2018 7th IEEE International Conference on Biomedical
Robotics and Biomechatronics (Biorob)
Enschede, The Netherlands, August 26-29, 2018

The RETRAINER Light-Weight Arm Exoskeleton: Effect of Adjustable Gravity Compensation on Muscle Activations and Forces*

Markus Puchinger, Nithin Babu Rajendra Kurup, Thomas Keck, Johannes Zajc,
Michael Friedrich Russold and Margit Gföhler

Abstract— **the recovery of voluntary arm movements is one of the most important goals during stroke rehabilitation in order to avoid long-term disability in activities of daily living. Support against gravity in order to reduce flexion synergy is reported as an effective strategy to enable upper extremity rehabilitation. In this study, the reduction of muscle activities and muscle forces with the gravity compensated RETRAINER upper limb exoskeleton were analyzed carrying out defined movements with healthy subjects. The EMG signals of the main active muscles as well as the kinematics of different defined motions were captured and compared. The joint kinematics and the joint moments were computationally determined using a 3D musculoskeletal model. The effectiveness of the upper limb gravity compensation could be shown in both mean values of EMG signals and resulting muscle forces, indicating that this compact and lightweight arm exoskeleton can serve as a powerful tool to support the rehabilitation process.**

I. Introduction

According to the World Health Organization, annually an estimated 5 million people are stroke survivors and a high percentage of these are permanently disabled. Stroke often causes permanent and complex long-term disability in adults, placing a burden on families, health professionals and communities in general [1].

In literature, upper limb hemiparesis is widely reported as one of the primary impairments after stroke. While many patients recover ambulatory function after dense hemiplegia, the restoration of arm motor skills is often incomplete and more than 60% of patients are not able to use their paretic hand in functional activities [2]. The recovery of voluntary arm movements is one of the most important goals during stroke rehabilitation in order to avoid long-term disability that may restrict activities of daily living (ADL), social and occupational activities and also leads to depression. Raising the upper-limb against gravity is documented as one of the hardest tasks for neurological patients as abnormal concurrent muscle activations result in synergistic joint torque coupling of shoulder abduction with external rotation, extension and often with elbow flexion [3], [4]. Specially adapted devices such as exoskeletons can enhance the rehabilitation process by assisting the patient in executing motor tasks [5] and it was shown that weight-support in upper limb exoskeletons benefit motor function recovery [6] especially if the load is varied during the therapy [7]. Over the last years, numerous upper limb exoskeletons for stroke survivors were developed but only few of them are used in clinical routine [8]. One of the few commercially available passive upper limb exoskeletons is the Armeo® Spring by Hocoma, based on the principle of the T-WREX exoskeleton, using a parallelogram linkage with elastic bands for arm weight compensation [9], [10]. Due to its dimensions and weight, the Armeo® Spring can only be used as a stationary device and cannot be installed on a wheelchair or chair. Additional to weight support, recent neurophysiological studies [11], [12] also advocated the use of neuromuscular electrical stimulation (NMES) simultaneously to generate voluntary muscle force in order to enhance beneficial effects of the motor re-learning process.

The RETRAINER arm exoskeleton is a modular, lightweight and non-cumbersome passive device based on the MUNDUS exoskeleton [13], [14]. Its aim is to facilitate recovery of arm motor function after stroke by providing a spring-based weight relief at the shoulder and elbow joints in order to reduce the muscular effort required to perform defined arm movements. In contrary to existing systems, the specific layout of the RETRAINER exoskeleton gravity support mechanism gives the possibility to adapt the progress and the level of the compensation torque to the therapeutic needs of the individual user. While using the RETRAINER exoskeleton, residual muscle function is utilized for actuation as far as possible and if additionally needed, NMES is used to generate additional muscle force.

For the design of effective training movements for stroke patients, it is essential to know how much individual muscle forces are reduced due to the gravity support. The aim of this study is to verify the functionality of the weight relief of the RETRAINER exoskeleton and to determine the decrease of required muscle force by comparing muscle activations and muscle forces with and without enabled gravity compensation during defined single degree-of-freedom (DoF) movements.

II. Methods

A. Exoskeleton

The RETRAINER upper limb exoskeleton (Fig. 5) is a modular, passive and lightweight device with 5 DoFs. Humeral rotation and wrist pro/supination are either controlled by residual muscle forces or locked at customized

*Research supported by European Union's Horizon 2020 research and innovation program under grant agreement No 644721.

Markus Puchinger, Nithin Babu Rajendra Kurup, Thomas Keck and Margit Gföhler are with TU Wien, Vienna, Austria.

(Markus Puchinger is the corresponding author; phone: +43 (1) 58801 30769; e-mail: markus.puchinger@tuwien.ac.at).

Johannes Zajc and Michael Friedrich Russold are with Ottobock Health Products GmbH, Vienna, Austria.

978-1-5386-8182-4/18/$31.00 ©2018 IEEE

positions. Shoulder elevation in sagittal and frontal plane as well as elbow flexion are actuated by residual muscle forces and additionally by NMES if needed and can be locked at any chosen position during the movement [15]. Gravity is not completely compensated so that the arm is slowly moved down by gravity without additional actuation. A universal mounting device allows an installation on both wheelchairs and normal chairs with different backrest shapes. In order to make the user feel less constrained, an inclination module allows to move the trunk when seated on a (wheel)chair wearing the exoskeleton. In combination with the shoulder joint unit the inclination module ensures unconstrained, physiological movements of the user's shoulder.

The spring-based gravity compensation mechanisms integrated into the shoulder and elbow joint modules, as shown in Figs. 1 and 2 respectively, provide compensation torques as functions of the respective joint angles, in order to compensate the torques due to gravity and to minimize muscular effort during arm movements.

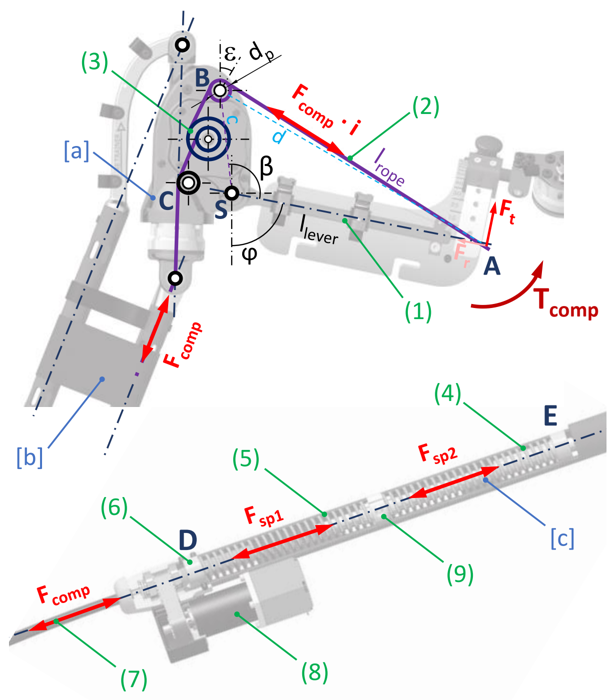

Figure 1. Schematic of the gravity compensation mechanism at the shoulder joint S.
([a] shoulder joint unit, [b] inclination mechanism, [c] external spring unit, (1) lever, (2) dyneema rope, (3) pulley assembly with gear ratio 2:1, (4) spring 1, (5) spring 2, (6) driven thread for automatic fine tuning, (7) bowden cable, (8) timing belt-gearbox-stepper motor combination, (9) lockable middle spring carrier)

Main element of the shoulder joint gravity compensation is a special alignment of a rope pulley B (diameter d_p) which guides a dyneema rope fixed to the end of the lever (length l_{lever}) in A. If the length AB (l_{rope}), which is a function of shoulder elevation angle φ, changes, consequently also the compensation torque T_{comp} is changed. The length l_{rope} is calculated as shown in (1). For a slim design of the module, the rope is guided through the inclination mechanism and then via a bowden cable to the spring unit mounted on the backrest. In the spring unit, the rope is attached to two springs connected in series. The length change of the distance AB causes a change of spring compression. To reduce the overall building size of the assembly, a twin pulley with a gear ratio of $i=2$ is integrated into the compensation mechanism between B and C. Based on the length change of l_{rope} (2), the spring force F_{comp} is calculated using (3). The resulting torque, generated by the tangential component F_t of the spring force, at the shoulder joint is given by (4).

The gravity compensation is designed for users with anthropometric measures between 5th female and 95th male percentiles. To adjust the gravity compensation for tall users, it is necessary to change from two-spring to one-spring mode. This is manually done by tightening a hexagon socket screw on the middle spring carrier (Fig. 1). Changing to one-spring mode results in a higher total spring stiffness k and consequently in a higher gravity compensation torque at the shoulder joint.

$$l_{rope}(\varphi) = \sqrt{d(\varphi)^2 + \left(\frac{d_P}{2}\right)^2} + \frac{d_P}{2}\varepsilon(\varphi) \tag{1}$$

$$\Delta l_{rope}(\varphi) = l_{rope}(\varphi) - l_{rope}\left(\frac{\pi}{2}\right) \tag{2}$$

$$F_{comp}(\varphi) \cdot i = k \cdot \Delta l_{rope}(\varphi) \tag{3}$$

$$T_{comp}(\varphi) = F_t\left(\varphi, F_{comp}(\varphi)\right) \cdot i \cdot l_{lever} \tag{4}$$

For fine tuning of the gravity support, a stepper motor combined with a timing belt and a driven thread changes the pre-load of the spring package (Fig. 1). The compensation level is electronically adjusted to the individual user by the operator at the beginning of each training session. In Fig. 3 the compensation torque gradient is shown as a function of the shoulder elevation angle.

The weight relief for the forearm is realized by a spring linked with a cable pull, which is manually adjusted for each individual user at the beginning of the training session (Fig. 2).

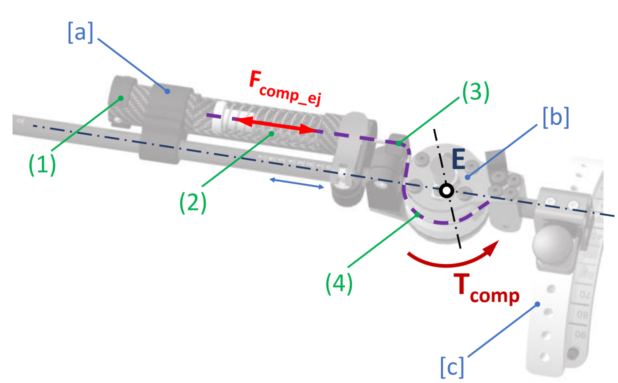

Figure 2. Schematic of the gravity compensation at the elbow joint *E*
([a] elbow joint compensation unit, [b] elbow joint unit, [c] humeral-rotation unit, (1) spring fixation clip,(2) spring, (3) rope pulley, (4) dyneema rope)

The level of the forearm weight compensation can be adjusted by shifting the elbow joint compensation unit to the desired position. The compensation torque at the elbow is independent from the spatial upper arm position. For special cases of humeral rotation settings above 40° with shoulder elevation angles higher than 50° the use of a forearm gravity compensation is not needed, because the flexion/extension of the elbow is performed in a more or less horizontal plane where gravity does not have a large impact. In that case, the compensation can be disabled by inserting a clip.

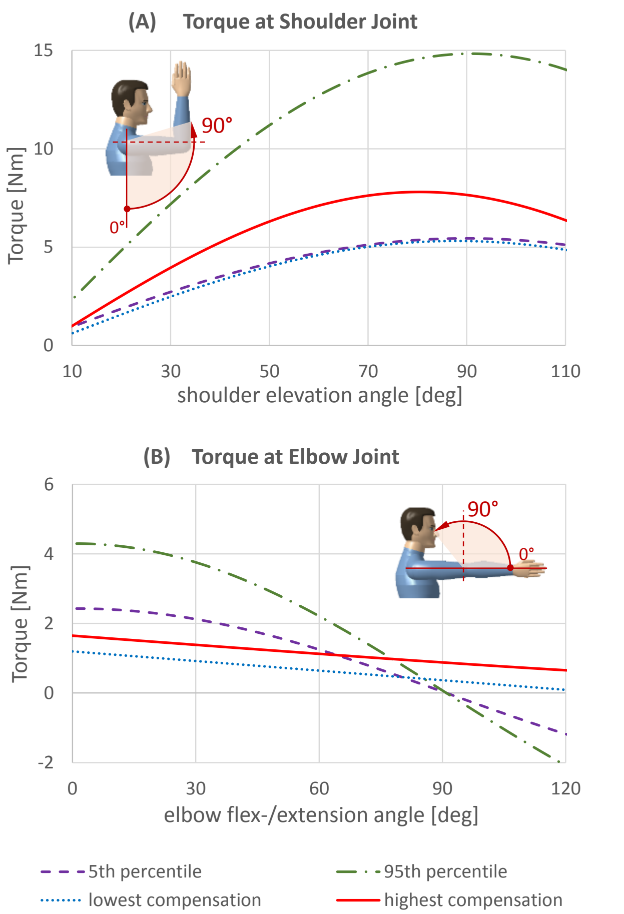

Figure 3. Gravity torques at shoulder joint *(A)* and elbow joint *(B)* for a 5th percentile (*purple dashed)* and 95th percentile *(green dashed and dotted)* subject in contrast to lowest *(blue dotted)* and highest *(red full)* possible compensation torques.

B. Musculoskeletal Model

A musculoskeletal model of the upper extremity of a 50th percentile adult male was used in this study, based on the work by Saul et al. [16]. The model consists of five body segments (thorax, upper arm, two segments for the forearm representing radius and ulna, hand) which are connected by 7 DoF: 3 DoF at the shoulder (thoracohumeral angle, elevation plane, shoulder rotation), elbow flexion/extension, forearm pro-/supination, wrist flexion/extension and wrist deviation. The model is actuated by 12 Hill type musculotendon actuators as shown in Fig. 4, with properties as defined by Thelen [17].

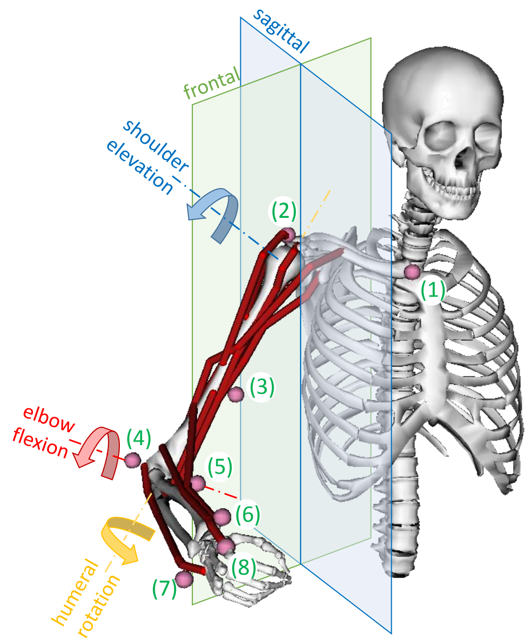

Figure 4. Musculoskeletal model of the upper limb with musculotendon actuators and virtual marker placement for motion recording. The musculotendon actuators include *m. Deltoideus ant., m. Deltoideus post.* and *med., m. Biceps short* and *long, m. Triceps long, m. Triceps med.* and *lat., m. Brachialis, m. Brachioradialis, m. Extensor carpi ulnaris* and *m. Extensor carpi radialis longus (the markers were placed at: (1) sternum, (2) acromion, (3) biceps, (4) lateral epicondyle, (5) forearm, (6) radial styoid, (7) ulnar styoid and (8) 2nd MCP joint)*

C. Subjects

Five healthy male subjects with no history of upper limb injuries were recruited for this study, mean age 32 ± 2.5 years and body height 1.80 ± 0.1 m. One subject was left handed, the other four were right handed. None of them had used a gravity compensated exoskeleton before. Informed consent for the experimental trials was provided by each of the subjects.

D. Measurement Setup

The exoskeleton was mounted on the backrest of a standard chair. As control of the stepper motor for adjustment of shoulder joint gravity compensation, a standard control (Nanotec SMCI33) in combination with the delivered software (NanoPro) was used. Kinematic data was collected at 120 Hz with an 8 camera motion analysis system (Motion Analysis Corporation). Electromyographic (EMG) signals were recorded with a wireless EMG signal detection system (Delsys Trigno Lab). The EMG signals together with the motion data were processed by the motion analysis software (Cortex, Motion Analysis). Maximum isometric joint torques were measured at each of the three DoFs using a force/torque sensor (K6D40, ME Messsysteme), which records three-dimensional forces and torques at 60 Hz. The systems were synchronized via a computer generated trigger impulse.

E. Data Collection

The exoskeleton was adjusted to each subjects individual shoulder height, upper arm length and forearm length. The upper limb mass of all five subjects required setting the exoskeleton to one-spring mode. Fine tuning of the gravity compensation was adjusted so that gravity slowly moved the fully extended arm (neutral elbow joint angle) downwards from an initial shoulder elevation angle of 90 degrees.

For the motion analysis, a total of 8 reflective markers were placed on the subject's torso and left upper limb as shown in Fig. 4. Four EMG electrodes were placed on the skin above the muscles *Deltoideus ant.*, *Deltoideus med.*, *Deltoideus post.* and *Biceps brachii*. The complete setup is shown in Fig. 5.

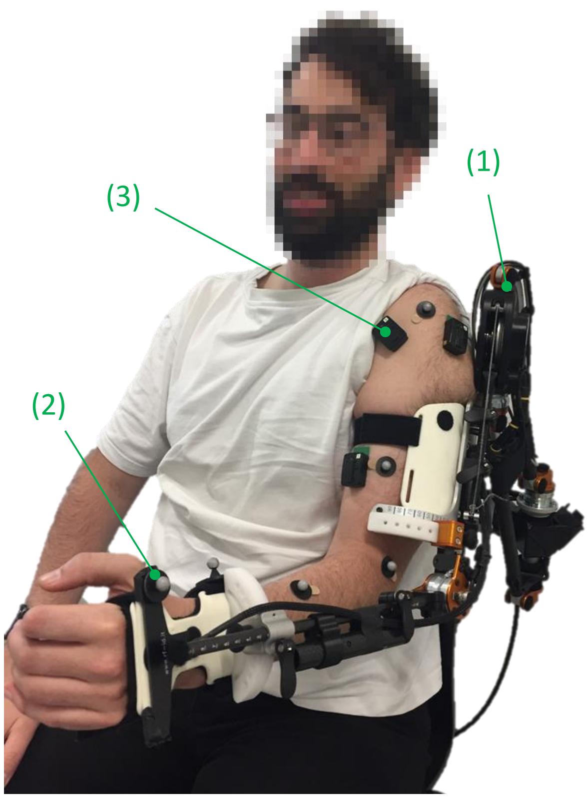

Figure 5. RETRAINER arm exoskeleton and measurement setup. On the left upper limb 8 reflective markes were placed on sternum, acromion, biceps, lateral epicondyle, forearm, radial styoid, ulnar styoid and 2nd MCP joint. *([1] exoskeleton, [2] reflective marker, [3] EMG sensor*)

After the donning phase and before the trials, the subjects were allowed to familiarize with the exoskeleton and the experimental setup. All experiments were carried out with an additional weight of 0.5 kg in the hand to enhance muscle excitations. Each subject had to perform three different movements: shoulder elevation in parasagittal plane (elbow neutral), shoulder elevation in frontal plane (elbow neutral) and hand-to-mouth task at humeral rotation 40°, each movement was repeated three times. Two sets of data were recorded, one with enabled and one with disabled gravity compensation. For disabled compensation the subjects were asked to raise the upper limb within 1.5 seconds. In case of enabled gravity compensation the subjects were asked to perform the movements with lowest possible muscle effort.

After performing the movements, for each subject the maximum isometric joint torques and the EMG at maximum voluntary contraction (MVC) of shoulder muscles and biceps were measured in isometric test routines [18].

F. Data Processing

The recorded EMG signals for each muscle and subject were smoothened by calculating the RMS with a fixed timeframe of 100 milliseconds. After normalization with the MVC the EMG signals were trimmed to the duration of the performed task. For each movement, the mean values and standard deviations of all three repetitions were calculated. For the general statistical analysis of all subjects, the mean and standard deviations over a normalized cycle time for every performed movement were calculated.

G. Simulation

The software package OpenSim [19], was used for the simulations. Joint angle trajectories for each movement were computed from motion capture data using the provided Inverse Kinematics tool. To determine muscle forces and activations, the computed muscle control (CMC) algorithm [20] was used, which uses forward simulation in combination with a feedback controller that tracks the input kinematics.

Average muscle activations and forces were determined for one subject representing an average of the test group. For the experiments with disabled gravity compensation, the muscle activations and forces were determined as described above. For the experiments with enabled gravity compensation (external joint torque), muscle activations and forces were estimated by comparing the measured EMG signal amplitudes with those from the measurements with disabled gravity compensation at the respective joint angles.

III. Results

Measured muscle activation levels at selected points of the movement cycles are shown in Table 1. The activation level of the *Deltoideus ant.* was highest during shoulder elevation in the parasagittal plane for all subjects. Also low activity of the *Deltoideus med.* was measured. Analogously, during shoulder elevation in the frontal plane, mostly the *Deltoideus med.* remains active in combination with low activity of the *Deltoideus post.*

The highest relative reduction of the muscle activity due to the enabled gravity compensation is shown to range from 30% to 80% of the motion time (corresponds to 25° to 75° shoulder elevation angle). For the hand-to-mouth task no significant differences of muscle activation levels between the trials with enabled and disabled gravity compensation were recognized.

The analysis of all performed tasks, with enabled gravity compensation, showed maximum peak excitations (normalized to the MVCs) for the *Biceps brachii* of 0.14 (0.04 std), *Deltoideus ant.* 0.40 (0.10 std), *Deltoideus med.* 0.31 (0.08 std) and for the *Deltoideus post.* 0.13 (0.10 std). In comparison with disabled gravity compensation maximum peak excitations for the *Biceps brachii* of 0.17 (0.04 std) (21% higher), *Deltoideus ant.* 0.54 (0.12 std) (35% higher), *Deltoideus med.* 0.44 (0.14 std) (41% higher) and for the *Deltoideus post.* 0.15 (0.11 std) (15% higher) were found.

For the shoulder elevation in sagittal and frontal planes, the peak values for the deltoid muscles were reached after

70% of the motion time (about 75° of shoulder elevation). During the hand-to-mouth movement the *Biceps brachii* showed peak activations between 30% and 80% motion time, *Deltoideus ant.* and *Deltoideus med.* showed normalized activation values in the range from 0.15 (0,04 std) to 0.34 (0.07 std). On average muscle activations were reduced up to 20% for elevation in the sagittal plane and up to 15% for elevation in the frontal plane over the complete duration of the movements.

TABLE I. NORMALIZED MUSCLE ACTIVATIONS FOR THE PERFORMED TASKS AVERAGED OVER ALL REPETITIONS AND SUBJECTS AT SELECTED CYCLE POINTS (% OF THE COMPLETE MOVEMENT); VALUES IN THE PARENTHESES ARE WITHOUT GRAVITY COMPENSATION.

Muscles	Shoulder Elevation Progress (sagittal plane)				
	20%	*40%*	*60%*	*80%*	*100%*
Biceps brachii	0.01 (0.023)	0.028 (0.061)	0.033 (0.058)	0.038 (0.057)	0.025 (0.062)
Deltoideus ant.	0.026 (0.08)	0.127 (0.249)	0.194 (0.432)	0.296 (0.469)	0.4 (0.472)
Deltoideus med.	0.023 (0.043)	0.036 (0.094)	0.07 (0.146)	0.094 (0.165)	0.131 (0.188)
Deltoideus post.	0.008 (0.006)	0.009 (0.013)	0.014 (0.018)	0.013 (0.019)	0.014 (0.017)
Muscles	**Shoulder Elevation Progress (frontal plane)**				
	20%	*40%*	*60%*	*80%*	*100%*
Biceps brachii	0.09 (0.02)	0.021 (0.037)	0.031 (0.052)	0.051 (0.062)	0.054 (0.078)
Deltoideus ant.	0.019 (0.059)	0.041 (0.103)	0.094 (0.179)	0.14 (0.247)	0.186 (0.315)
Deltoideus med.	0.051 (0.125)	0.123 (0.227)	0.189 (0.337)	0.246 (0.362)	0.315 (0.444)
Deltoideus post.	0.015 (0.056)	0.022 (0.093)	0.059 (0.128)	0.111 (0.136)	0.119 (0.148)
Muscles	**Hand-to-Mouth Task (drinking)**				
	20%	*40%*	*60%*	*80%*	*100%*
Biceps brachii	0.053 (0.085)	0.125 (0.135)	0.123 (0.117)	0.119 (0.046)	0.104 (0.037)
Deltoideus ant.	0.234 (0.244)	0.265 (0.288)	0.304 (0.322)	0.320 (0.300)	0.298 (0.270)
Deltoideus med.	0.130 (0.114)	0.129 (0.114)	0.135 (0.125)	0.121 (0.107)	0.111 (0.097)
Deltoideus post.	0.029 (0.027)	0.031 (0.028)	0.030 (0.029)	0.029 (0.025)	0.028 (0.023)

The simulations predict required peak muscle forces to be 20% to 80% lower with enabled gravity compensation as shown in Table 2. As illustrated in Fig. 6 the simulation results for the muscle activations during the movements came close to the measured trajectories. Simulated activations have smoother trajectories compared to the EMG measurements of the chosen subject.

In Fig. 7 a comparison of the predicted muscle forces with enabled and disabled gravity compensation during a full movement cycle is shown. It can be seen, that muscle forces are significantly reduced for the *Deltoideus ant.* for elevation in sagittal plane and for the *Deltoideus med.* and *Deltoideus post.* for elevation in frontal plane. The highest muscle force reductions are visible from 20% to 75% of motion time.

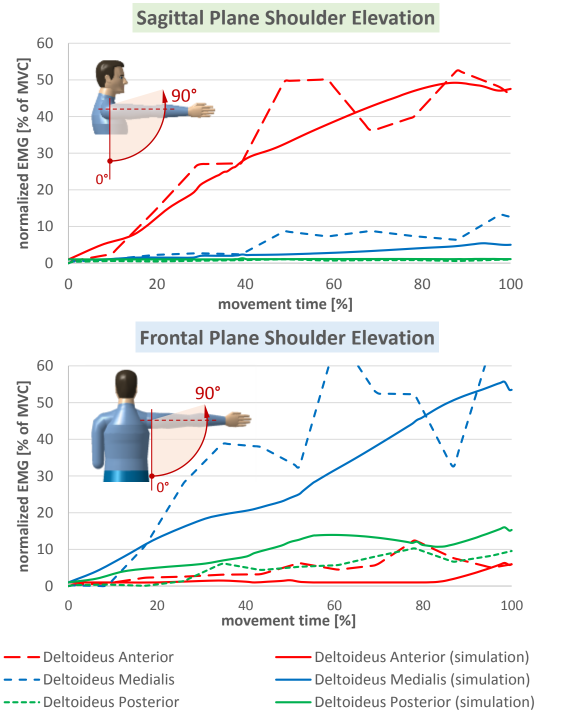

Figure 6. Measured muscle activations for disabled weight compensation in comparison with the simulation results for muscle activations for the displayed movements shown for one subject.
(the dashed line represents the measured muscle activations, the full line represents the results of the simulation)

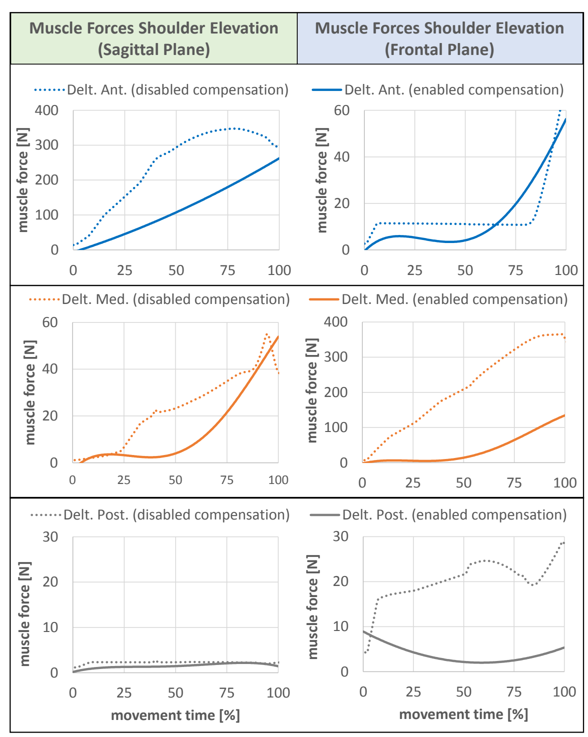

Figure 7. Comparison of predicted shoulder muscle forces from simulation for enabled *(full lines)* and disabled *(dashed lines)* gravity compensation for one subject.

TABLE II. PREDICTED PEAK FORCES FOR ALL DELTOID MUSCLES FOR PERFORMED MOVEMENT WITH (W.C.) AND WITHOUT (WO.C.) GRAVITY COMPENSATION

Muscle Peak Forces for Shoulder Elevation				
Muscle	*Sagittal Plane*		*Frontal Plane*	
	w.c.	*wo.c.*	*w.c.*	*wo.c.*
Deltoideus ant.	173 N	347 N	11 N	11 N
Deltoideus med.	31 N	38 N	67 N	343 N
Deltoideus post.	2 N	3 N	9 N	25 N

IV. DISCUSSION

The experiments with the RETRAINER exoskeleton have shown, that the muscle activations were reduced during shoulder elevation movements when gravity compensation was enabled, what underlined the effectivity of the gravity compensation mechanism. The muscle activations determined by the simulations were close to the measured values, indicating the validity of the musculoskeletal model. Previous studies on the ranges of motion in ADL have shown, that for most tasks no extreme angular positions are reached, motions stay within 20°-80° of possible joint ranges [21]. The results of this study show, that the highest rates of force reduction due to the weight compensation (60% *Delt. ant.*, 80% *Delt. med.*, 75% *Delt. post.*) are generated in this range as well.

For the hand-to-mouth task, no significant differences in muscle activations between enabled and disabled gravity compensation were recognized. As the movement of the forearm stays mainly in an approximately horizontal plane, gravity compensation is not useful here. Consequently, adjustment of the elbow gravity compensation has to be adapted to the planned movements.

One possible shortcoming of this study is, that the EMG signals, which were recorded with surface electrodes, might have been influenced by small relative movements between skin and muscles. Also, the measurements were done with healthy test subjects, this could influence the results as it is difficult to relax and provide only the required muscle forces, which are lower than expected from experience. This may also lead to activation of antagonists to control the movement.

To summarize, this study indicates that beside its compact and lightweight design the gravity compensation of the RETRAINER exoskeleton can significantly reduce required muscle forces for arm movements and hence effectively support the rehabilitation process. Based on the results specific training movements for stroke patients can be assembled according to their individual needs.

REFERENCES

[1] J. Mackay and G. A. Mensah, *The Atlas of Heart Disease and Stroke*, vol. 19. 2004.

[2] A. Fusco *et al.*, "The ABC of tDCS: Effects of anodal, bilateral and cathodal montages of transcranial direct current stimulation in patients with stroke - A pilot study," *Stroke Res. Treat.*, 2013.

[3] T. Krabben *et al.*, "Influence of gravity compensation training on synergistic movement patterns of the upper extremity after stroke, a pilot study," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, 2012.

[4] M. D. Ellis, T. Sukal, T. DeMott, and J. P. A. Dewald, "Augmenting clinical evaluation of hemiparetic arm movement with a laboratory-based quantitative measurement of kinematics as a function of limb loading," *Neurorehabil. Neural Repair*, vol. 22, no. 4, pp. 321–329, 2008.

[5] G. Kwakkel, B. J. Kollen, and H. I. Krebs, "Effects of robot-assisted therapy on upper limb recovery after stroke: A systematic review," *Neurorehabilitation and Neural Repair*, vol. 22, no. 2. pp. 111–121, 2008.

[6] S. J. Housman, K. M. Scott, and D. J. Reinkensmeyer, "A randomized controlled trial of gravity-supported, computer-enhanced arm exercise for individuals with severe hemiparesis," *Neurorehabil. Neural Repair*, vol. 23, no. 5, pp. 505–514, 2009.

[7] M. D. Ellis, C. Carmona, J. Drogos, and J. P. A. Dewald, "Progressive abduction loading therapy with horizontal-plane viscous resistance targeting weakness and flexion synergy to treat upper limb function in chronic hemiparetic stroke: A randomized clinical trial," *Front. Neurol.*, vol. 9, no. FEB, 2018.

[8] R. Riener, M. Guidali, U. Keller, A. Duschau-Wicke, V. Klamroth, and T. Nef, "Transferring ARMin to the Clinics and Industry," *Top. Spinal Cord Inj. Rehabil.*, vol. 17, no. 1, pp. 54–59, 2011.

[9] T. Rahman, W. Sample, and R. Seliktar, "Design and testing of WREX," in *Advances in rehabilitation robotics: human-friendly technologies on movement assistance and restoration for people with disabilities*, 2004, vol. 306, pp. 243–250.

[10] S. J. Housman, L. Vu, T. Rahman, R. J. Sanchez, and D. J. Reinkensmeyer, "Arm-Training with T-WREX After Chronic Stroke: Preliminary Results of a Randomized Controlled Trial," *Rehabil. Robot. 2007. ICORR 2007. IEEE 10th Int. Conf.*, vol. 0, no. c, pp. 562–568, 2007.

[11] S. D. Iftime-Nielsen, M. S. Christensen, R. J. Vingborg, T. Sinkjær, A. Roepstorff, and M. J. Grey, "Interaction of electrical stimulation and voluntary hand movement in SII and the cerebellum during simulated therapeutic functional electrical stimulation in healthy adults," *Hum. Brain Mapp.*, vol. 33, no. 1, pp. 40–49, 2012.

[12] M. Gandolla *et al.*, "Re-thinking the role of motor cortex: Context-sensitive motor outputs?," *Neuroimage*, vol. 91, pp. 366–374, 2014.

[13] E. Ambrosini *et al.*, "The combined action of a passive exoskeleton and an EMG-controlled neuroprosthesis for upper limb stroke rehabilitation: First results of the RETRAINER project," in *IEEE Int.l Conference on Rehabilitation Robotics*, 2017, pp. 56–61.

[14] A. Pedrocchi *et al.*, "MUNDUS project: MUltimodal Neuroprosthesis for daily Upper limb Support," *J. Neuroeng. Rehabil.*, vol. 10, no. 1, 2013.

[15] M. Puchinger *et al.*, "The ReTrainer arm exoskeleton - a modular lightweight device that combines adjustable gravity compensation, lockable joints and neuromuscular electrical stimulation," in *Proceedings of the 12th Vienna Int. Workshop on FES*, 2006, pp. 11–14.

[16] K. R. Saul *et al.*, "Benchmarking of dynamic simulation predictions in two software platforms using an upper limb musculoskeletal model," *Comput. Methods Biomech. Biomed. Engin.*, vol. 18, no. 13, pp. 1445–1458, 2015.

[17] D. G. Thelen, "Adjustment of Muscle Mechanics Model Parameters to Simulate Dynamic Contractions in Older Adults," *J. Biomech. Eng.*, vol. 125, no. 1, p. 70, 2003.

[18] R. K. W. Douma, R. Soer, W. P. Krijnen, M. Reneman, and C. P. van der Schans, "Reference values for isometric muscle force among workers for the Netherlands: A comparison of reference values," *BMC Sports Sci. Med. Rehabil.*, vol. 6, no. 1, 2014.

[19] S. L. Delp *et al.*, "OpenSim: Open-source software to create and analyze dynamic simulations of movement," *IEEE Trans. Biomed. Eng.*, vol. 54, no. 11, pp. 1940–1950, 2007.

[20] K. R. S. Holzbaur, W. M. Murray, and S. L. Delp, "A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control," *Ann. Biomed. Eng.*, vol. 33, no. 6, pp. 829–840, 2005.

[21] J. Karner, W. Reichenfelser, and M. Gfoehler, "Kinematic and kinetic analysis of human motion as design input for an upper extremity bracing system," in *Proceedings of the IASTED International Conference Biomedical Engineering*, 2012, pp. 376–383.