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DOI
10.1109/ICORR55369.2022.9896556

Publication date
2022

Document Version
Final published version

Published in
Proceedings of the International Conference on Rehabilitation Robotics, ICORR 2022

2022 International Conference on Rehabilitation Robotics (ICORR)
25-29 July, 2022, Rotterdam, The Netherlands

Towards Unsupervised Rehabilitation: Development of a Portable Compliant Device for Sensorimotor Hand Rehabilitation

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2022 International Conference on Rehabilitation Robotics (ICORR) | 978-1-6654-8829-7/22/$31.00 ©2022 IEEE | DOI: 10.1109/ICORR55369.2022.9896556

Abstract— **Sensorimotor impairments of the hand after stroke can drastically reduce the ability to perform activities of daily living. Recently, there has been an increased interest in minimally supervised and unsupervised rehabilitation to increase therapy dosage and to complement conventional therapy. Several devices have been developed that are simple to use and portable. Yet, they do not incorporate diversified somatosensory feedback, which has been suggested to promote sensorimotor recovery. Here we present the prototype of a portable one-degree-of-freedom hand trainer based on a novel compliant shell mechanism. Our solution is safe, intuitive, and can be used for various hand sizes. Importantly, it also provides rich sensory feedback through haptic rendering. We complement our device with a rehabilitation game, where we leverage interactive tangible game elements with diverse haptic characteristics to provide somatosensory training and foster recovery.**

I. INTRODUCTION

Stroke is considered a major cause of disability worldwide with over 12 million new cases being globally reported each year [1]. Up to three out of four stroke survivors experience long-term hand impairments such as reduced strength, mobility, and dexterity. Moreover, approximately half of all stroke patients suffer from sensory loss in the hemiplegic upper limb and hand [2], [3]. This results in a reduced ability to independently perform activities of daily living (ADL) [4], [5].

The recovery of sensorimotor functions requires a long and demanding rehabilitation program [6] that includes highly intense [7] and repetitive training [8]. Additionally, it has been suggested that sensory training should also be an integral part of rehabilitation after stroke [9], [10]. Numerous robotic devices have been developed, mainly to address the recovery of motor functions (see [11], [12] for reviews). Yet, robotic devices could also be employed to promote the recovery of sensory functions. Specifically, haptic rendering – i.e., the simulation and physical representation of interaction forces with virtual tangible objects – could be an important source of somatosensory feedback that might enhance recovery [13], [14], [15].

*This work was supported by the Swiss National Science Foundation through the Grant PP00P2163800 and the Dutch Research Council (NWO) Talent Program VIDI TTW 2020

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Motivation has been associated with improved rehabilitation outcomes after stroke [16]. By leveraging elements of game design, e.g., scores, a high level of patient's motivation can be maintained during the training of repetitive movements [17]. Thus, gamification together with virtual reality (VR) is included in many rehabilitation programs [18].

The use of robotic devices together with motivating VR games in minimally supervised or unsupervised home-based training could help to further increase therapy dosage by providing continuous care [19]. Portable devices are an essential tool to accomplish this and could, consequently, also reduce the increasing pressure on healthcare [20], [21].

There is no consensus on the definition of portability in literature; here we consider that a portable device should be compact and lightweight to be carried by a moderately impaired stroke patient. Complex multifunctional and multi-degree-of-freedom hand rehabilitation devices are often associated with technical challenges, poor usability, and low clinical acceptance [22]. A crucial criterion for technology acceptance by therapists and patients is a short and easy setup [23], [24]. Therefore, devices for unsupervised rehabilitation should be safe, simple, have obvious functionalities, and be intuitive to utilize [19].

When it comes to portable devices, several hand exoskeletons and gloves have been recently developed that are capable of guiding or detecting movements of each finger (e.g., [25], [26], [27], [28]). However, the donning of exoskeletal devices and gloves often requires an advanced level of finger mobility and dexterity from the patient. Portable devices that consist of none or only few moving parts could serve as an easy-to-use alternative. For example, the Armeo®Senso [29] consists of a rigid sensorized hand module and wearable motion sensors that are strapped on the upper limb to provide patients with performance feedback through a graphical user interface. Based on a similar principle, the Pablo® [30] device additionally includes a force sensor to measure grasping forces. The FitMi [31] is composed of two devices that are equipped with motion sensors and vibrotactile actuators to provide performance feedback. Finally, in addition to a motion sensor, the GripAble [32] also features a passive spring element that allows small grasping movements. Although these devices are portable and could potentially be used in a minimally or unsupervised setting, they generally do not provide diversified somatosensory feedback from the interaction with tangible virtual objects, limiting their potential to train realistic sensorimotor exercises that resemble ADL. Dedicated haptic devices (e.g., [33], [34], [12]), however, are often non-portable.

Here we present our first steps towards the development of a novel portable device for unsupervised sensory hand rehabilitation. We introduce a prototype based on an actuated compliant shell that interacts with the entire palmar side of the hand. Despite its mechanically simple hand-device interaction, our solution provides somatosensory feedback through haptic rendering and supports the training of grasping and releasing objects. We complement our robotic device with a rehabilitation game specifically designed to leverage somatic feedback and motivation.

II. METHODS

A. Requirements

1) Mechanical Design: To maximize **usability**, we opted for a design that is simple and intuitive to use. In order to achieve the desired **portability**, we required our device to be as compact and lightweight as possible. Furthermore, we considered **safety** as one of the most important design requirement for a device to be used in unsupervised rehabilitation. Consequently, the risk of skin getting pinched in gaps between moving parts should be minimized at all costs. Limiting the number of gaps could also help designing a device that is easy to clean and disinfect. On top of that, the device should be **ergonomic**, the patient's hand should not be left in an uncomfortable position (e.g., full finger extension) for long times and patients should be able to independently release the hand in case of power loss.

We aimed for a device that mimics a large-diameter **power grasp**, i.e., simultaneous flexion or extension of all fingers with abducted thumb. This is one of the grasps most frequently used in ADL [35] and is effectively trained in clinics [36]. Based on our experience with a previously developed hand rehabilitation device [12], we targeted a continuous grasping force of approximately 15 N. Besides, a design that allows to perform a power grasp, regardless of the patient's hand size and without adjustments or swapping of parts, would likely reduce the setup time.

2) Rehabilitation Game and Haptic Rendering: In an online survey we conducted with 33 participants working in neurorehabilitation, we found that they give high priority to the use of games that allow for **adjustable task difficulty** and the quantity of game objects (see Fig. 1). Furthermore, it has been suggested that rehabilitation games should be designed to be believable while not necessarily realistic [37].

During the game design, we aimed at rendering interactions with virtual objects that have **rich dynamics** – e.g., deformable objects or viscous fluids. The interaction with such objects has been reported to result in a stronger activation of sensory-related brain areas when compared to objects with simple dynamics [38], and thus have the potential to further improve the rehabilitation outcomes.

To keep the overall complexity of our solution low, we decided to realize the haptic rendering using open-loop impedance control. In this control scheme, the interaction force is computed as a function of the user's fingers displacement without measuring the true interaction force between the user and the device. Thus, no additional force sensors

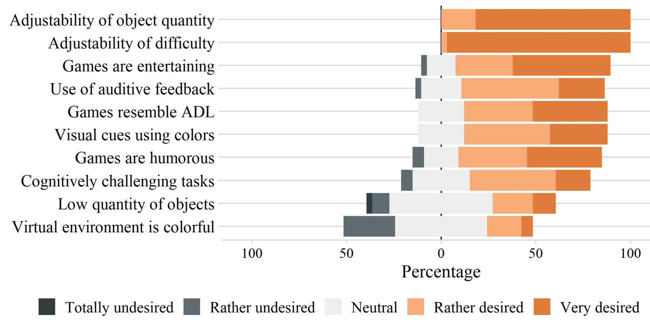

Fig. 1. Results of the online survey with 33 participants working in neurorehabilitation [24]. Therapists were asked to rate how desired the items listed above would be in a rehabilitation game that would be used in conjunction with a robotic device.

are needed [39]. This does, however, require a backdrivable and mechanically transparent transmission.

B. Mechanical Design

1) Compliant Shell: We propose a handle consisting of a single piece to satisfy our requirements. Specifically, we present a compliant shell that replicates the curvature of the fingers and the thumb during power grasping as shown in Fig. 2. The shell is composed of both flexible and locally reinforced rigid sections in the area between the metacarpophalangeal joints of the fingers and the thumb to provide a firm contact area for the palm. The shell endpoints in the area around the fingertips (Fig. 2) are also reinforced and act as the mechanical interface between the shell and the transmission underneath (see Fig. 3). The shell geometry, and thus the bending behaviour during grasping, was determined iteratively by means of rapid prototyping and comfort testing with seven healthy persons (hand lengths: 18.5±1.25 cm).

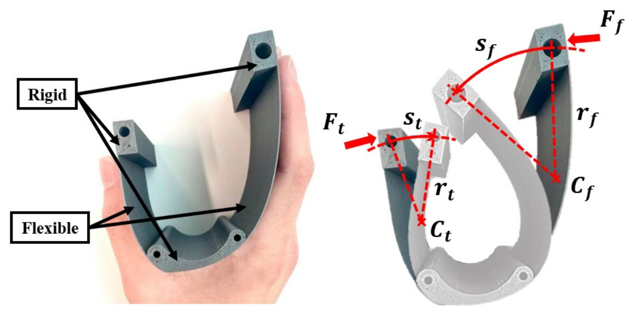

Fig. 2. Top view of the shell. Left: The flexible shell is made out of a single part. Local reinforcements are located at the shell endpoints and palm area. Right: Bending of the shell when forces F_t and F_f are applied.

2) Actuation and Transmission: To achieve a compact design and reduce the overall weight, we decided to drive our solution with only one DC motor and couple the motion of the thumb and the fingers. We aimed to achieve this coupling and the shell actuation with a transmission consisting of a synchronous belt and gears as shown in Fig. 3. A relatively low transmission ratio was desired, as this has been shown to be beneficial for mechanical transparency [40].

The fingertips and the thumb tip can be considered to be approximately coincident with the shell endpoints. They

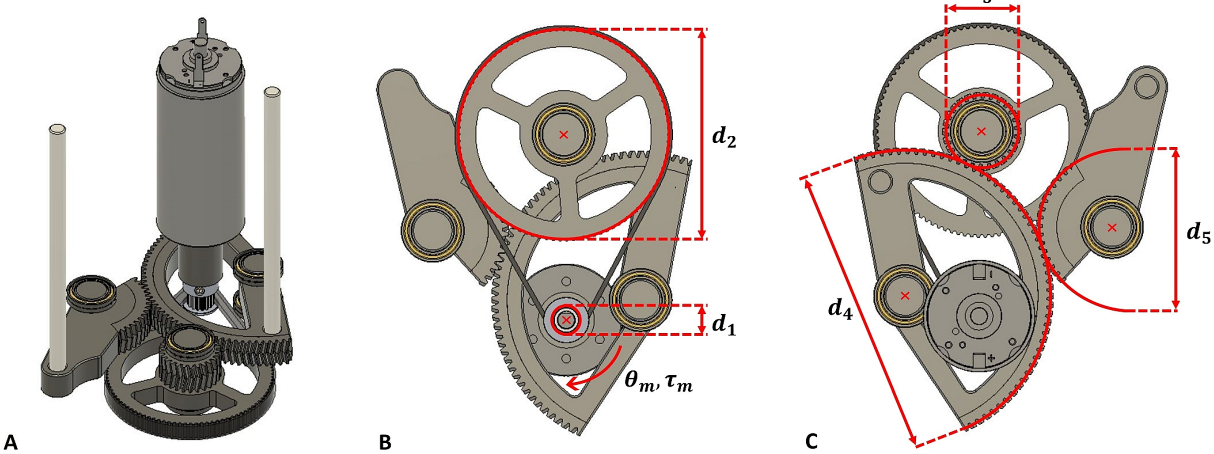

Fig. 3. **A**: 3D view of the transmission showing the three transmission stages and aluminium rods that are attached to the shell endpoints. **B**: Bottom view of the transmission. First transmission stage consists of a belt connection starting from the DC motor. **C**: Top view of the transmission. The second and third transmission stages are made up of herringbone gears.

move along the arcs with radii r_t, r_f and center points C_t, C_f as shown in Fig. 2. These parameters were defined in the iterative design process of the shell. We define the thumb and fingertip displacements s_t and s_f as arc lengths, with initial values $s_t = 0\,\mathrm{mm}$ and $s_f = 0\,\mathrm{mm}$ when the shell is in maximum extension (see Fig. 2). Given the motor shaft angle θ_m and the effective gear diameters d_1 to d_5 (Fig. 3), the displacement of the fingertips can be described as follows:

$$s_t = \frac{r_t}{i_t}\theta_m \quad \text{with} \quad i_t = \frac{d_2 d_4}{d_1 d_3} \tag{1a}$$

$$s_f = \frac{r_f}{i_f}\theta_m \quad \text{with} \quad i_f = \frac{d_2 d_5}{d_1 d_3} \tag{1b}$$

The motor torque τ_m depends simultaneously on the thumb force F_t and the finger force F_f, and is given by:

$$\tau_m = \frac{\partial s_t}{\partial \theta_m}F_t + \frac{\partial s_f}{\partial \theta_m}F_f = \frac{r_t}{i_t}F_t + \frac{r_f}{i_f}F_f \tag{2}$$

If we assume that an equal amount of grasping force F is applied at the thumb and the fingertips, i.e., $F_t = F_f = F$, we obtain:

$$\tau_m = r'F \quad \text{with} \quad r' = \frac{r_t}{i_t} + \frac{r_f}{i_f} \tag{3}$$

This relation between τ_m and F will be used to compute the required motor torque for simulated hand-object interaction forces.

C. Rehabilitation Game and Haptic Rendering

Since our goal is a functional and believable rehabilitation game with rich haptic rendering, we designed a three-dimensional game using Unity3D (Unity Technologies, USA). It is played on a computer monitor (but could eventually also be played with a smartphone or tablet) and controlled using our device and a keyboard. Special attention was paid to the creation of dynamic, tangible game elements with a wide range of (adjustable) haptic characteristics.

The interaction with those game elements was simulated by implementing a virtual wall, which is common practice for haptic devices [41]. To consider the coupled movement of thumb and fingertips, we first defined their combined displacement s and the speed $\dot{s}$ thereof as:

$$s = s_t + s_f = r'\theta_m \tag{4a}$$

$$\dot{s} = \dot{s}_t + \dot{s}_f = r'\dot{\theta}_m \tag{4b}$$

The rendered force F_r is then computed in the rehabilitation game according to Eq. (5):

$$F_r = \begin{cases} K(s - s_0) + B\dot{s}, & \text{for } s > s_0 \text{ and } \dot{s} > 0 \\ K(s - s_0), & \text{for } s > s_0 \text{ and } \dot{s} \leq 0 \\ 0, & \text{otherwise} \end{cases} \tag{5}$$

The values of the parameters K and B represent a virtual spring and a virtual viscous damping, and can be adjusted to render interactions with virtual objects with different dynamic characteristics. The parameter s_0 indicates the combined finger displacement s right in the moment a tangible object is touched. By setting $F = -F_r$ in Eq. (3), the desired motor torque to render the haptic interaction can be computed.

III. RESULTS

A. Design and Actuation of the Device

The final handle geometry – found through repeated comfort testing – incorporates a shell with a thickness of 0.6 mm and height of 100 mm following the curvature of the palmar side of the hand when performing a power grasping motion (Fig. 2). Hook and loop fasteners are used to secure the fingers to the handle and to ensure that the device also follows the users' self-initiated finger extension movements. A thumb strap did not appear to be required during the comfort testing due to the coupled thumb and finger motion. It was therefore omitted in favor of easier setup but could easily be added if desired/needed by patients. The shell endpoints are attached to the transmission mechanism by aluminium rods (Fig. 3A). The transmission was designed in a space-efficient arrangement and placed into an enclosure underneath the handle as shown in Fig. 4. Aside from the components being inaccessible to the user and thus preventing any safety issues such as pinching, this enclosure also serves as a base on which the hand can rest while operating the device.

The transmission consists of three stages: In a first stage, a synchronous belt (2GT profile) leads to a reduction ratio of 6:1 ($d_2 : d_1$; Fig. 3B). The second and third stage consist of herringbone gears that couple the motion of the thumb and the fingers with transmission ratios of 3:1 ($d_4 : d_3$; Fig. 3C) and 1:2 ($d_5 : d_3$), which leads to overall transmission ratios of $i_t = 18$ and $i_f = 12$ from Eq. (1). Both the shell and the components of the transmission were manufactured by means of 3D-printing using poly-lactic acid (PLA).

A DC motor with integrated optical encoder (3272CR and IER3-4096, Faulhaber, Germany) with a continuous torque of 75 mNm was found to satisfy our torque requirements. Given the arc radii of $r_t = 34\,\text{mm}$ and $r_f = 41\,\text{mm}$, it led to a maximum continuous force of $F = 14.1\,\text{N}$ from Eq. (3). The motor was placed inside the handle resulting in a compact design that does not interfere with the shell bending motion nor the patients' hand. Grooves in the base plate mechanically limit the displacement of the shell endpoints to $s_t = 20\,\text{mm}$ and $s_f = 36\,\text{mm}$ for the thumb and the fingers respectively, corresponding to an anatomical thumb and finger range of motion (RoM) of approximately [20, 53]° and [35, 85]°.

The device is controlled by an ESP32 microcontroller (Espressif Systems, China) in combination with an Escon Module 50/5 motor controller (Maxon, Switzerland) and communicates with the host PC via USB. The dimensions of the device are 122 mm x 147 mm x 132 mm, with a weight of 700 g (without electronics).

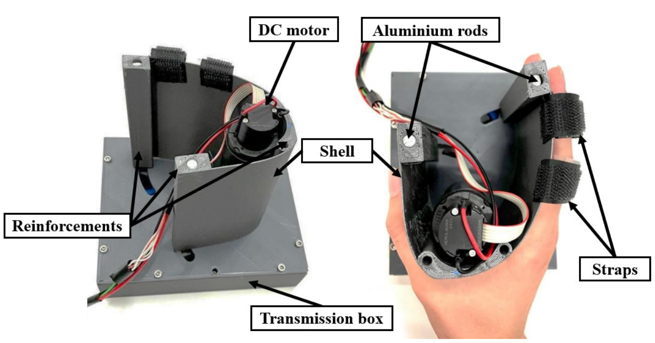

Fig. 4. Final design of the portable robotic rehabilitation device for grasp training. The 3D printed flexible shell is actuated at the fingertips with a single DC motor. Straps are used to attach the fingers to the shell. The DC motor is placed inside the handle resulting in a compact device. The entire structure is placed on top of a rectangular transmission housing.

B. Rehabilitation Game and Haptic Rendering

We designed a VR game scene that takes place in a virtual bar with glasses, liquid dispensers, and a hand avatar. The goal of the task is to fill glasses that appear on the screen as fast as possible by correctly squeezing the appropriate dispenser, i.e., with a determined fingertip force. Different dispensers contain liquids of different viscosities. The number and position of the dispensers are adjustable, while only one glass at a time is visible to help patients direct their attention more easily to the dispenser to be employed. Such a VR game incorporates all the elements listed in the requirements and methods section: low complexity, believability, and adaptability of virtual elements with different rich haptic characteristics.

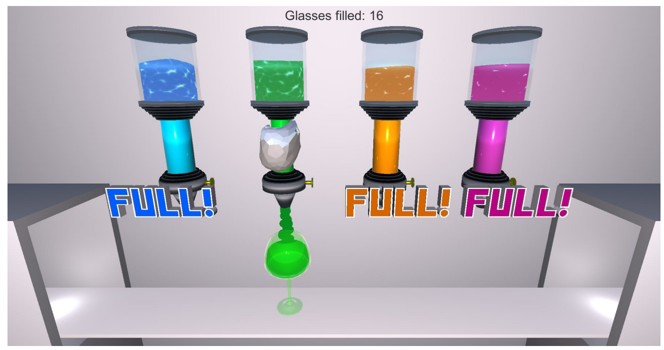

Fig. 5. Screenshot of the game showing four dispensers, glasses, and hand avatar. Whenever a glass is filled, a new one will spawn underneath a different dispenser. The text "FULL!" appears to signify that a glass has been filled. Each dispensers have different squeezing behaviours that are haptically rendered by the device.

Squeezing the handle causes a grasp to be initiated in the rehabilitation game. When engaging in a grasping movement, the hand avatar first automatically moves forward and touches the dispenser. Once the hand avatar encloses the dispenser (at a displacement of $s = s_0$), the hand-object interaction force is simulated by the virtual wall using Eq. 5, i.e., the user feels resistance as if a physical dispenser was grasped and squeezed. Thus, visual and haptic information is perceived when an interaction with a dispenser takes place. In order to promote the use of rich and diverse somatosensory information, each liquid and consequently each dispenser possesses different simulated properties – i.e., combinations of different values of K and B (up to $K = 2\,\text{N/mm}$ and $B = 0.1\,\text{Ns/mm}$). Therefore, each dispenser has to be squeezed differently to fill the glass as fast as possible, i.e., different fingertip displacement and interaction forces are needed. If squeezed too little, only a small amount of fluid is released, resulting in a slow filling speed. However, when squeezed too hard, the dispenser starts sputtering and spilling the fluid outside of the glass (Fig. 6). Whenever a glass is filled, the text "FULL!" appears and the next glass spawns. To move to a different dispenser, the hand has to be opened, i.e., fingers extended. Navigating to a different dispenser is then performed with the key arrows on a keyboard.

The computation of the virtual wall is executed in a 1 kHz loop on the microcontroller. Numerical differentiation with consecutive first-order low-pass filtering is utilized to compute the motor shaft speed. Via a USB connection, the finger displacement s, computed by Eq. (4), is sent to the host

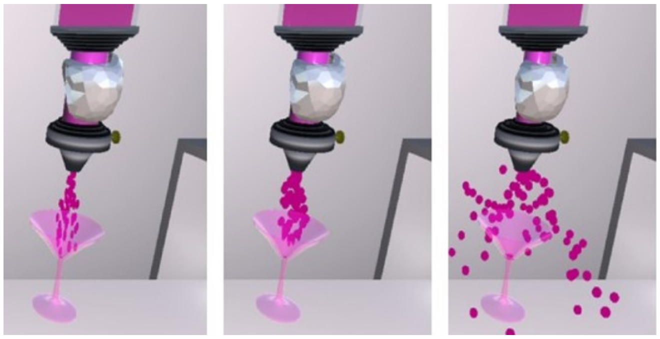

Fig. 6. Visual cues. Squeezing too little causes only a limited amount of droplets to fill the glass (left), while squeezing too hard makes the fluid sputter, which does not contribute to filling the glass (right). Applying the optimal amount of force causes the most droplets to enter the glass (center).

PC where the rehabilitation game is played. The parameters K and B are then computed and sent to the microcontroller.

IV. DISCUSSION

We presented the prototype of a portable one-degree-of-freedom device for unsupervised hand rehabilitation after stroke based on a novel compliant shell design. The device provides haptic rendering during training and can be used to train hand opening and closing.

Our distinctive shell design that consists of only one piece can be used safely regardless of patients' hand size. Its inherent safety stems from the absence of pinching risks, its backdrivable transmission, limited RoM, and limited force of $14.1\,\mathrm{N}$. This potentially enables its application in an unsupervised setting. By placing the motor inside the handle and engineering a space-efficient transmission, we obtained a compact and portable device, which could be beneficial for training at patients' home [19], [26]. Importantly, our device introduces the novelty of providing haptic rendering to portable devices. Our solution has the potential to provide rich somatosensory input through the simulated interaction with a wide variety of virtual tangible objects, instead of only tactile vibration, as is often the case in portable devices (e.g., FitMi [31], Gripable [32]).

To complement our device, we developed a rehabilitation game with a believable scenario and an intuitive task. It is designed such that it encourages patients to haptically explore the behaviour of the different dispensers and use somatosensory feedback to perform the task. Similar to other, more complex systems, (e.g., [33], [34]), our device does not require a real-time communication between the microcontroller and the host PC. An increased communication latency with the host PC could only lead to a delayed update of the parameters K and B. This could potentially negatively impact the training experience, but it can not cause instability of the haptic controller [41], which would be safety-critical.

Our current prototype also possesses some drawbacks. The flexible handle of our solution acts as a spring ($\approx 0.03\,\mathrm{N/mm}$). The shell thus tends to return to its initial, unloaded shape. However, the maximum inherent restitution force exerted by the shell is rather low (i.e., $\approx 1.5\,\mathrm{N}$), which limits the risk of causing any physical harm. Still, it might confound the haptic rendering since the aforementioned force is currently not compensated in our control algorithm. Although we utilized a low transmission ratio that minimizes reflected motor inertia and motor friction, the mechanical transparency of our transmission could be further improved. This could, for example, be achieved by further reducing friction by replacing the 3D-printed gears with a low-friction solution such as a cable transmission [12], [40]. Finally, an inherent limitation of the shell design is that the length of the shell (i.e., exterior circumference from endpoint to endpoint) is constant throughout the entire RoM. This could pose a problem for designs where a larger RoM is desired, because the palmar circumference of the hand (i.e., arc length of the palmar side of thumb and fingers) shortens during flexion, resulting in sliding between the hand and the shell. However, due to the chosen RoM, this is not noticeable in our prototype.

We plan to further improve our device before performing tests with patients. First, the electronics need to be fully integrated and enclosed in the device. Second, we will realize a wireless setup. This will include the addition of a battery to power the device and the inclusion of a wireless transceiver to connect the device to a smartphone or tablet. Third, we would like to add an orientation sensor (i.e., inertial measurement unit, IMU), which – in combination with a rounded bottom surface of the device – would allow to train pronosupination movements. This could be used to navigate between different dispensers instead of the currently required keyboard. Resulting in reduced complexity, this could be advantageous for unsupervised training. Ultimately, the long-term durability and fatigue behaviour of our flexible shell need to be investigated.

In future studies, we will evaluate the usability and effectiveness of our device with stroke patients. In particular, we are interested in the usability of our system and the feasibility of minimally supervised or unsupervised hand training. On top of that, the enjoyment and level of motivation provided by the rehabilitation game will be assessed.

V. CONCLUSIONS

With its unique and intuitive shell design, our device holds the potential to be utilized in minimally supervised therapy or in an unsupervised setting at home. The final design will be fully portable and – in contrast to existing devices – feature haptic rendering to not only train motor functions, but to allow for simultaneous sensory and motor training.

ACKNOWLEDGMENT

The authors would like to thank the Department of Neurology, University Hospital (Inselspital) Bern, Switzerland for the precious collaboration. In particular, we are grateful for the numerous fruitful discussions with René Müri and Anna-Liisa Saare.

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