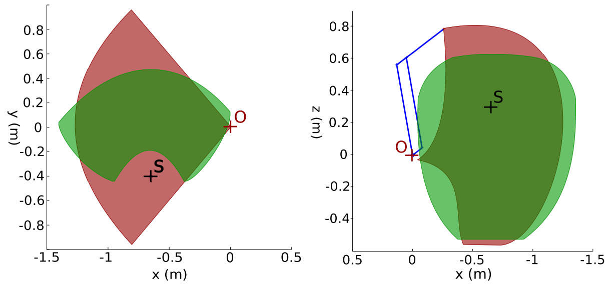

Fig. 3. Top (left) and Front (right) views of the system workspace (red) and human arm workspace (green) for a subject with limb lengths of $0.34m$ and $0.27m$. Points (O) and (S) respectively denote the robot origin and subject shoulder position. One extreme robot configuration is shown on the front view (blue).

The bushing is threaded on its external surface, thus allowing the capstan wire to sit in the groove of the thread. This has lower friction compared to geared or belt-driven options as there is no rubbing component in the motion. Secondly, the hybrid serial-parallel structure — and subsequent position of the motors — further reduces the inertia of the device and allows the use of high power (and heavy) motors. Finally, the moving arms of the EMU are constructed out of light-weight, hollow, aluminium tubes.

Each axis of the current prototype is powered by a 86BL71 brushless motor (Fulling Motor) with nominal torque of 0.7Nm and peak torque of 2.1Nm, driven by a Copley 503. Each capstan has a reduction ratio of $\frac{300}{13} = 23$ leading to a peak output torque of 48.5Nm for each joint. This implementation results in a system with an average end-effector force of 48N in the horizontal plane and 38N in the vertical plane in its usable workspace, which can be adjusted in future iterations through resizing of the motors or capstan arrangement but is sufficient to support the arm of a 80kg subject (see Section IV).

B. Electrical, Electronic and Software Design

The EMU utilises a CompactRIO (National Instruments, USA), which includes a microprocessor running Real Time (RT) Linux, and Input/Output channels connected through an FPGA. This controller is connected via ethernet to a host computer, which runs user interface software. Analogue Outputs (AO) are used to command the motor drives. Incremental encoders, fitted on each motor shaft, are connected via high speed Digital Inputs (DI). Potentiometers, providing absolute angular measurement of each of the 6 axes, are connected to Analogue Inputs (AI).

The software is designed in a hierarchical manner, with higher priority time-critical processes running on faster, deterministic hardware and deterministic software threads, and lower priority tasks running as non-RT software on the host computer. This arrangement can be seen in Figure 4. Specifically, the software limits (angular, velocity and torque limits), the open-loop (feedforward) gravity and friction compensations [13] and an impedance controller [14] run at 10kHz on the FPGA whereas higher level controllers (including path and trajectory planners) run at 1kHz on the

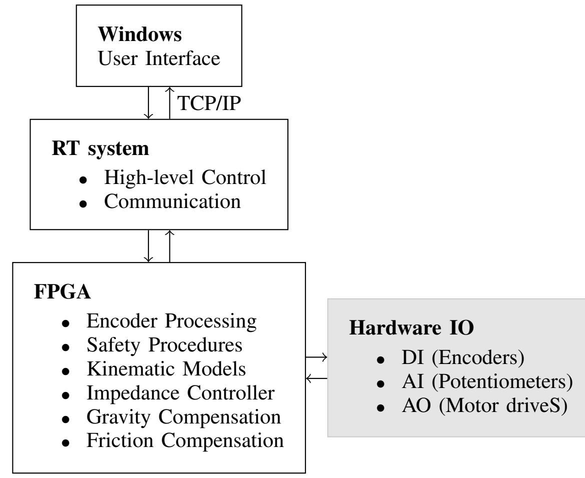

Fig. 4. The Software and Electronic Architecture of the EMU

RT controller. A Personal Computer running Windows OS (Microsoft, USA) is used as a host PC for the user interface. The software was written in LabVIEW.

III. TRANSPARENCY EVALUATION

The role of a robotic device in neurorehabilitation is to impart force onto the subject whilst they attempt to complete a movement, in order to encourage the use of certain movement or muscle activation patterns. As such, forces which are applied unintentionally may result in the promotion of unintended movement patterns within the subject. Therefore, it is critical that a robotic device design for rehabilitation is as mechanically transparent as possible.

In evaluating transparency, a traditional method is to use a force and torque sensor to measure the forces applied at the end-effector when a given motion is performed. In this case, the smaller the magnitude (of the force and torque), the better. Alternatively, within the context of rehabilitation of the upper limb, transparency can also be evaluated by having human subjects perform reaching actions while they are attached and not attached to the rehabilitation robot. The trajectories of the movements in these two conditions can then be compared. In an ideal case, the trajectories for the same intended motion would be identical — i.e. the robotic device does not affect the movements of the subjects. Previous similar studies on an existing rehabilitation device have highlighted how significantly the movement patterns may change [15], [16], [17]. In this paper, the latter approach is taken to evaluate the transparency of the EMU.

A. Experimental Methods

Five healthy subjects were involved in the experiment after providing an informed consent. A similar protocol to the one utilised in [16] by the authors was then used. Subjects were asked to reach to virtual targets in two conditions: in the robot and out of the robot. Magnetic sensors (3d Guidance trakSTAR, Ascension Corp) were attached to the subject's elbow and wrist. The position of the wrist was mapped to a virtual cursor, and subjects were asked to reach from a fixed starting position (in the sagittal plane in line with the shoulder, with elbow flexed to approximately 45 degrees) to one of six targets — directly forward, to the left and to the right, and the same with a vertical elevation. The subjects were asked to reach each target in one second.

Two conditions were tested — (1) the subject completely free to move, not in any way connected with the robot, with only the magnetic sensors attached — “Free”; and (2) the subject attached to EMU using the wrist splint — “Robot”, where the EMU was set to its transparent mode (compensation of its own weight and friction). Each subject reached to each target 10 times in both conditions. The order in which the conditions were presented was randomised between subjects.

The impact of the EMU on subject performance was measured using five metrics dependent on wrist position only, as described in [16]: (1) Peak Speed — The largest speed (as calculated in real-world coordinates, using a first-order Euler approximation on the position data); (2) Time of Peak Speed — The time of the peak speed relative to the start of the movement; (3) Smoothness — Spectral Arc Length (SAL) Smoothness as defined in [18]; (4) Curvature — Measured as the integral of the distance of the reaching trajectory from a straight line connecting the home position and the final position (at $t = 1$s); and (5) Accuracy — Defined as the shortest distance of the cursor to the target in virtual coordinates at $t = 1$s.

These metrics were chosen for their relevance to rehabilitation [19]. The metrics are evaluated in two ways. First, a Wilcoxon Signed Rank Test is used to compare the movements in the *Free* versus the *Robot* conditions. Secondly, a comparison between the data presented here for the EMU, and those for using the ArmeoPower (Hocoma, Switzerland), as presented in a previous work [16].

B. Experimental Results

Figure 5 illustrates the change in metrics with respect to the two reaching conditions. It is noted that performing the actions within the robot does affect a significant difference in the movement patterns illustrated by these metrics.

Figure 6 illustrates the percentage change from *Robot* to *Free* for the ArmeoPower and the EMU. It can be seen that the EMU affects the metrics less in all metrics, with the exception of curvature, suggesting that the EMU provides a more mechanically transparent environment for rehabilitation.

Movements made within the EMU were found to be affected compared to those made outside it. However, these changes are relatively small, with Peak Speed, Time to Peak Speed, Smoothness and Curvature affected by less than 15%. Accuracy is affected more significantly, with a 50% decrease. However, it is noted that the absolute change is in the order of $3mm$ in magnitude. The limited effect on these metrics suggests that, although the subjects were aware of being attached to the EMU, its impact was minor. Despite this, it is important to note that these changes are not directly the result of the forces — the subjects are likely to have, in some way, accommodated for the interaction forces, and/or changed their movement patterns slightly due to the change in context. Regardless, these small effects indicate that the interaction forces are minimal — at the very least the subjects are capable of easily overcoming these forces to ‘correct’ for the changes.

Comparisons are also made against the commercially available rehabilitation (active) exoskeleton ArmeoPower. In this comparison, it can be seen that the changes in the metrics introduced by the EMU are two to four times lower than the ArmeoPower. There are a number of reasons for this. First, the ArmeoPower is a full exoskeleton, and thus is attached to the arm at multiple points. This provides additional locations at which force can be imparted on the subject, causing changes in the movement patterns. Secondly, the ArmeoPower’s serial structure naturally leads to a heavier system and thus more inertia which must be compensated for — particularly in the relatively fast movements considered here. The EMU has most of its mass located at its base, and as such less mass must move when the arm moves — again reducing the force applied to the subject’s arm.

The study thus showed that the movements are affected when using the EMU compared to those made in free reaching conditions, however, the design of the EMU leads to a significantly smaller effect than exoskeleton based rehabilitation robotic devices, (in this case, represented by the Armeo Power) allowing more refined interactions with the subjects and a greater capability to detect or react to movement contributions.

IV. GRAVITY COMPENSATION

A common and useful feature amongst rehabilitation robots is the ability to ‘de-weight’ the arm [20], such that the force threshold for movement is lower — that is, the muscles do not need to overcome the weight of the arm first, before the arm accelerates.

A. A 3D manipulandum specific problem

The construction of the robotic device has an obvious effect on how the gravity compensation must be achieved. For example, horizontal planar manipulanda do not require active gravity compensation — the structure of the device itself restricts movement in the vertical direction. On the other hand, exoskeletons do require active compensation. This compensation can be achieved by estimating the mass of each arm segment (upper arm, forearm, hand), and compensating for the associated gravitation force with torques at each robotic joint.

By design, a three dimensional manipulandum can only provide directional force at one point on the patient’s arm. As such, the approach taken for gravity compensation is to calculate and apply the force at this one point to cancel the torque required by the shoulder to counteract the weight of the arm.

Within this analysis, the arm is modeled as a fixed two-link mechanism, upper-arm and forearm, with respective lengths l_{ua} and l_{fa}. Each link is assumed to be a point-mass, centred

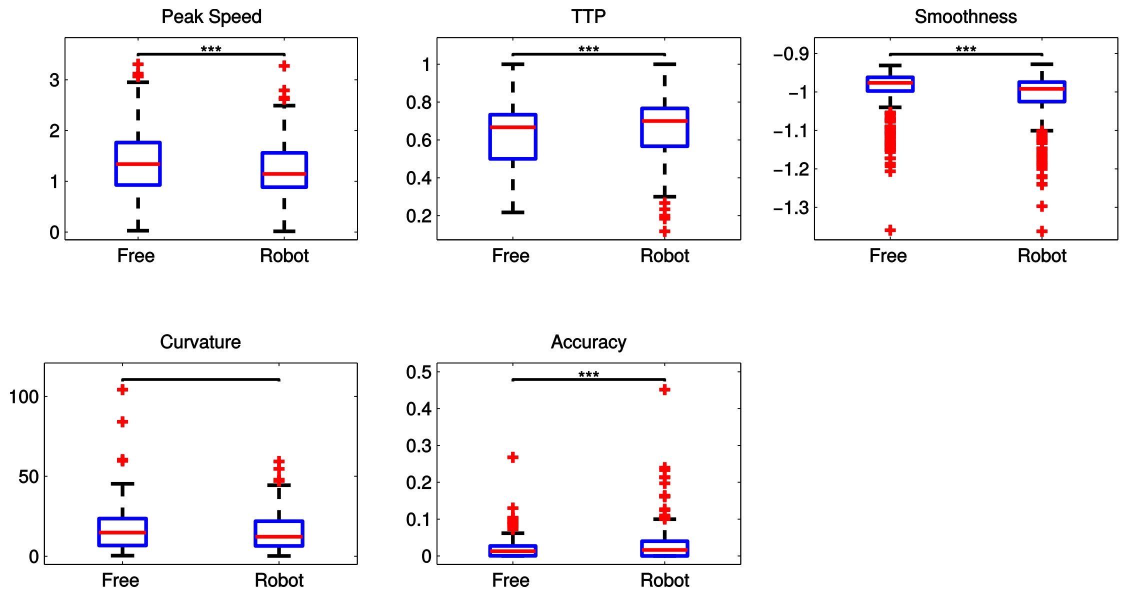

Fig. 5. Comparison of metrics between movements performed within EMU, and when the same movements are performed outside. *** indicates significant difference $p < 10^3$

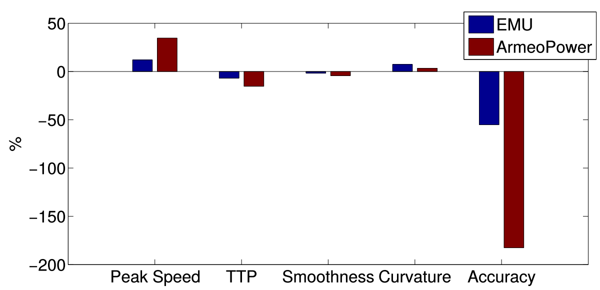

Fig. 6. Comparison of change in metrics (percentage) when performed within EMU, and when performed within the ArmeoPower

along the link at points U and F, respectively noted m_{ua} and m_{fa} (see Figure 7).

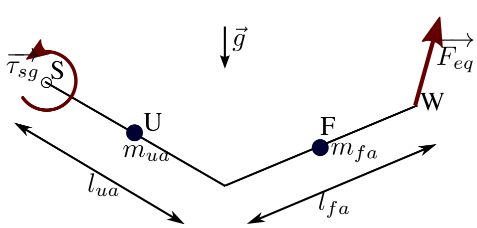

Fig. 7. Schematic representation of the human arm as a two link mechanism in the sagittal plane. W represents the location of the wrist, $\overrightarrow{\tau_{sg}}$ is the shoulder torque required to support the arm weight, and $\overrightarrow{F_{eq}}$ an 'equivalent' force applied by the robot.

The shoulder torque τ_{sg} required to support the weight can be expressed as:

$$\overrightarrow{\tau_{sg}} = m_{ua}\overrightarrow{SU} \times \overrightarrow{g} \; + \; m_{fa}\overrightarrow{SF} \times \overrightarrow{g} \tag{1}$$

It is noted that the required shoulder torque is variable and dependent on the arm posture. Thus, to compute the appropriate gravity-compensation force, the system needs to measure this posture and not only the forearm pose. This can be achieved in a number of ways using external sensors (such as IMUs, RGBD cameras or magnetic sensors such as those used in the experiments presented in Section III).

B. Proposed gravity compensation

In order for the 3D manipulandum to compensate for the shoulder torque τ_{sg}, the equivalent force $\overrightarrow{F_{eq}}$ which must be applied at the end-effector point (*i.e.* the wrist center W) must satisfy:

$$\overrightarrow{\tau_{sg}} = \overrightarrow{SW} \times \overrightarrow{F_{eq}} \tag{2}$$

The solution of the minimal norm is given by:

$$\begin{aligned} \overrightarrow{F_{eq}} &= \frac{\overrightarrow{\tau_{sg}} \times \overrightarrow{SW}}{\|\overrightarrow{SW}\|^2} \\ &= \frac{\left(m_{ua}\overrightarrow{SU} \times \overrightarrow{g} \; + \; m_{fa}\overrightarrow{SF} \times \overrightarrow{g}\right) \times \overrightarrow{SW}}{\|\overrightarrow{SW}\|^2} \end{aligned} \tag{3}$$

This theoretical analysis indicates that the magnitude and direction of the gravity compensation force $\overrightarrow{F_{eq}}$ is dependent on both the arm parameters (lengths and masses) and posture. Figure 8 (left) shows an example of how this equivalent force varies whilst the wrist is moving in the sagittal plane in line with the shoulder, for arm parameters ($l_{ua} = 0.34m$, $l_{fa} = 0.27m$ and $m_{ua} = m_{fa} = 2.2kg$ — corresponding to the arm mass of a 80kg average adult).

For these parameters, the required gravity compensation force ranges from 0N to 38N even in this restricted workspace, indicating the importance of taking into account the human arm posture when providing the gravity compensation. Figure 8 (right) provides an indication that the capability of the current prototype is sufficient to produce

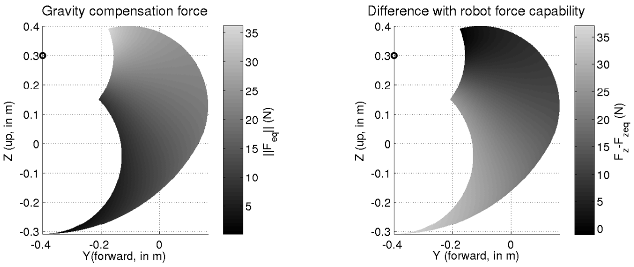

Fig. 8. Left: magnitude of gravity compensation force required of the robot at its end-effector when applied to point W in Figure 7. Right: difference between required vertical force and maximal vertical robot force at this point. Black circle represents the shoulder point.

this force. The proposed solution thus suggests one possible method of providing arm gravity compensation for 3D manipulanda given that the upper-limb posture is known. This method of gravity compensation is yet to be experimentally validated.

V. CONCLUSIONS AND FUTURE WORK

This paper introduces the EMU — an electromechanical manipulandum for upper limb rehabilitation. This work highlights the design and features of the device, including its mechanical transparency, large workspace, and gravity compensation. In particular, it is demonstrated that the dynamics of the EMU has a weak impact on the resulting movements made with the arm, and a large enough workspace to cover the active range of motion of healthy subjects. The EMU's capability in providing a useful force over a large 3D workspace while remaining transparent demonstrates that the proposed design yields an appropriate balance between classes existing upper-limb rehabilitation systems — exoskeletons and planar manipulanda.

Future work will continue on this platform through the development of upper limb rehabilitation specific control implementations. The motorised and dynamically transparent platform thus allows the practical realisation of various repetitive exercise motions investigated in the robot assisted rehabilitation literature, such as reviewed in [21], and the realisation of assistive strategies, such as [22] [23], in a spatial workspace.

Furthermore, the EMU has been intentionally designed to allow free movement of the hand. Whilst the majority of robotic devices for rehabilitation utilise a virtual environment, studies indicate the importance of context in effective rehabilitation exercises [11]. The use of virtual environments is useful for motivation (the exercises can be 'gamified'), an additional mapping between the real and virtual worlds is required, thus questions remain regarding generalisation of these exercises. Furthermore, traditional rehabilitation exercises are generally goal-orientated — for example, using a spoon to feed oneself. As such, the ability to have a free hand to work with physical object is an important part of the construction of the EMU, and thus the implementation of control strategies utilising this characteristic will feature in future developments.

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