Physical Burden in Manual Patient Handling: Quantification of Lower Limb EMG Muscle Activation Patterns of Healthy Individuals Lifting Different Loads Ergonomically

Anna Brinkmann[a], Conrad Fifelski-von Böhlen[b], Sandra Hellmers[c], Ole Meyer[d], Rebecca Diekmann[e] and Andreas Hein

Assistive Systems and Medical Device Technology, Carl von Ossietzky University of Oldenburg, 26129 Oldenburg, Germany

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Abstract: Manual patient handling is a challenging part of daily care and leads to high mechanical loads as well as to the development of degenerative diseases, e.g. lower back pain. To prevent musculoskeletal overload effects, the use of ergonomic working techniques is essential as well as improving caregivers' functional ability. However, most of the studies do not consider these aspects and biomechanical evaluations including dynamic electromyography (EMG) are rarely analyzed. In this work, we focus on the quantification of lower limb EMG muscle activation patterns of healthy caregiver students in an experimental setup. The extent of lifting different loads ergonomically is analyzed and similarities/dissimilarities of dynamic EMG data of three lower limb muscles are investigated via cross-correlation calculation. One of the main findings of our investigation is an indication of a more consistent mean activity of the quadriceps and hamstring musculature, as the load to be lifted increases. Furthermore, we found an intra- as well as an interindividual similarity of EMG muscle activation patterns regarding time and shape of the signals generated during all of the conducted lifting tasks with a predominantly high cross-correlation coefficient for the selected muscles of the lower limb.

1 INTRODUCTION

Manual patient handling is one of the most significant challenges in care and leads to high mechanical loads as well as to the development of degenerative diseases, e.g. lower back pain (Hwang et al., 2019; Choi and Brings, 2016; Jäger et al., 2013). In particular, lift, hold and handle especially overweight and obese patients manually is physically demanding and leads to a compressive strength of the lumbar spine of up to 9 kN (Choi and Brings, 2016; Jäger et al., 2013).

To prevent musculoskeletal overload effects significantly, the correct use of technical devices as well as ergonomic caregiving strategies like supervised ergonomic exercise training programs are essential (Hwang et al., 2019; Choi and Brings, 2016; Weißert-Horn et al., 2014; Jäger et al., 2013; Michaelis and Hermann, 2010).

A functional approach to ergonomic working strategies and the improvement of caregivers' power as well as functional ability is squat training (Kusma et al., 2015; Jäger et al., 2013; Baum et al., 2012) as the squat is biomechanically as well as neuromuscular similar to many activities of daily living, e.g. standing up from a chair, demanding the musculoskeletal system of the human body more than 50 times per day (Wang et al., 2019). As squatting positions are also part of ergonomic manual caregiving routines, e.g. standing a patient up for transfer, the squat is frequently used in exercise programs of strength and conditioning as well as in physical therapy (Yavuz and Erdag, 2017). In this case, lumbosacral loads can be compensated by strengthening the lower limb

[a] https://orcid.org/0000-0001-5228-4947
[b] https://orcid.org/0000-0002-6118-2755
[c] https://orcid.org/0000-0002-1686-6752
[d] https://orcid.org/0000-0002-9964-5591
[e] https://orcid.org/0000-0001-9793-3832

Brinkmann, A., Böhlen, C., Hellmers, S., Meyer, O., Diekmann, R. and Hein, A.
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and back muscles. Thus, effective load transfer from the lumbar spine to the pelvis is achieved and shearing of the sacroiliac joints through compression is prevented (Vleeming and Stoeckart, 2007; Richardson et al., 2002).

Common faults in squatting exercises of daily living as well as in the context of professional caregiving are faster rising hips than shoulders and thus resulting in an increasing flexion of the trunk (Hellmers et al., 2021; Yavuz and Erdag, 2017). In this case, the distance between hips and shoulders is diminished in vertical direction when rising upright from squatting and the lumbar load increases (Hellmers et al., 2021; Yavuz and Erdag, 2017). The aim of squatting exercises is to train quadriceps musculature around the knee and hip joints, thereby strengthening the lower back (Yavuz and Erdag, 2017). By using the squat as an ergonomic working strategy for lifting patients, quadriceps muscles are activated, resulting in a more consistent mean activation of the back extensor muscles (Brinkmann et al., 2020a; Brinkmann et al., 2020b).

In recent literature, various scientific articles on analyzing caregiving activities exist. These deal with both the identification of psychological as well as physical stress of healthcare workers and the enhancement of existing strategies for preventing job-related back disorders (Cheung et al., 2020; Vinstrup et al., 2020; Hwang et al., 2019; Choi and Brings, 2016; Höhmann et al., 2016; Zhao et al., 2016; Kusma et al., 2015; Weißert-Horn et al., 2014; Jäger et al., 2013; Baum et al., 2012; Aiken et al., 2012; Michaelis and Hermann, 2010). However, most of the studies do not consider functional aspects in this context, such as physical functionality of the caregivers. In addition, the effects of applying ergonomic working techniques and its biomechanical evaluation, including dynamic electromyography (EMG) for quantifying muscle activation patterns objectively, are rarely analyzed (Cheung et al., 2020; Vinstrup et al., 2020; Hwang et al., 2019).

In this work, we focus on the quantification of lower limb muscle activity of healthy caregiver students in an experimental setup. The extent of lifting different loads ergonomically is analyzed and similarities/dissimilarities of dynamic EMG data of three lower limb muscles are investigated via cross-correlation. We hypothesize a similar EMG activation for all three conditions while mean muscle activity increases with lifting heavier weights. The aim is the assessment of potential amplitude independent changes of EMG muscle activation patterns as a function of different ergonomic lifting conditions.

2 MATERIALS & METHODS

2.1 Study Design

In the case study (ethical vote: Drs.EK/2019/004), five healthy caregiver students (n = 5, 3 female and 2 male students aged between 21 to 45) conduct three different dynamic lifting tasks:

(1) lifting the own body weight by rising from a chair to an upright position (Figure 1a, (1)),
(2) lifting a patient simulator (13 kg) from the edge of a motorized care bed to a standing position (Figure 1a, (2)),
(3) lifting a patient-imitating subject (patient) (female, 28 years, 63 kg) upright from the edge of a motorized care bed (Figure 1a, (3)).

To avoid overloading the caregivers while lifting, a physiotherapist supervises the tasks.

In the first task (Figure 1a, (1)), the caregivers' initial position is a vertical trunk with crossed arms. The feet are placed flat on the floor in shoulder width and the knee angle is > 90° to avoid extreme joint load (Slater and Hart, 2017). While rising to fully standing upright, stability is maintained through muscle activation. In this case, the components balance, coordination and lower limb strength as well as power are covered, which are important in view of analyzing caregivers' physical function quantitatively (Hardy et al., 2010). All in all, each caregiver student repeats the task five times. The second as well as the third lifting task (Figure 1a, (2) and (3)) are conducted in accordance to the Kinaesthetics (Hatch, 2003) care conception. Therefore, the lifting is executed in different stages and is foresighted ergonomic planned with a consistent use of aids. The care bed is adjusted to an appropriate working height, so that the patient's as well as the patient simulator's feet are flat on the floor while sitting at the bed's edge. The caregiver stands parallel to the patient and slightly squats bending down to the patient for lifting up. The knee angle is > 90°, as extreme joint load is thus prevented (Slater and Hart, 2017). In the second task, the caregiver puts his arms around the simulated patient and lifts while rising from squatting (Figure 1a, (2)). Then, both are in an upright position. Compared to the second lifting task, there is an active interaction between patient and caregiver in the third lifting task. In this case the patient puts his arms around the caregiver (Figure 1a, (3)), so that the functionality of the patient can be used for cooperation while lifting. Then, the caregiver also puts his arms around the patient and shifts the own body weight while remaining a straight back to finally lift up.

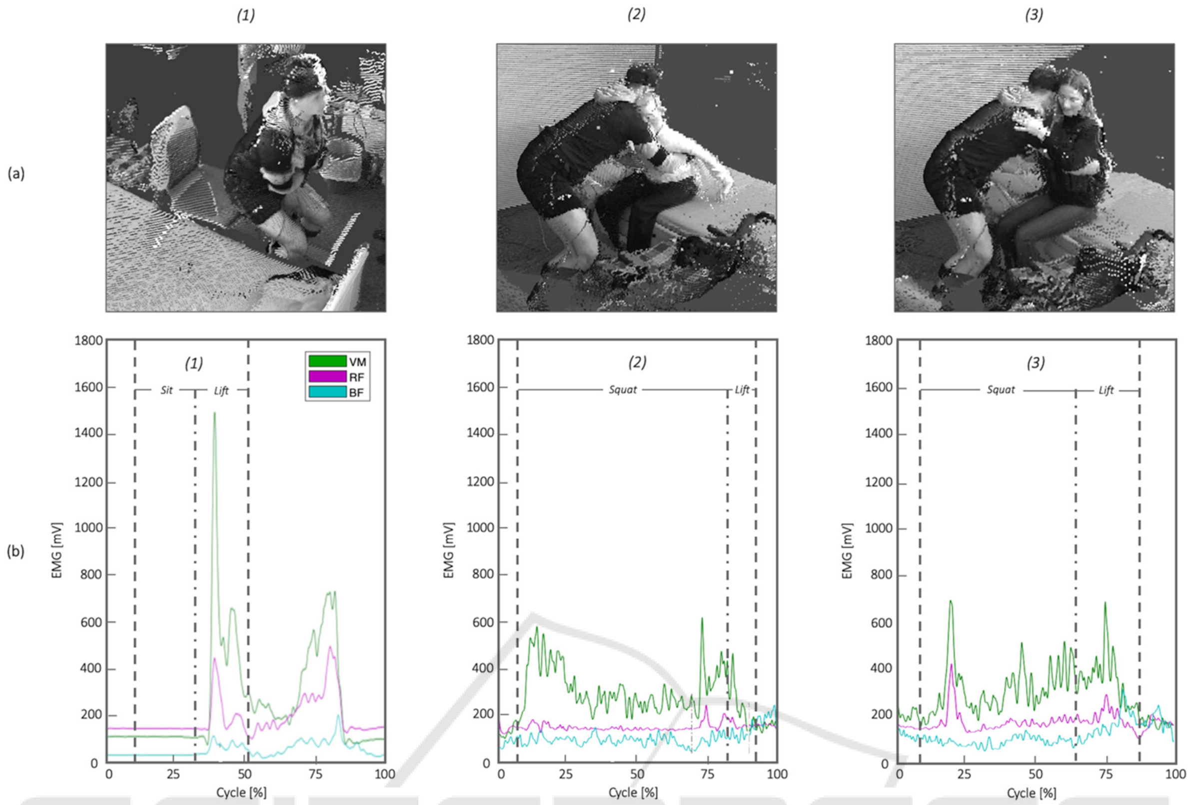

Figure 1: Kinematic (a) and EMG data (b) of three different ergonomic lifting tasks of one exemplary study participant. Kinematic data is shown in three-dimensional point clouds of the respective lifting task. The time courses for muscle activity data of three muscles of the lower limb (vastus medialis (VM), rectus femoris (RF) and biceps femoris (BF)) are shown as a function of the respective lifting task cycle for squatting and lifting for task (1) – (3).

2.2 Biomechanical Data Collection

The procedure for biomechanical data collection is based on our existing Healthcare Prevention System (Brinkmann et al., 2020a; Brinkmann et al., 2020b). Thus, the kinematics of the moving body and its segments, the kinetics (external ground reaction forces) and muscle activities of the caregivers' lower limb are recorded in order to quantify, assess and evaluate the executed processes biomechanically.

By direct measurement techniques, a 3D multi-depth image camera system (Fifelski et al., 2018) record the data required for motion analysis and a force plate is used for the measurement of occurring external ground reaction forces while transferring the patient. Non-invasive surface EMG records electrical action potentials associated with muscle contraction and is the main focus in this work. EMG is used in order to gain information on the activation behavior of the following selected muscle groups of the caregivers' thigh in task (1) – (3): vastus medialis (VM), rectus femoris (RF) and biceps femoris (BF). These muscles are part of the knee extensors as well as the hip extensors and are thus primarily active during the conducted dynamic squat exercises. The electrodes are placed in accordance with SENIAM guidelines (Hermens et al., 1999). For the acquisition process, Dasy-Lab 4.010 software as well as an EMG device from Biovision (Biovision Inputbox) and bipolar surface electrodes (Ø 14 mm; 10 mm inter-electrode distance) are used (GE Medical/Hellige). By local amplifiers, an amplification of the signal with 2500 Hz is done.

2.3 Data Analysis

In a first step, recorded dynamic EMG data is rectified and then smoothed via Root Mean Square (RMS) (Figure 2). Then, the data recorded while lifting the own body weight is cut according to kinematic and kinetic data representing the basis for analyzing the different lifting tasks. In task (1), a lifting cycle starts with sitting and ends with fully standing (Figure 1b, (1) and Figure 2).

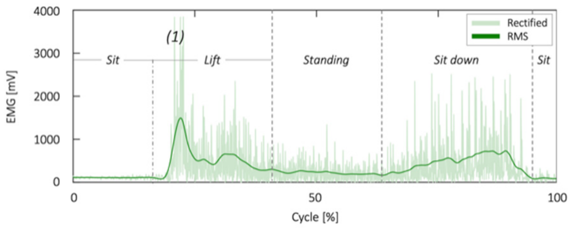

Figure 2: Rectified and RMS smoothed EMG data of VM activity in task (1) for one exemplary study participant.

Due to the fact, that every cycle reveals a slightly different duration for each participant as well as between all participants, time normalization is applied using linear interpolation function. Thus, a mean lifting signal is calculated separately for each participant and muscle. Then, an intraindividual as well as an interindividual normalized cross-correlation analysis is done by calculating cross-correlation coefficient (*R*-value) at zero time lag to test dynamic EMG data for similarities/dissimilarities (Geiger et al., 2019; Nelson-Wong et al., 2009; Wren et al, 2006) for each muscle and each lifting task as follows:

$$R_{xy}(\tau) = \frac{\frac{1}{N}\sum_{i=1}^{N}(x_i - \overline{x})(y_{i+\tau \cdot f_s} - \overline{y})}{\frac{1}{N}\sqrt{\sum_{i=1}^{N}(x_i - \overline{x})^2 \sum_{i=1}^{N}(y_i - \overline{y})^2}} \quad (1)$$

with x_i and y_i as the two signals to be compared. τ is the discrete temporal time shift, N is the number of data points in the respective signal and f_s is the original sample frequency (Nelson-Wong et al., 2009).

Via cross-correlation the comparison of two signals regarding timing and shape is possible, while amplitude is not considered. Therefore, the signals mean power is also reflected by using RMS, while its mean value qualifies gross innervation input for respective muscles. This step is then followed by comparing mean EMG activity for each study participant when lifting their own body weight in task (1). In this case, we analyse the data due to similar mean muscle activation patterns while lifting with an interindividual point of view. The participants, who show plausible mean muscle activation patterns in task (1) while rising from a seated position to an upright position and thereby lifting the own body weight (Wang et al., 2019; Roldán-Jiménez et al., 2015; Cuesta-Vargas and Gonzáles-Sanchez, 2013; Roebroeck et al., 1994) are therefore constituted to one functional group and considered for the evaluation via cross-correlation calculation.

3 RESULTS

For concentric knee extension when lifting (Figure 1a), VM and RF contract at the same time while hamstring muscle activation (BF) sustains the hip (Figure 1b). All in all, three out of five study participants show plausible mean muscle activation patterns while rising from a seated position to an upright position and thereby lifting the own body weight in task (1) (Figure 1a, (1)). This means, that the gross innervation input of the analyzed muscles is highest for VM, followed by RF and BF (Wang et al., 2019; Roldán-Jiménez et al., 2015; Cuesta-Vargas and Gonzáles-Sanchez, 2013; Roebroeck et al., 1994). Accordingly, these participants are constituted to one functional group and therefore considered for the evaluation (Figure 3). The other group of participants show different mean activation patterns with a predominantly high activation level of RF and are not considered for further analysis in this work.

For the functional group of study participants, mean muscle activities of VM, RF and BF while lifting the own body weight in task (1) (Figure 1a, (1)) are: VM = 350 mV, RF = 240 mV and BF = 80 mV (Figure 3, (1)).

While lifting the patient simulator in task (2) (Figure 1a, (2)), mean muscle activities of VM, RF and BF are: VM = 220 mV, RF = 190 mV and BF = 75 mV (Figure 3, (2)).

In task (3), while lifting the patient (Figure 1a, (3)), mean muscle activities of VM, RF and BF are: VM = 250 mV, RF = 245 mV and BF = 220 mV (Figure 3, (3)).

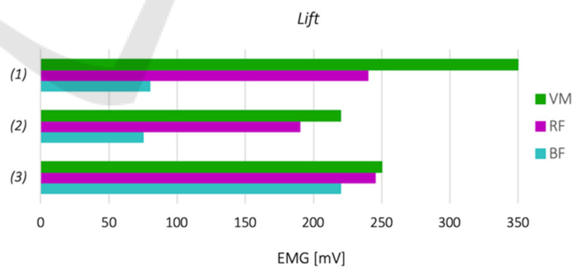

Figure 3: Mean muscle activity data of VM, RF and BF for task (1) – (3).

Comparing the dynamic mean muscle activity data of the conducted lifting tasks regarding the functional group of study participants, similar mean muscle activation patterns are present (Figure 3). Due to the gross innervation input of the analyzed muscles while lifting different loads ergonomically, the highest value is found for VM, followed by RF and BF (Figure 3). Comparing the tasks (1), (2) and (3), the highest mean muscle activity values for VM and RF are found for lifting the own body weight from a seated position (Figure 3, (1)), followed by lifting the patient (63 kg) (Figure 3, (3)) and lifting the patient simulator (13 kg) (Figure 3, (2)).

The delta between mean muscle activity of VM and RF is 110 mV while standing up, 30 mV while standing the patient simulator up (13 kg) and 15 mV while standing the patient up (63 kg). Accordingly, the delta between mean muscle activity of RF and BF is 160 mV while standing up, 115 mV while standing the patient simulator up and 25 mV while standing the patient up. Furthermore, there is an increase of BF's mean muscle activity for all of the three lifting tasks of up to 145 mV while standing the patient up compared to lifting the own body weight as well as lifting the patient simulator. In detail, comparing task (2) and (3), the deviation of the delta of RF and BF is 78% while lifting the higher weight (63 kg) in comparison to lifting the patient simulator (13 kg) (Figure 3). In this case, the deviation of the delta of VM and RF is 50 %.

Figure 4 shows the dynamic EMG data of the lifting parts of task (1), (2) and (3) for one exemplary study participant and the calculation of the *R*-values is presented as a function of phase shift with an intraindividual high similarity of EMG activation for all three conditions.

All calculated intraindividual *R*-values for the functional group of study participants are shown in Table 1. Here, the *R*-values show high to very high correlation for each muscle among the different lifting tasks and for each study participant. However, lower *R*-values and greater variability are found for the BF within the execution of the different lifting task of one participant (437, Table 1).

Interindividual comparison of the dynamic EMG data (Table 2) show very high *R*-values for VM, averaging > 0.90. In this case, the similarity for the VM within the functional group of study participants is $R = 0.96 \pm 0.005$ for test (1), $R = 0.96 \pm 0.014$ for test (2) and $R = 0.96 \pm 0.005$ for test (3) (Table 2). The interindividual cross-correlation result for RF is $R = 0.91 \pm 0.026$ for test (1), $R = 0.89 \pm 0.069$ for test (2) and $R = 0.97 \pm 0.012$ for test (3) (Table 2). By comparing muscle activity data for BF within the different study participants (Table 2), the *R*-value is $R = 0.95 \pm 0.022$ for test (1), $R = 0.96 \pm 0.025$ for test (2) and $R = 0.86 \pm 0.043$ for test (3).

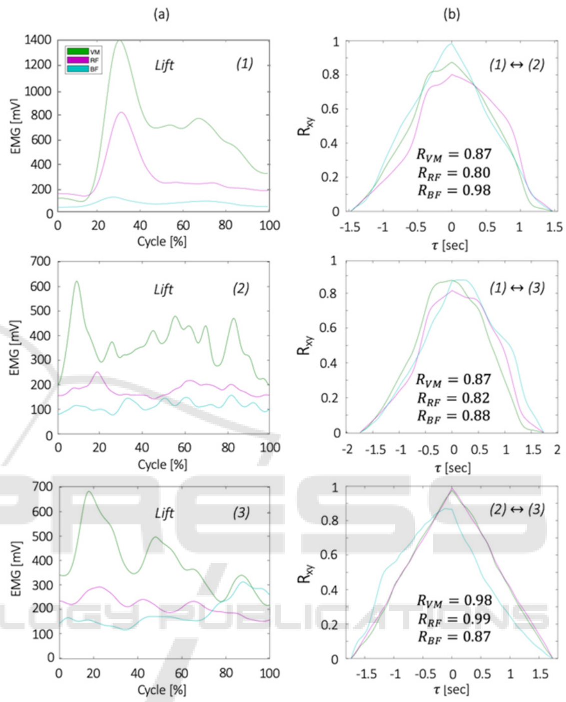

Figure 4: EMG data (a) and *R*-values (b) for task (1) – (3) and for one exemplary study participant. EMG time courses of VM, RF and BF are shown as a function of the respective lifting task cycle. *R*-value calculation is presented as a function of phase shift.

Table 1: Intraindividual *R*-values for each muscle and the respective lifting task correlation.

ID	Muscle	*R*		
		(1) ↔ (2)	(1) ↔ (3)	(2) ↔ (3)
424	VM	0.87	0.87	0.98
	RF	0.80	0.82	0.99
	BF	0.98	0.88	0.87
437	VM	0.89	0.86	0.95
	RF	0.89	0.94	0.97
	BF	0.68	0.80	0.67
471	VM	0.89	0.90	0.96
	RF	0.99	0.99	0.99
	BF	0.96	0.91	0.92

Table 2: Interindividual *R*-values for each muscle and the respective lifting task correlation.

Muscle	Test	*R*		
		424 ↔ 437	424 ↔ 471	437 ↔ 471
VM	(1)	0.95	0.96	0.96
	(2)	0.94	0.97	0.97
	(3)	0.96	0.96	0.97
RF	(1)	0.93	0.87	0.92
	(2)	0.83	0.99	0.86
	(3)	0.96	0.99	0.97
BF	(1)	0.93	0.98	0.94
	(2)	0.99	0.95	0.93
	(3)	0.92	0.82	0.84

4 DISCUSSION

We focused on the quantification of lower limb EMG muscle activation patterns of healthy caregiver students while lifting different loads ergonomically. In an experimental setup in the field, the extent of kinematic, kinetic and muscular activity is investigated while three different dynamic lifting tasks are conducted (Figure 1).

In each task, the caregivers' stability is maintained through muscle activation while distributing the own body weight evenly before standing up from a seated position as well as lifting the simulated patient (13 kg)/the patient (63 kg). In accordance with literature findings (Yavuz and Erdag, 2017; Aspe and Swinton, 2014; Paoli et al., 2009; Boyden et al., 2000; McCaw and Melrose, 1999), mean muscle activity increases with lifting higher loads in our experimental case study. A more consistent mean activity of the quadriceps and hamstring musculature is indicated, as the load to be lifted gets higher. Thereby, concentric knee extension and eccentric resistance to knee flexion activates the quadriceps muscles (Figure 1b). The hamstrings are quadriceps' antagonists, as these muscles oppose knee extensor moments (Yavuz and Erdag, 2017). However, in squatting exercises RF and BF paradoxically co-contract. With increasing load, BF muscle activity increases as well (Figure 1b, (3)). The effect of increasing mean muscle activity of BF in our case study could be due to co-contraction for stabilizing the knee as well as the pelvis while turning from an eccentric to a concentric movement. In future research, muscle fatigue could be another relevant topic. Literature findings indicate an increasing muscle fatigue of the knee extensors with increasing task repetitions (Roldán-Jiménez et al., 2015). For this purpose, the repetitions of task (1) should be increased in future studies.

For the quantification of the EMG muscle activation patterns generated in our experimental case study, we use cross-correlation calculation for comparing the data from different lifting scenarios and different individuals objectively. In a first step, the muscle activation patterns of lifting the own body weight in task (1) are intraindividual analyzed for the functional group of study participants. Here, cross-correlation results (R-values) show similar activation for the five lifting cycles with slightly differences in form as well as in duration. By using linear interpolation function for intraindividual normalized cross-correlation analysis at zero time lag, the R-values indicate a high similarity between different lifting patterns (Table 1) and a significant similarity when comparing task (2) and (3) with somewhat moderate correlation for the BF of one participant (Table 1, 437, $R = 0.72 \pm 0.060$). This may reflect a greater variability regarding muscle activation within different lifting scenarios as well as a sensitivity of this muscle due to its biarticular function (Wren et al., 2006). In this case, a greater variability could be also due to a knee angle < 90°, which has to be verified in future studies. Furthermore, muscle's length and overlying fat mass could be the reason for an increasing sensitivity to EMG electrode placement (Wren et al., 2006). Although the BF as well as the RF in some cases exhibit a greater variability than the VM (Table 1), the R-values are still high. Comparing the dynamic EMG data of the functional group of study participants interindividual (Table 2), very high R-values for VM, averaging > 0.90, are found for all lifting scenarios. In future research, the constitution of study participants to functional groups needs to be further investigated. In this case, cross-correlation analysis could be used to verify inter- as well as intra-individual similarities/dissimilarities.

It should be noted, that no real patient was recruited for our case study. Although, the use of the patient simulator in task (2) prevents unintentional subliminal cooperation and supportive behavior throughout the tests, the variety of possible non-cooperating patient behavior of e.g. anesthetized or obese patients is not fully covered. This is due to the low weight of the patient simulator (13 kg). However, the weight of the patient (63 kg) is within a realistic range and by using the patient's functionality in task (3), cooperative patient behavior is represented. Nevertheless, it still has to be distinguished from lifting a real patient. Therefore, it can be assumed, that muscle activity data under realistic circumstances may be higher than provided in this work.

The main findings of our experimental case study are an intraindividual as well as an interindividual similarity of EMG muscle activation patterns regarding time and shape of the signals generated during all of the three conducted lifting tasks. In this case, the R-values are predominantly high for the selected muscles of the lower limb, especially for the VM. These results provide a first insight into the quantification of EMG muscle activation patterns of healthy caregivers lifting different loads ergonomically and serve as a basis for further investigations with a larger study population. Based on future research, the results may enhance both supervised ergonomic exercise programs in the education of caregivers and to allow for a more targeted use in training interventions from a functional point of view.

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