The neuromechanical and behavioural adaptations to dynamic arm supports in neuromuscular disorders

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J.M.N. Essers

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Johannes Maria Nicolaas Essers

The research presented in this thesis was performed within NUTRIM School of Nutrition and Translational Research in Metabolism and conducted out at the Department of Nutrition and Movement Sciences, Maastricht University Medical Centre+ (MUMC+), Maastricht, The Netherlands and at the Department of Human Movement Sciences, University Medical Center Groningen (UMCG), Groningen, The Netherlands. This work is part of the project ADAPT (with project number 13523) of the research programme Symbionics Perspectief Program which is (partly) financed by the Dutch Research Council (NWO).

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DISSERTATION

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CHAPTER 1

General introduction

Interaction with our surroundings is an essential element of our activities of daily life. Whether it is to prepare and eat a meal or to write a PhD thesis on a computer, we need to interact with objects where necessary. Unfortunately, for people suffering from a neuromuscular disorder (NMD) this way of life is restricted as the disorder causes muscular weakness and consequently limits movement. NMDs affect about 153 people per 100.000 in the Netherlands, and 160 per 100.000 worldwide [1]. Hereditary disorders such as muscular dystrophy often affect people's life at an early age. Furthermore, most NMDs, such as muscular dystrophy, are progressive, meaning that muscular weakness worsens over time and so does the performance during daily life activities (ADL) [2].

These disorder characteristics can be described using the International Classification of Functioning, Disability, and Health (ICF) model [3] (Figure 1), which consists of three components; body functions and structure, activity, and participation, and two contextual factors; environmental and personal. Problems and difficulties identified in each component are indicated as impairments, limitations, and restrictions, respectively. The most common impairments in NMDs involve a reduction of muscle quality, such as loss of muscle fibres and fat infiltration, and of upper extremity mobility and stability, thus limiting the movement capabilities of the upper extremity in terms of strength and range of motion. As a result, motor control appears to be altered and compensatory movements from neighbouring joints are required to perform ADL tasks. Yet, it remains unclear how motor control is altered in NMDs. The consequence is that the performance of ADL is limited and participation in a social environment is restricted [4, 5].

Assistive devices can alleviate some of the upper extremity limitations and thereby assist in the activities and participation of daily life. Dynamic arm supports (DAS) are assistive devices that provide gravity compensation on the lower arm to relieve upper body function impairments [4]. A dynamic arm support reduces the required muscle efforts and enhances upper extremity mobility, improving the capabilities to interact with the environment. However, over time most users stop using the assistive device altogether for reasons that remain unclear [6]. Environmental and personal factors (e.g. costs and aesthetics) are thought to largely contribute to this discontinued use [6]. Our hypothesis is that the progressive nature of NMDs and its associated loss of function is one of the major determinants of discontinued use of arm supports as users eventually lack the strength and mobility to handle the device's inertia in addition to overcoming their own lack of mobility [6-8]. Strength and mobility loss and motor control alterations are the forerunners of limitations and restrictions in daily life experienced by subjects suffering from muscular dystrophy

[9, 10]. Personalized arm support settings could be sufficient to restore mobility in the short term. However, compensatory movements may be preferred over a dynamic arm support in the long term if its limitations, e.g. difficulty in controlling the device due to excessive upper support force or kinematic limitations of the device, outweigh its benefits. Yet, it remains unclear how motor control is altered in NMDs by the use of a dynamic arm support. Moreover, it is also unclear how these alterations are affected by the progressive nature of NMDs. Therefore, this thesis aims to better understand how people with NMDs interact with such assistive devices.

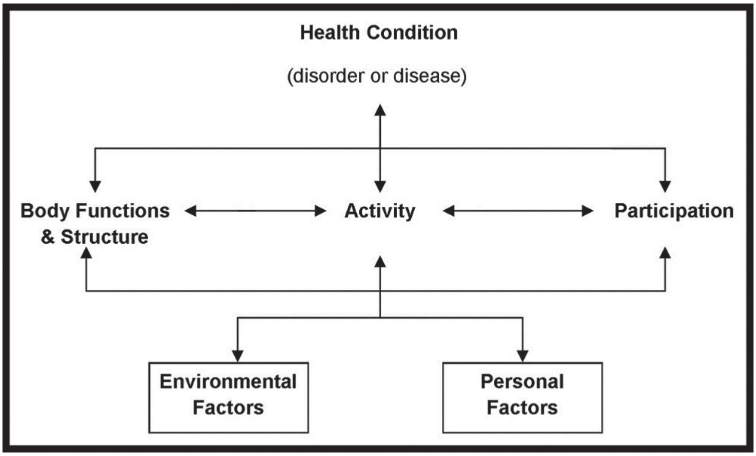

Figure 1. International Classification of Functioning, Disability and Health model.

1.1 DYNAMIC ARM SUPPORTS

Numerous dynamic arm support devices have been developed in the past 80 years for various populations, including polio patients and stroke survivors, and people with NMDs [5]. Generally, there are three types of support devices implemented in these devices: 1) passive through counterweights or elastic bands, 2) active through an electromotor, or 3) hybrid, which provides a combination of passive and active support. One such hybrid device is the Gowing [11] (Figure 2), which is designed to facilitate anti-gravity and hand to mouth movements via support on the lower arm, elbow, and wrist (optional). While these devices are commonly well-received by its users, there are some issues that may contribute to its discontinued use. For example, the devices' mobility is often limited in the side- and downward directions, forcing the user to reconsider the benefit of having increased upward mobility at reduced efforts against the loss of overall range of motion and opposing force when moving downwards [7, 12]. Furthermore, as with many types of assistive devices, the DAS device enlarges the arm's dimensions and inertia with a resulting increase of collisions with the environment (e.g. hitting the kitchen table while eating/drinking). Therefore, interaction with a device requires adjustments to how users move. These user-device interactions are best studied by looking at the neuromechanical adaptations of the user, as it has been shown in other studies on the design, actuation, and control of assistive devices [13-15]. For example, a dynamic arm support alters muscle efforts and joint kinematics during reaching, which indicates that the respective coordination and movements have been adapted accordingly.

Figure 2. The dynamic arm support Gowing from Focal Meditech B.V.

1.2 NEUROMECHANICS

Neuromechanics is described as a field that seeks to understand how the neurophysiological and biomechanical aspects, such as muscles, sense organs, motor pattern generators, and brain, interact to produce coordinated movement [16]. An important aspect of neuromechanical adaptations are the changes in motor control and output. Motor control is defined as the way the central nervous system coordinates muscle activity to deal with the redundancy of degrees of freedom in joints and muscles present in the human body [17]. Motor output can be described by the forces and motion, or kinetics and kinematics, respectively, that the body produces. In NMDs, it is known that motor control and its resulting output are generally affected by impairments in either the nervous system, musculoskeletal system, or both. In contrast with other neurological diseases, the nervous system in muscular dystrophy is presumed to be least affected. Thus, musculoskeletal impairments such as muscular weakness and decreased mobility are expected to have the largest influence, especially over time. However, in muscular dystrophy, and particularly Facioscapulohumeral dystrophy (FSHD) with a prevalence of 4 per 100.000 in the Netherlands [18], these alterations in motor control and output remain mostly unclear. In particular, how shoulder muscles are coordinated during activities of daily life, and whether the presence of an arm support changes this coordination, remain to be investigated.

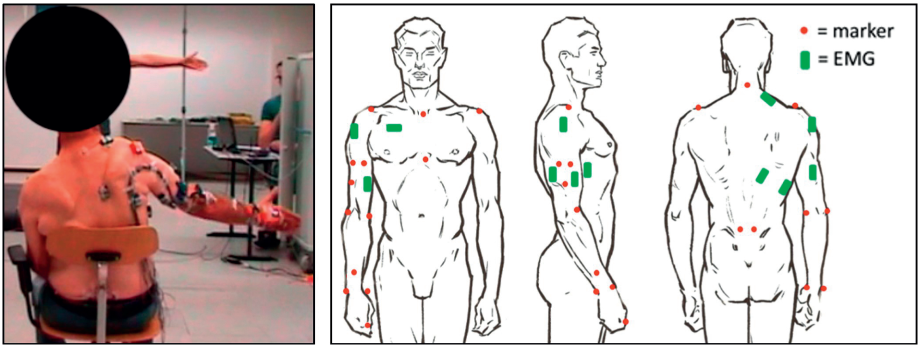

Figure 3. Surface electromyogram (EMG) electrodes on a participant (left) and complete overview including reflective markers for 3D motion capture (right).

Coordinated muscle activities, as part of motor control, can be investigated with surface electromyography and non-negative matrix factorization that provide one or more sets of muscle synergies as a mathematical approximation. Surface electromyography is a common method to quantify muscle activity by measuring the electrical current of the skin on top of a muscle (Figure 3) [19]. Non-negative matrix factorization is a method that simplifies the activation of multiple muscles into a lower dimensional spatiotemporal output of synergistic contributions (weights) and activation patterns (coefficients) [17] (Figure 4). The variability and alterations in motor control is then evaluated based on the number of muscle synergies needed, the variances accounted for per synergy, and synergy similarities. In people post-stroke, muscle synergy analysis of the upper extremity revealed alterations during isometric force generation [20] and dynamic tasks [21]. Furthermore, in various daily activities, the affected and unaffected arm muscle synergies were highly similar and showed the presence of compensatory strategies by Trapezius and Pectoralis muscles during reaching tasks [20, 22, 23]. Concerning dynamic arm support devices, the muscle coordination of healthy (older) participants was influenced slightly regardless of the level of support [13, 15, 24]. However, it can be postulated that for people with FSHD such a device would alter the selection of muscle synergies in a more distinct way than in healthy individuals. For example, due to the muscle weakness of the arm adductors, overcoming the gravity compensation could require altered synergies compared to healthy individuals. However, in people with FSHD, muscle synergies of daily life activities and alterations therein from using a support device with respect to healthy individuals remain unclear.

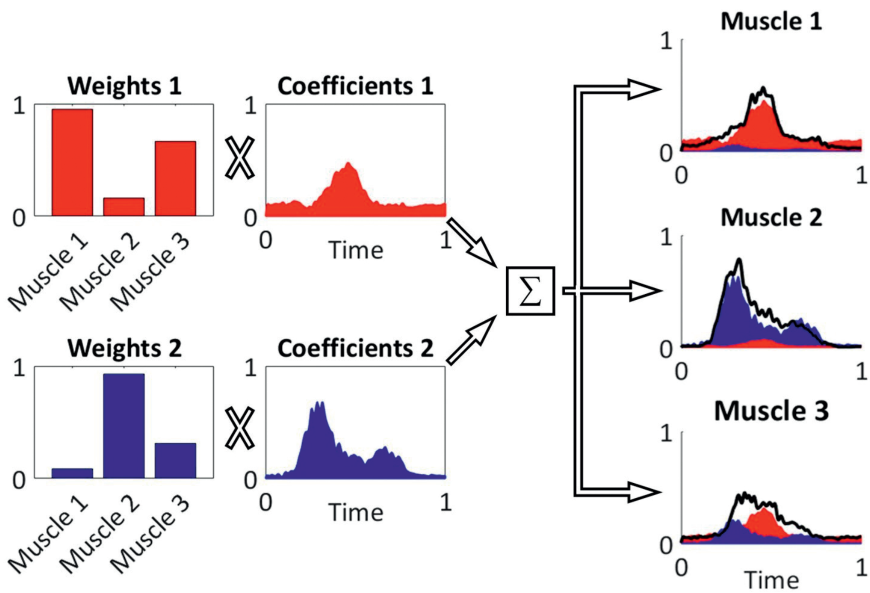

Figure 4. Example overview of EMG reconstruction by the sum of activations of muscle synergies. The total activity of individual muscles (right, black line) is the result from summation of muscle synergies (blue and red) based on respective spatial contributions (left, weights) for shared temporal patterns (left, coefficients).

Kinetics and kinematics, as part of motor output, reflect the strength and mobility of the user, respectively. Kinetics are measured with force sensors [25] and often used to describe strength of muscle groups and interaction with the environment. Kinematics are commonly recorded by three directional motion capture (3DMoCap) systems in controlled environments and activity monitors in a home environment.

3DMoCap systems capture the motion of markers placed on specific location on the body (Figure 3). However, the controlled environment in which 3DMoCap systems are used is limited in replicating the daily life situations. Therefore, activity monitors (Figure 5) [26, 27] are used to quantify motions, such as intensity and orientation, and to estimate efforts during daily life activities at home. This approach has been successfully applied in various populations such as children with neurodevelopmental disorders, people post-stroke, and upper limb prosthesis users [28-32]. Previously [28], we used a multi-sensor network to classify upper and lower arm activities of children with Duchenne Muscular Dystrophy during ADL. These classified activities, intensity, orientation, and frequency of arm elevations provided valuable insights into the daily activity levels, such as the timing, intensity, and duration of activities. Using this novel approach in the shape of a wearable sensors network, the benefits of using an arm support can similarly be monitored in the users' daily environment, which provides a better understanding of discontinued use with respect to disease progression.

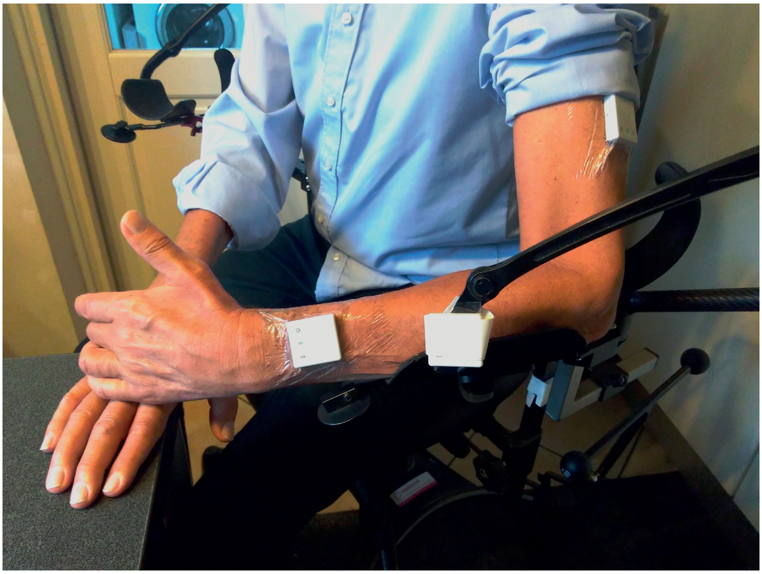

Figure 5. Measurement setup demonstration on a support device user wearing the MOX on the upper arm, lower arm, support brace, and device base (not shown in figure).

1.3 THESIS AIMS AND OBJECTIVES

The aim of this thesis is to better understand how people with NMDs, in particular those with musculoskeletal impairments, interact with dynamic arm supports to establish the contributing factors for discontinued use. We first approached this aim by addressing current literature and expert opinions (Chapter 2). The goal was to create a structured overview of current knowledge and a framework to facilitate future research. Second, we investigated the effect of specific impairments and limitations (Chapter 3) of FSHD, a common muscular dystrophy type, on motor control in a controlled setting compared with healthy individuals. The revealed effect would provide a better understanding of the interactions between body functions and activity and participation and consequently guide disease management. We hypothesized that motor control shows alterations in terms of muscle synergies. Third, we investigated the neuromechanical changes with a dynamic arm support (Chapter 4) during activities of daily life. Similarly, the objective was to get a better understanding of the ICF components' interactions, but now in combination with the user-device interaction. Furthermore, the focus on activities of daily life would approach reoccurring situations of usage at home. Our hypothesis was that motor control under influence of a dynamic arm support would show less alterations in terms of muscle synergies, but would be influenced by the type of activity performed. Fourth, we expanded the user-device interaction measurement to a home environment to examine usage and performance in daily life (Chapter 5). In addition, the goals were to capture determinants for device benefits and discontinued use and to test the design's long-term practicality in an uncontrolled environment. Our hypothesis was that device benefits were mostly related to motions against gravity and discontinuation of use would be gradually present. Finally, a discussion (Chapter 6), valorisation (Chapter 7), and a summary (Chapter 8) were formulated to emphasize the implications for arm support users, developers, clinicians, and researchers. Ultimately, we aim to promote collaboration across expert fields to enhance dynamic arm support usage through a better understanding of the disease and the user-device interaction.

CHAPTER 2

Recommendations for studies on dynamic arm support devices in people with neuromuscular disorders: a scoping review with expert-based discussion

J. M. N. Essers, A. Murgia, A. Peters, M. M. H. P. Janssen, K. Meijer.

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2.1 INTRODUCTION

Neuromuscular disorders (NMD) are characterized by muscle weakness that limits upper extremity mobility and can affect people of all ages [33]. The worldwide prevalence of NMD is 160 per 100,000, which is similar to that of Parkinson's disease and double that of multiple sclerosis [1]. People with NMD commonly experience difficulties to perform movements against gravity, such as lifting the arms or reaching for an object [34]. These functional limitations translate into problems with activities of daily life (ADL) and participation in society. Various types of dynamic arm support devices (DAS) have been developed that improve engagement in ADL by providing gravity compensation [5, 6]. Generally, such devices relieve upper extremity limitations that stem from muscular weakness [14, 15, 35]. However, the impact might differ between devices intended for rehabilitation and research and wearable devices intended for daily life. As approximately 7 to 24% of the Dutch population with NMD, and 8.5% of people with Duchenne Muscular Dystrophy (DMD) worldwide [36], use a DAS in daily life [8], it is relevant to understand how and for what purposes the DAS are used on a daily basis.

The perceived benefits of a DAS vary from complete satisfaction to no perceived added value, where daily life usage has been reported to be discontinued over time [7, 37, 38]. Current research highlights the importance of evidence-based recommendations for DAS development and prescription. However, determining recommendations has proven to be difficult [6], due to the diversity in NMD, DAS, and study designs. This is further complicated by the lack of standardized and validated evaluation tools in current research [6, 38, 39]. To optimize DAS development and its use in daily life, it is important to investigate the users' characteristics, DAS function, and resulting user-device interactions in the short and long term. According to the Consortium for Assistive Technologies Outcome Research framework, used by Heide et al. 2015, the WHO's International Classification of Functioning, Disability and Health (ICF) model could act as the primary guidelines [3, 6, 40].

Current research indicates that DAS impact the users abilities across all ICF components [6]. However, it is commonly assumed that a DAS primarily affects body functions through gravity compensation, which shapes the effects on activity and participation. Furthermore, Heide et al. 2015, indicated that the ability of a DAS to support ADL does not guarantee higher performance or even utilization in a home environment [6]. Therefore, it is important to account for the contextual differences in which the activities are performed. For example, there is a difference in what people are able to do in a standardized environment by following instructions of an examiner compared to what they actually do in their daily life. Holsbeeke et al. 2009 described three constructs, or concepts, for these contextual differences: motor capacity (can do in a standardized environment), motor capability (can do in a daily environment), and motor performance (actually do in a daily environment) [41]. We propose that a combination of the ICF model as primary guidelines and the three contextual constructs as secondary guidelines would provide a suitable framework to structure the evidence of DAS evaluation and recommendations for future development.

Technological advances in DAS, such as wearable robotics, are developing rapidly and it is expected that they will become more pervasive in daily life support systems [8, 36]. Yet the research on DAS evaluation in patients is relatively new and under development [6, 39]. Perspectives on the state of the art from third parties who are either engaged in development and prescription of these devices, or are end users, could provide important insights which are often lacking in the literature. The current study aims to synthesize the literature with expert opinions in order to provide an overview of current evidence and identify knowledge gaps that may limit the development of DAS. A secondary aim is to provide research recommendations to establish a standardized and validated approach for DAS evaluation in people with NMD.

2.2 METHODS

2.2.1 Literature search

Inclusion criteria for the literature review focussed on studies that evaluated a DAS intended for daily life situations that supported the lower arm through gravity compensation. Studies with healthy participants only were included if the DAS was a finalized prototype designed for daily use by people with NMD. Furthermore, studies needed to report measures involving body functions described as neuromusculoskeletal and movement-related functions [3]. Other inclusion and exclusion criteria can be found in the appendix table 1. Both scientific and non-commercially published literature were searched.

Scientific literature was searched in PubMed and Web of Science in August 2018 and updated in July 2020. The search strategy for each database can be found in the appendix table 1. Titles and abstracts were independently screened by two researchers where remaining articles were compared and finally in-/excluded.

Included articles' authors and reference lists were then searched for additional articles.
Non-commercially literature published in the past 5 years (2015-2020) was searched from a government clinical trial database (clinicaltrials.gov), DAS suppliers' websites identified from previously included articles, and research mentioned by the experts. The search strategy for the government clinical trial source consisted of the combination of 'Neuromuscular Disorders' or 'Neuromuscular Diseases' with either 'Robot', 'Exoskeleton', or 'Arm support'. One researcher gathered the information and another researcher checked for consistency with the in-/exclusion criteria. Any inconsistency was discussed until agreement was reached.

2.2.2 Focus groups

Five focus groups were formed from a patient community, DAS developer, clinical, rehabilitation, and research setting. The groups consisted out of fifteen experts in total: two members of a patient community with a neuromuscular disorder, five DAS developers/suppliers, one physician, five therapists, and two researchers. At that time, one member with a NMD used a DAS in daily life and the other was orientating. The experts were most experienced with muscular weakness from atrophy/dystrophy or lesions to the central nervous system. Two developers/suppliers also had direct contact with some clients. Eight people were involved in research: two full-time researchers, one member with a NMD evaluated research proposals in a funding comity, and two developers, two therapists, and one physician were partially involved in projects involving DAS development, training, and improvement of DAS selection procedure.

The groups were interviewed by three researchers in person on separate occasions. The interviews were in Dutch and semi-structured as they were guided by (originally Dutch) questions. The questions were formed based on preliminary review of the literature and discussion between the research team. Information was distilled of the focus groups' views on 1) current impairments/limitations of people with NMD, 2) effectiveness and requirements of a DAS linked to the previous, 3) previous research projects and remaining questions, and 4) research priorities. The interviews were audiotaped in support of the keywords, which were noted during the interview by at least two researchers and evaluated afterwards. Keywords were formulated based on the terminology used in previous literature and descriptions were added for clarification purposes.

2.2.3 Evidence synthesis

Fundamental topics from literature and interviews were tabulated and summarized to synthesise current evidence with expert opinions. The tabulation framework was formed with the ICF model components (body functions, activity and participation, and environmental and personal factors) as rows and contextual constructs (motor capacity, - capability, and - performance) as columns. Furthermore, ICF model components were further divided as categories and sub-categories to cover the different movement and impairment aspects. Each table cell represents a unique combination of one sub-category and one contextual construct. Current evidence matching expert opinions were indicated in each table cell by the reference number. Cells where current evidence was lacking, despite experts' interests, were considered knowledge gaps and indicated by a dash symbol. We formulated our synthesized current evidence, knowledge gaps, and research recommendations according to the terminology used for conceptual modelling of assistive technology device outcomes [40].

2.3 RESULTS

2.3.1 Synthesized evidence

Current evidence and knowledge gaps were synthesized per category based on literature findings and expert opinions and presented in table 3. There were five categories identified for body functions, three for activity and participation, and one for environmental and personal factors. The user-device interaction was considered handling an object and therefore categorised under activity and participation. Body functions was consequently linked to user-device interaction as a sub-category. Literature covered roughly nine out of 51 cells within the body functions component, which also covered eight out of 19 cells within the activity and participation component, and two out of four cells within the environmental and personal factors.

2.3.2 Literature review

The literature search resulted in 635 hits (PubMed: 209, Web of Science: 426) of which 587 unique articles (figure 1). Then, 546 articles were excluded based on title and abstract, and another 35 after a subsequent full read of the article. Two articles were added after reference cross checks. Finally, eight articles were included for reviewing after the inclusion process [12, 42-48] (table 1). Two studies, ongoing until December 31st 2020 and 2019, respectively, were identified from the government clinical trial source [49, 50] (table 2).

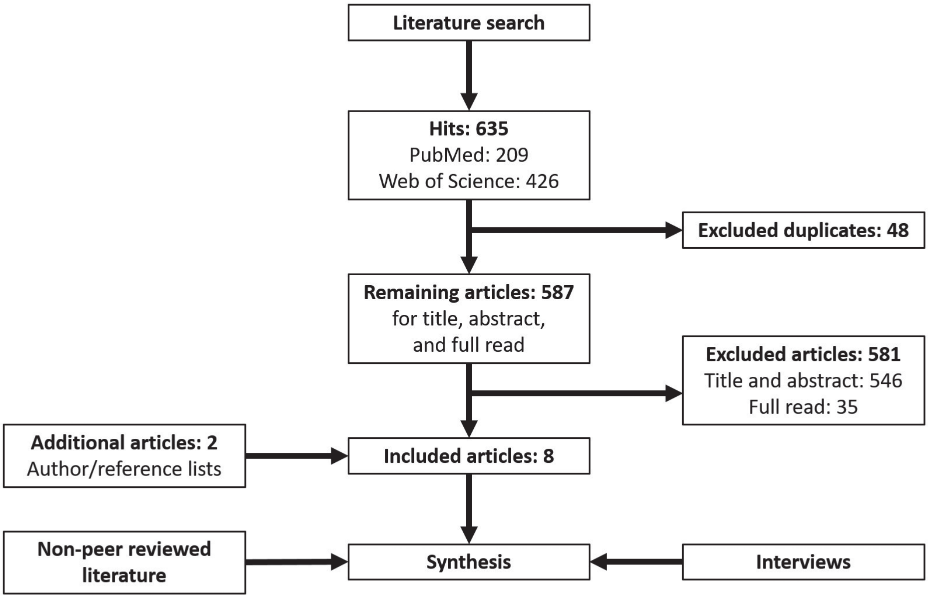

Figure 1. Flowchart of mixed-methods process.

2.3.3 Dynamic arm support evaluations

The evaluated DAS were a prototype A-gear [43, 51], and commercially available devices Armon Ayura [49, 50, 52], Armon Edero [48, 52], Gowing [47, 53], JAECO MultiLink Arm with Elevation Assist [48, 54], JAECO WREX [45, 46, 49, 50, 54], SLING [12, 44, 53], and Top-Help [12, 53], and a JAECO WREX modified with a trunk support prototype [42] (table 1). These DAS provide gravity compensation through adjustable counter-weights (SLING) or springs (A-gear, Armon Edero, JAECO MultiLink Arm with Elevation Assist, JAECO WREX, Top-Help) with an additional actuator that adjusts the springs' tension (Armon Ayura, Gowing). All studies investigated the body functions, several also investigated activity and/or participation [12, 49, 50], and one the satisfactory levels of using the DAS [47]. All study designs were set up to cross-sectionally investigate motor capacity during functional and ADL tasks w/o a DAS. Effects were mostly quantified for muscle function, joint mobility, and upper body functionality. Main effect identified were: DAS can lower up to 15% activity of the Trapezius muscle in healthy people; DAS can increase arm elevation from 25 up to 100° and elbow flexion from 10 up to 120° in people with NMD; Performance of the Upper Limb (PUL) scores were improved by 20 up to 30% with a device in people with NMD and showed a large effect (>1.15) in the modified Upper Extremity Performance Test for the Elderly (TEMPA); and the ability to perform ADL w/o a support device was participant specific where a device could have a limiting and beneficial effect. The ongoing studies perform longitudinal interventions of eight and five weeks, respectively, to field-test two DAS [49, 50] (table 2). We considered these studies to investigate the motor capacity, - capability and - performance regarding upper body functionality and activity/participation levels.

2.3.4 Expert opinions

Experts expressed their opinions mainly in identifying DAS effects on muscle function, joint mobility, and upper body functionality (table 3 and appendix table 1). They viewed people with NMD as impaired in these aspects of body functions and believed a DAS could compensate specific impairments. For example, a DAS improves joint mobility and lowers muscle efforts. This combination allows a user to perform multiple repetitions in a greater workspace needed in self-care activities such as eating and drinking. However, a DAS often does not fully support the complete hand to mouth motion and is then considered limited in their effectiveness. Therefore, experts believe further research is necessary to support DAS development not only on body functions but on all ICF components and contextual constructs. Furthermore, experts advocated to expand and improve the descriptions of the (active) population and disease progression. Specifically, device utilization in the daily environment was of interest to bridge the knowledge from motor capacity to motor capability and motor performance.

Table 1. Study characteristics of completed studies. SD: Study Design, P: Population,

Reference	Research aim	Study design, population, DAS, and tasks
Dunning (2016)	To explore the effect of the JAECO WREX with trunk motion capability on the total range of motion and balancing quality of the arm support.	SD: Cross-sectional P: Healthy (N=3) DAS: JAECO WREX with prototype trunk motion capability T: Functional task and ADL (single joint motions, reaching and touching various places, and drinking w/o DAS)
Essers (2013)	To compare the influence of gravity compensation between joint moments in populations performing a reaching task.	SD: Cross-sectional P: Muscular Dystrophy (Facioscapulohumeral Dystrophy) (N=3) and Healthy (N=3) DAS: SLING T: Functional tasks (reaching at shoulder level w/o DAS)
Estilow (2018)	To evaluate the efficacy of the JAECO WREX on improvement of active range of motion of the upper extremity and ADL performance.	SD: Cross-sectional P: Muscular Dystrophy (Duchenne) (N=9) DAS: JAECO WREX T: Functional task and ADL (active shoulder motions, eating/drinking, simulated facial grooming, and item retrieval w/o DAS)
Haumont (2011)	To describe the JAECO WREX and to assess the functional improvement in the upper extremity.	SD: Cross-sectional P: Arthrogryposis Multiplex Congenita (N=1) and Spinal Muscular Atrophy (N=2) DAS: JAECO WREX T: Functional tasks and ADL (shoulder motions, reaching, and eating/drinking w/o DAS)
Heide (2017)	To evaluate the Motion and Muscle Ambulatory Activity System at a home setting and investigate range of motion and ADL performance w/o DAS.	SD: Cross-sectional P: Stroke (N=2), amyothophic lateral sclerosis (N=1), muscular dystrophy (N=1), and spinal stenosis (N=1) DAS: Top-help and SLING T: ADL (reaching and eating/drinking w/o DAS)

N: number of participants, T: Tasks, DAS: Dynamic Arm Support, w/o: with and without.

Outcome parameters and interpretations	**Research Recommendations**
- Range of Motion (kinematics) - Normalized muscle activity levels Range of motion increased by 10% and less muscle activity is needed with support.	Use the same degrees of freedom in the DAS as a human arm which allows for better alignment and natural behavior. The effect of a body-bound DAS with trunk motion capability was perceived as important by both healthy subjects and three DMD patients who shortly tried the DAS without measurements. Movements were easier and more natural to perform and DMD patients mentioned that it also allowed for making compensatory movements.
- Joint moments (kinetics) - Execution time (kinematics) Joint moments were lower with DAS but execution time was longer.	Designing better support devices should include biomechanical considerations. Further research should be focused on expanding the investigation on the influence of gravity compensation in muscular dystrophy subjects in order to quantify the real benefits of an arm support device.
- Active Range of Motion (kinematics) - Manual muscle strength testing Improvements in active range of motion of shoulder and elbow joints depended on the patients' strength.	Further exploration of agonist and antagonist strength ratios is needed with respect to the muscle groups that receive assistence. Longitudinal studies are necessary to evaluate the WREX's regarding muscle strength and ARange of motion preservation. Future research should include the use of diagnosis-specific measures and an ordinal ADL scale. In addition, research is needed to identify clinical measures that predict successful performance with the WREX to allow clinicians to appropriately evaluate and recommend the WREX for patients with neuromuscular diseases.
- Range of Motion (kinematics) There was greater mobility of the upper extremity in motions against gravity with the DAS.	Future DAS should assist in pronation or supination as these motions are an integral part of the feeding movement.
- Range of Motion (kinematics) Shoulder range of motion increased with DAS but varied greatly between individuals.	To understand the real effects of these devices it is recommended to take this individual variation into account and that benefits are also assessed in a broader daily life context. Insight into benefits of DAS and influencing factors could support the provision and the selection process of DAS.

Table 1. (Continued)

Reference	Research aim	Study design, population, DAS, and tasks
Kooren (2015)	To develop, pilot test, and present the characterization and validation of the A-gear.	SD: Cross-sectional P: Duchenne (N=3) and Healthy (N=1) DAS: A-gear (prototype) T: Functional tasks and ADL (shoulder and elbow motions, eating/ drinking, use phone/computer, self-care, and PUL w/o DAS)
Lebrasseur (2019)	To evaluate the usability of the Gowing power-assisted arm support.	SD: Cross-sectional P: Multiple sclerosis Spinal (N=3), muscular atrophy (N=3), and muscular dystrophy (N=3) DAS: Gowing T: Functional tasks and ADL w/o DAS
Cruz (2020)	To examine Armon Edero and Multilink with Elevation Assist on upper limb function and ADL.	SD: Cross-sectional P: Muscular dystrophy (Duchenne) (N=4) DAS: Armon Edero and Multilink with Elevation Assist T: Functional tasks and ADL w/o DAS

Outcome parameters and interpretations	Research Recommendations
- Range of Motion (kinematics), - Distance covered (kinematics), - Comfort (self-reported) PUL score was increased with less compensatory movements. Arm movements forward and upward became easier. The DAS was comfortable.	One-hundred percent weight compensation is not always preferred by patients. One of the patients wanted less supporting force, which felt more comfortable to him. The reduction of compensatory movements is very important, as compensatory movement consumes a lot of energy and therefore they restrict the endurance to perform daily activities. Two sided support is preferred to avoid a skew posture. Forward lean capability is much appreciated. DAS preferably does not run between arm and trunk, or add considerable volume underneath the forearm and elbow. Such components create an uncomfortable environment for arm relaxation and can clash with tabletops.
- Disabilities of the Arm, Shoulder and Hand (DASH) - modified Upper Extremity Performance Test for the Elderly (TEMPA) - Quebec User Evaluation of Satisfaction with assistive Technology (QUEST) The DAS improved TEMPA scores and two third were quite or very satisfied with the device.	Different actively actuated devices could be compared in manual tasks and in the long-term to assess long-term benefits. Future designs should include hand exoskeleton or wrist support on a DAS to investigate the influence of fine movement improvements. Proper use of DAS in research might depend on correct installation and should receive attention in the experimental setup.
- PUL - Duchenne Muscular Dystrophy Upper Limb function Patient Reported Outcome Measure (DMD UL PROM) - Semi-structured interviews PUL mid-level score was increased, but distal level score decreased with DAS. Mostly eating and drinking was enhanced. Decline in strength was the main reason for discontinuation.	ROM and standardized assessment measures do not necessarily predict the degree of functional improvement possible with DAS use in real life settings. Both objective and subjective outcome measures should be used when evaluating the effectiveness of a DAS. Input may also be required for ongoing support from a clinician and/or supplier experienced in DAS fitting and adjustment, to ensure issues can be addressed in a timely manner as the disease progresses. Future DAS designs should focus on independent stabilization of the arm when it is positioned in the DAS and improve sensitivity between DAS tension levels. Furthermore, a DAS should fit within the existing wheelchair dimensions and allow the DAS to be secured in a fixed position when not in use to offer increased device utility and improve user experience.

Table 2. Study characteristics of ongoing studies. DMD: Duchenne Muscular Dystrophy,

Reference	**Research title and aim**	**Study design, population, DAS, and tasks**
Bendixen (2019)	Title: Use of Dynamic Arm Support Devices for Upper Limb Function in Non-Ambulatory Men With DMD Aim: To promote participation in activities of daily living in non-ambulatory individuals with DMD with upper extremity weakness.	SD: Longitudinal, randomized, and interventional P: Duchenne Muscular Dystrophy DAS: Armon Ayura and JAECO WREX T: - Baseline (two weeks) - Device trial (four weeks) - Post device trial (two weeks)
Pedrocchi (2019)	Title: USEFUL: User-centred Assistive SystEm for Arm Functions in neUromuscuLar Subjects Aim: To field-test the improvement in arm functions provided by DAS and assessing their impact to users' quality of life and independence	SD: Longitudinal, cross-over, and interventional P: Muscular Dystrophies (Duchenne, Becker, Limb-Girdle Type 2) DAS: Armon Ayura and JAECO WREX T: - Baseline (cross-sectional) - Short Training DAS 1 (three days) - Domestic use DAS 1 (two weeks) - Short Training DAS 2 (three days) - Domestic use DAS2 (two weeks)

SD: Study Design, P: Population, T: Tasks, DAS: Dynamic Arm Support.

Outcome parameters	**Research relevance**
Primary: - Change in Upper Extremity Acceleration through Actigraphy - Change in Upper Extremity Position through Actigraphy Secondary: - Goal Attainment Scale - Physical Motor Assessment - PUL	Upper extremity performance will be further quantified with use of a physical motor assessment, the PUL assessment, and patient reported outcomes. Data gleaned will provide important knowledge and objective results regarding the potential benefit of DAS in individuals with DMD with limited functional use of their upper extremities.
Primary: - PUL Secondary: - Motor Function Measures scale - Brooke scale - ABILHAND - PedsQL - PROMIS FATIGUE - Personal Adjustment and Role Skills Scale III - Technology Acceptance Model - System Usability Scale	Field-testing is essential to assure a widespread accessibility to these devices for most of the potential users, possibly providing health providers with direction and guidance towards Health Technology Assessment.

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Table 3. Tabulation and summary of literature (L) and expert views (E) and synthesis of research recommendations. Current evidence, or knowledge gaps, within experts' interests were indicated in each cell by the reference number(s) or a dash, respectively. Blank cells either lacked experts' interests or were considered theoretically impossible.

ICF component	Category	Sub-category	Motor Capacity	Motor Capability	Motor Performance
Body function	Muscle	Activity	1	—	—
		Coordination	—	—	—
		Fatigue	—	—	10
		Pain	—	—	—
		Stiffness	—	—	—
		Strength	4	—	—
	Joint	Kinematics	1, 2, 4, 5, 6	—	—
		Kinetics	3	—	—
		Pain	—	—	—
		Stiffness	—	—	—
	Upper body	Functionality	7, 8, 9, 10	—	—
		Hand/Wrist function	—	—	—
		Motion/Posture	2	—	9
Body function	Population description	Disease progression	—	—	—

NMD: NeuroMuscular Disorders, ADL: Activities of Daily Living, DAS: Dynamic Arm Support, w/o: with and without. 1. (Dunning et al., 2016) 2. (Kooren et al., 2015), 3. (Essers et al., 2013), 4. (Estilow et al., 2018), 5. (van der Heide et al., 2017), 6. (Haumont et al., 2011), 7. (Lebrasseur et al., 2019) , 8. (Cruz, 2020), 9. (Bendixen, 2019), 10. (Pedrocchi, 2019).

Expert views (E) and literature findings (L)	**Synthesized research recommendations**
E: for which muscles can a DAS reduce muscle activity? L: activity was reduced in the Trapezius, Deltoid, and Pectoralis Major muscles. E: does a DAS require compensatory strategies? E: how can we measure and prevent fatigue? L: no findings available from ongoing studies. E: can a DAS relief muscle pain? E: can a DAS reduce muscle stiffness? E: what are required strengths to operate a DAS? L: potential DAS effects are related to residual muscle groups' strength.	Explore the effects of external force generated by DAS on muscle function of the musculoskeletal system. Determine whether an external force introduces a burden on other muscles or if it reliefs pain and stiffness. Investigate the relation between muscle effort reduction and muscle coordination of the shoulder girdle. Prescribe user strength requirements to a device.
E: can we map the differences in motion w/o DAS? L: a DAS positively affects the reachable workspace. E: can we map the interacting forces for motions with a DAS? L: gravity compensation reduces joint torques and thus effort. E: can a DAS relief joint pain? E: can a DAS overcome joint rigidity and feelings of stiffness?	Explore the effects of an external force generated by DAS on the joints of the musculoskeletal system. Investigate the relation between improvements in joint load reduction and mobility and joint stability, pain, and stiffness. Investigate the relationship between levels of external force, joint loading, and mobility across the range of motion.
E: can we distinguish different aspects of upper body functionality improvements? L: the DAS had a large positive effect on most items. No findings available from ongoing studies. E: what is the influence of a DAS on hand/wrist function? E: how can we support natural motion/support? L: an integrated DAS with trunk support positively affected participants' posture. No findings available from ongoing studies.	Investigate the effects of a DAS on the upper body as a whole by including other regions than the shoulder area. Explore what can be improved by integrating devices that support other regions than the shoulder area.
E: can we expand knowledge of the (active) population on muscle/joint/upper body aspects?	Limited knowledge on disease progression is presented on a population level for DAS users. Perform a review on disease progression in NMD on functionality and disability.

Table 3. (Continued)

ICF component	Category	Sub-category	Motor Capacity	Motor Capability	Motor Performance
	Others	Short term tests	—	—	—
		Long term tests	—	—	9, 10
		Test accuracy and discriminative properties	—	—	—
Activity & Participation	User-device interaction	Body function aspects	1-10	—	8, 9, 10
		Functionality	5, 7, 8, 9, 10	—	8, 10
		Quantified device utilization			—
		Quality of performance			9, 10
Activity & Participation	Population description	Needs and goals		—	9, 10
		Functionality	5, 7, 8, 9, 10	—	8, 9, 10
	Others	Ambulant/ Body mounted	—	—	—
		Test accuracy and discriminative properties	—	—	—

Expert views (E) and literature findings (L)	Synthesized research recommendations
E: current measurements are too time-consuming for participants. E: cross-sectional studies are not representative for disease progression. L: no findings available from ongoing studies. E: current methods do not capture disease progression properly enough.	Performance tests are considered too long due to the rapid onset of fatigue within this population. Furthermore, disease progression or DAS effects are not captured to a sufficiently distinctive level in a cross-sectional design. Develop tests which require less from the population and can be applied on multiple occasions.
E: what are the effects of a DAS on body function aspects and how do they interact? L: a DAS reduces efforts and increases reachable workspace, depending on residual muscle strength. No findings available from ongoing studies. E: what types of ADL are possible, impossible, or limited w/o a DAS? L: the DAS positively affected the collectively measured ADL. No findings available from ongoing studies. E: can we quantify device utilization and recognize non-usage? E: can we monitor performance of ADL at home? L: no findings available from ongoing studies.	Research should combine multiple aspects to investigate interactions within body functions. Relate the performance of ADL to body functions and identify areas for device development. Identify non-usage in daily life as an indicator for disease progression. Monitor performance and identify negative adaptations.
E: what are the needs and goals of people with NMD? L: no findings available from ongoing studies. E: what types of ADL are people with NMD capable of and do they perform those at home? L: observations revealed participant-specific possibilities for ADL. No findings available from ongoing studies.	Describe the population so it connects the desired motor capabilities and performance to the capacity. Integrate aspects of personal and environmental factors.
E: what is the influence of being ambulant and/or wearing a body-mounted DAS on device utilization? E: current methods are limited in capturing user-device interaction properly.	User-device interaction evaluations should be applicable to a large variety of users, yet discriminative. It is suggested to have a flexible and tailored method, such as wearable sensors, to measure continuously in varying settings.

Table 3. (Continued)

ICF component	Category	Sub-category	Motor Capacity	Motor Capability	Motor Performance
Environmental and Personal factors		Home setting and outdoors		—	8
		Psychological drives			—
		Distribution of energy			8

2.4. DISCUSSION

2.4.1 Current evidence and knowledge gaps

The aim of this scoping review was to provide research recommendations for DAS evaluation based on a synthesis of literature and expert opinions. We primarily focused on body functions and secondarily on other ICF model components: activity and participation, and environmental and personal factors. To structure the evidence and identify gaps we used a framework that combined the ICF model components with contextual constructs: motor capacity, capability, and performance. Most included studies focused on the user-device interaction within the framework cells of body functions and motor capacity. Typically, they studied the introductory phase of using a DAS, with just a few studies addressing the long-term adaptations. The lack of standardized evaluation tools posed difficulties in creating comparable evidence [47] and the synthesis of current evidence [6]. The following knowledge gaps were identified: first, we poorly understand the adaptations that may ensue following skill acquisition, fatigue, or disease, which alter the support requirements over time. Second, it is yet unclear how abilities across ICF components and contextual constructs are related. For instance, it is unclear how changes in body functions influence the activity and participation and how these changes are, in turn, the result of environmental setting and task requirements. Finally, various aspects, such as comfort levels, were considered important by the experts that have not yet gained sufficient attention in the scientific literature.

Expert views (E) and literature findings (L)	Synthesized research recommendations
E: what are factors in the user's environment that influence the user-device interaction? L: perceived facilitators were related to independence stimulated by internal and external motivation. Perceived barriers were related to disease progression, support from experts, and integrating DAS use with the use of a wheelchair.	Environmental and personal factors are under-investigated influencers of user-device interactions. Studies should consider barriers and facilitators for activity monitoring.
E: what is the influence of motivation on user-device interaction?	
E: can we estimate energy costs through activity monitoring and effort reduction by DAS? L: endurance was enhanced with the use of a DAS, but depended on the residual capabilities of the user.	

2.4.2 Adaptations over time

From this review, it is clear that the ability to adapt following skill acquisition, fatigue, or disease's progression, have not yet been properly investigated over time. In order to benefit from a DAS, it is crucial that the user acquires the skills to operate the device and retains them over time. Previous studies have shown that training with a DAS is feasible in people with NMD [55, 56]. However, it is currently unclear which skills are needed and which ones need to be learned to increase a device's benefits. Moreover, due to the significant loss in upper extremity functionality and increasing fatigue due to the progressive nature of some NMDs, handling the device can become increasingly difficult over time [7, 37, 38, 48]. The perceived benefits, which vary between users, can even decrease so much over time that the user decides to stop using the device completely. Similarly, a recent systematic review on the short-term benefits of wearable devices found that as the disability level changes the device benefits change as well [57]. To prevent discontinuation, researchers and developers are promoting intuitive and adaptable DAS that counteract pathological changes due to disease progression [5, 43]. However, such developments require extensive insights into a user's ability to adapt motor capacity, capability, and performance over time. Activity monitoring through wearable sensors might provide a solution to acquire evidence over such long periods and some recent advances have been made in this field [28, 49, 50].

2.4.3 Relations across framework cells

Evidence from experts suggests that the ICF model and contextual constructs are considered during the design process and the formulation of device requirements, however, this is not reflected in the literature findings. It is commonly assumed that a DAS primarily affects body functions, which influences performance in the activity and participation. However, most literature focused mainly on the technical and design requirements necessary to overcome body function impairments, mostly neglecting the relationship between motor ability and ADL performance. Only one study directly investigated the relation between joint mobility and the ability to perform common ADL, concluding that improvements in joint mobility alone does not directly translate to changes in ADL performance in a home environment [12]. For instance, increased arm elevation with a device from 26.4° to 67.1° did not result in the ability to comb one's hair, while peers could execute the same task at an elevation angle of 44.6° with and without a device. Furthermore, Heide et al. also indicated that environmental and personal factors, such as adjustments in the home setting and compensatory movements, have an important influence on ADL performance in multiple studies [6, 7, 12]. These factors could affect the relationship between changes in joint mobility and the ability to complete tasks. In addition, Cruz et al. 2020, stated that lack of or delay in funding, lack of support from experts, and lack of proper device integration with the wheelchair resulted in discontinuation of the DAS [48]. The authors therefore recommended to include multiple factors when evaluating the effectiveness of a DAS, especially in a home environment. As a result, we propose that future research considers the device requirements of multiple ICF model components and contextual constructs, i.e. motor ability, capability and performance, within the same study design.

2.4.4 Unaddressed framework cells

Our mixed-methods approach revealed that literature focuses on selected framework cells which are often considered separately, thus limiting the evidence on the interaction between cells. For instance, current evidence shows that muscle activity and joint mobility are both influenced by load reduction [12, 42-46]. However, as also pointed out by experts, the interaction between 1) muscle activity and joint mobility and 2) how this is affected by disease and 3) how this could be restored by the device should be investigated. Stabilizing and facilitating the shoulder girdle requires relatively complex muscle coordination, which is affected in people with NMD [4, 58, 59]. Experts believe that insights into how muscle coordination is affected would benefit the development of a universal DAS and to optimize a device to fit the individual requirements [5, 60]. Other symptoms, such as stiffness, pain, and early fatigue, were also regarded as important factors by experts and were present in the literature findings, however these topics were not clearly addressed in the study designs. Three studies investigated comfort levels with limited evidence on stiffness, pain, or fatigue [43, 47, 48] and were therefore not represented in the respective cells. In contrast, while pain and stiffness are highly prevalent and should be reduced to comfortable levels, they might not be the limiting factors for ADL performance in people with NMD [8]. Bergsma et al. 2017, found that participants who had high pain and stiffness levels also reported relatively few activity limitations, which indicates an overuse in their body functions. However, it is unclear whether a DAS positively relieves these symptoms and how effects differ across motor capacity, capability, and performance. Therefore, future research should consider the importance of these unaddressed cells for device development as possible influencers or as main device requirements.

2.4.5 Research recommendations

From our analysis it is clear that integration and inclusion of ICF components and contextual constructs are needed to bridge the knowledge gaps in the development and evaluation of DAS. To realize these two tasks, we propose four steps with each a focus point, examples from our analysis, and suggestions for the design of future studies.

First, we propose that future research incorporates multiple ICF components and contextual constructs within one study design. It is commonly assumed that body functions are primarily affected by a DAS, which shapes the effects on activity and participation. Therefore, we suggest to focus on the relation between these two components before proceeding to examine the effect of environmental and personal factors. Our analysis shows that muscle activity and joint mobility are affected by load reduction, but their relation has not been investigated nor linked to ADL performance. Gandolla et al. 2020, deducted similar conclusions from their focus on activity and participation [57]. Therefore, we recommend to investigate the relation between the two ICF components in a biomechanical framework under various levels of load reduction during functional tasks to optimize effort reduction and mobility improvements during common ADL.

Secondly, the influence of environmental and personal factors should be investigated when deploying a device. Barriers within these factors have been ascribed to personal preferences, such as performing ADL without support or conserving energy altogether, and home setting, such as limited space or a fixed location of the device, but also to a lack of funding or support from experts [7, 48].

Thirdly, research should include short term, such as within- and between-day repeatability, and longitudinal measurements, such as yearly follow ups, to monitor adaptations over time. For example, limited evidence indicated that a DAS delays fatigue onset and reduces fatigue, but it is unclear if and how this affects ADL performance throughout the day. From two ongoing studies and previous literature we consider activity monitoring a method to quantify device utilization and a proxy for motor performance [28, 49, 50, 61]. In addition, characteristics of ADL, such as a diminished frequency and variety, can be used to monitor adaptations in activity and participation [28]. Furthermore, we propose to include muscle strength and joint mobility measures to monitor disease progression [10, 62]. A longitudinal cohort study should investigate the relationship between disease progression and adaptations in daily activity of people with NMD over the course of a year. Disease progression and daily activity could be sampled every few months. Daily activity should then be averaged over the timespan of several days to include a range of ADL. The relationship could be expressed as a correlation between disease progression factors, muscle strength and joint mobility, and device utilization, and motor performance.

Lastly, experts and literature agree that user satisfaction, such as perceived benefits and comfort, should be taken as guidance to evaluate the device effectiveness. Cruz et al. 2020, and Gandolla et al. 2020, also recently promoted the use of objective and subjective measures as both measures provide equally important evidence of the functional status of the user [48, 57]. Therefore, device requirements should align with the needs and goals of the user and additionally aim to relief symptoms of pain, stiffness, and fatigue. Furthermore, future research should validate and incorporate subjective measures related to the respective ICF model components and contextual constructs. A possible longitudinal study could include a questionnaire that links symptoms of pain, stiffness, and fatigue in people with NMD to the motor capability and capacity to perform ADL w/o a DAS and perceived benefits over the course of a year.

2.4.6 Recommendations for developing evaluation tools

When the above evidence is combined, it will provide the basis for understanding how standardized device benefits result in daily device utilization accounting for changes in disease progression and users' needs and goals over time. Evaluation tools developed along these insights should be standardized and validated with focus on international consensus, as indicated by recent research [39]. We suggest several minimal requirements for the development of such tools. First, the tools require an integration of translatable cells that cover at least two of the contextual constructs. Second, the tools should be applicable alongside the development and after deployment of DAS. Third, subjective measures, such as perceived benefits and comfort, should be included in a device's evaluation of effectiveness near the product final development stages.

2.5 CONCLUSION

Three knowledge gaps were identified and given synthesized research recommendations based on the integration and inclusion of ICF model components and contextual constructs. First, adaptations due to altered support requirements over time are poorly understood. Second, relations between ICF model components and contextual differences are limited. Finally, several framework cells, such as comfort levels, were brought to our attention by experts that were not covered sufficiently in scientific literature. We promote the use of multiple ICF model components and contextual constructs within research to benefit the development of DAS. Research should quantify device benefits and daily device utilization with respect to disease progression and users' needs and goals over time. Furthermore, we suggest several minimal requirements for the development of evaluation tools of DAS. The tools are required to cover multiple framework cells and to be applicable in various environments, for various users, and on multiple time points. Moreover, the tools should integrate objective and subjective measures to evaluate device effectiveness.

2.6 APPENDIX

Appendix table 1. Literature search terms, criteria, and strings.

	Search Terms	Inclusion criteria	Exclusion criteria
Population	Neuromuscular Diseases Neuromuscular Manifestations Muscular Dystrophies Musculoskeletal Diseases Disabled Persons	Neuromuscular Diseases Neuromuscular Manifestations Muscular Dystrophies Musculoskeletal Diseases Disabled Persons	Animals Cadaver
Impairment	-	-	Amputees Fractures, Bone Cognition Disorders Muscle-Tendon problems
Region	Upper Extremity Shoulder Arm Elbow Scapula	Upper Extremity Shoulder Arm Elbow Scapula	Lower Extremity Spine Hand Head Psychological*
Intervention	Self-Help Devices Orthotic Devices Gravity Compensation Exoskelet* Robot* Support Devices Arm Support Arm Assistance	Self-Help Devices Orthotic Devices Gravity Compensation Exoskelet* Robot* Support Devices Arm Support Arm Assistance	Medicine General Surgery Electrical Stimulation Drug Therapy Transcranial Magnetic Stimulation Magnetic Field Therapy Virtual Reality Exposure Therapy
Evaluation	Investigative Techniques (PubMed only)	Investigative Techniques Activities of Daily Living Range of Motion, Articular Torque Reachable Workspace Joint Loading Muscle Activity Body Function Activity	-
Others	-	-	Publication Date: 01-01-2000 up until 30-06-2020

Appendix table 1. (Continued)

	Search Terms	Inclusion criteria	Exclusion criteria
Search string (PubMed)	(Neuromuscular Diseases[All Fields] OR Neuromuscular Manifestations[All Fields] OR Muscular Dystrophies[All Fields] OR Musculoskeletal Diseases[All Fields] OR Disabled Persons[All Fields]) AND (Upper Extremity[All Fields] OR Shoulder[All Fields] OR Arm[All Fields] OR Elbow[All Fields] OR Scapula[All Fields]) AND (Self-Help Devices[Title/Abstract] OR Orthotic Devices[Title/Abstract] OR Gravity Compensation[Title/Abstract] OR Exoskelet*[Title/Abstract] OR Robot*[Title/Abstract] OR Support Devices[Title/Abstract] OR Arm Support[Title/Abstract] OR Arm Assistance[Title/Abstract]) AND (Investigative Techniques[All Fields]) AND (“2000/01/01”[PDAT] : “2020/06/30”[PDAT])		
Search string (Web of Science)	ALL=(Neuromuscular Diseases OR Neuromuscular Manifestations OR Muscular Dystrophies OR Musculoskeletal Diseases OR Disabled Persons) AND ALL=(Upper Extremity OR Shoulder OR Arm OR Elbow OR Scapula) AND TS=(Self-Help Devices OR Orthotic Devices OR Gravity Compensation OR Exoskelet* OR Robot* OR Support Devices OR Arm Support OR Arm Assistance) Timespan=2000-2020		

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Appendix table 2. Expert views as tabulated keywords from the collective focus group

ICF model	Impairments & Limitations	Effectiveness & Requirements
Body functions	Common: Difficulty with performing motions against gravity; shoulder pain; muscle weakness/ pain/ fatigue, coordination; inactivity of muscle groups or other detrimental effects due to compensational strategies; joint rigidity; hand function Solved: Amount of range of motion; shoulder pain; muscle fatigue and coordination; posture Unsolved: Larger range of motion; hand function; muscle weakness/ pain/ fatigue (required residual forces for device usage); finer coordination; integration with wrist support; joint rigidity; DAS induced inconvenience; limited shoulder motion/ stabilization; pronation/ supination	Effective: Device usage is intuitive; it supports and motivates usage of affected side; effectiveness is mostly subjective Ineffective: DAS has limited range of motion; it requires additional energy; disease and device utilization shows progression; not used as intended according to specifications; user has compensation strategies; ineffectiveness is mostly subjective Minimum requirement: DAS has controlled interaction from joints (shoulder, elbow, wrist driven control); promotes normal posture, natural motion, and range of motion especially above shoulder/ head level; provides adaptive support; usable in training; frictionless; usage is intuitive; provides shoulder support; decreases fatigue
Activity	Common: Limitations in repetitions; independence/ self-care (eat/ drink, writing); interaction with environment; DAS acceptance (embarrassment); computer work Solved: Certain level of dependency/ self-care (eat/drink, computer work) Unsolved: Activities outside range of motion; interaction with environment; natural motion hand to mouth after elevation (eating/ drinking, pouring a drink); writing; easy don/ doff; limited activities supported; problems with computer work still exist; other types of DAS are sometimes better (adjusted spoon), perhaps a combination is necessary	Effective: DAS changes users' inability to ability or improves a realistic user need (repetitions); completing tasks within acceptable range of motion at comfortable level Ineffective: Mostly subjective Minimum requirements: Mostly subjective; DAS should support independent self-care; improvement of ADL; allow control of wheelchair; enable writing; motivate users

interviews. DAS: Dynamic Arm Support, w/o: with and without.

Previous projects & Remaining questions	**Research priorities**
Previous projects: Influence of gravity compensation; basic demands for ADL tasks (gravity compensation, range of motion); arm positioning/ balancing; biomechanical interaction with support; force transposition from DAS to user; head support; arm brace for shoulder stabilization; improvement for proprioceptive feedback Remaining questions: Supporting posture; interaction disease-device; kinematic profile (abilities, limitations); fatigue/ pain/ residual strength; required residual force/ strength; compensation support and muscle rigidity; feedback regarding shoulder motion	Interests: Biomechanics of usage (interaction forces, utilization of range of motion, feedback regarding shoulder motion as therapy tool); interaction ambulant user-device; fatigue Device user's interests: Upper extremity performance (RoM/ motion progression, frequency of usage, speed/effort, training, time/ effort); how to prevent inactivity/ fatigue Gaps: Biomechanical interaction user-device (forces); required residual strength capacity; supporting natural posture/ motion; accurate description/ indication disease progression; need for short-term measurements; adequate tests for pain/ stiffness/ fatigue/ functionality in combination with DAS; cross-sectional tests are not representative, need for longitudinal tests
Previous projects: Role of DAS in independency/ self-care; motion intention detection; obscure interaction with environment; evaluation of ADL support; motion freedom w/o DAS Remaining questions: What are the kinematic profiles (activities and limitations) w/o DAS; how to increase the set of supporting activities	Interests: What are individuals' needs or goals; device usage duration/ frequency; area map of gravity compensation linked to range of motion; identification and performance evaluation of self-care activities at home (eating/ drinking, computer/ tablet usage) Device user's interests: Recognizing non-usage; types of activities (can/not and w/o DAS)/ activity profile; user-device/needs-specs match; distribution of energy over activities/ motivation; activity monitoring with moderated feedback (users indicate they are aware of their activities/ energy) Gaps: Recognizing use/ non-use; activities/ functional capacity w/o DAS in home setting with respect to ADL needs; description of active population; environmental factors

Appendix table 2. (Continued)

ICF model	Impairments & Limitations	Effectiveness & Requirements
Participation	Common: Dependency on caregiver; trying to blend in; reaching individual goals; inactivity due to body function problems; age and disease dependent; eating at a restaurant Solved: Certain level of dependency/ self-care (eat/ drink at restaurant); reaching out hand for contact Unsolved: Integration with wheelchair; trying to blend in/ embarrassment; limited social contact due to caregiver dependency and hand contact; DAS not accessible/ transportable	Effective: Mostly subject to user needs and social context; if a user becomes more independent Ineffective: Mostly subject to user needs and social context; progression of usage; environmental factors limiting usage Minimum requirements: DAS is transportable; allows interaction with environment; double-sided DAS to stimulate both arms
Other	Common: Users are very intuitive with compensational strategies and masking problems Solved: Mounted to wheelchair provides constant access Unsolved: Aesthetics/ hardware/ size/ transportation; DAS not adaptive to disease progression; limited knowledge from user/ therapist on DAS; costs/ repair; user-device mismatch due to shortcomings in selection procedure/ disease progression; variation in user capabilities/ no universal method	Effective: DAS wear/utilization (actual usage); acceptation; user motivation Ineffective: DAS prototypes that do not reach the market Minimum requirements: Mechanistic balance; application to monitor user-device match; smaller size; smaller increments/ adjustments for gravity compensation; adaptive support; proper manual/ guidance device usage

Previous projects & Remaining questions	**Research priorities**
Previous projects: Inquiring user needs; ADL and home usage; Quality of Life; user-DAS match procedure/ evaluation (does it work and can it be optimized?) Remaining questions: Relation between device usage and psychomotor factors (motivation, mood, good/bad motion profiles, energy costs); environmental factors (home setting and outdoors) and personal factors; aesthetics/ embarrassment	Interests: Device usage in home settings; DAS in ambulant users as body mounted Device user's interests: Psychological drives; social interaction; experiencing normal puberty; barriers of confrontation Gaps: What is important/ positive for users; individual wishes and possibilities; quality of performance at home
Previous projects: General DAS information (also for insurance companies); involvement of young users Remaining questions: Fine-tuning prototypes with simulation modeling; transposing applied force to user; reducing internal friction/ required force from user; terminology of all DAS related items; aspects of user-device match (individual assessments; differences between DAS)	Interests: Time DAS motor is active; commands send to readjust motors; integration of devices and other DAS; aesthetics; use/ disuse and affected/ unaffected side; Device user's interests: Mostly subjective; allocate time to develop DAS acceptance in selection procedure Gaps: Problem cases; intensity of device utilization; DAS as measuring tool; matching wishes/ needs users; limited knowledge from care providers about DAS; structured method to decide which DAS and when; easy yet universal device for users and therapists; limited choice in DAS (insurance/ costs); test period at home; communication between professionals should improve to find the best user-device match Other: Psychomotor and complexity of advice procedure should be taken into account; need to know actual usage regarding ADL; what are other unknowns; user-device adaptation; therapy to improve usage; website with all DAS information; adequate tests to indicate need for DAS; more research for practical things (selection procedure, costs, aesthetics); rehabilitation can observe ADL but progress is training dependent

CHAPTER 3

Superficial shoulder muscle synergy analysis in Facioscapulohumeral Dystrophy during humeral elevation tasks

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3.1 INTRODUCTION

Facioscapulohumeral Dystrophy (FSHD) is characterized by progressive muscle wasting which primarily affects the face and shoulder area [18, 33]. Muscle quality decreases due to fat infiltration, but is weakly correlated with age where age onset varies greatly [33, 63]. Commonly occurring body impairments and functional limitations include scapular winging, joint instability, and a decline in upper extremity functionality [34, 64-67]. In a questionnaire-based survey, reaching and lifting objects above shoulder level were reported as “most limited” activities by 45% of FSHD participants [34]. Relative surface area, as a measure of the reachable workspace, decreases by 23 to 87% depending on the level of strength loss, in people with FSHD [10, 68]. Muscles attaching to the scapula are the most affected, with the Trapezius and Serratus Anterior muscles becoming atrophied and showing fat infiltration in more than 85% of individuals with FSHD [69]. These losses in tissue quantity and quality become evident at the earliest stages of the disease [69, 70] and translate into a diminished strength of the scapular rotator muscles. In turn, this limited muscle function could result in incomplete rotation and stabilization of the scapula.

Electromyographic assessments of muscle function can provide insight in the muscle activation strategies used for scapular stabilization and mobilization in people with FSHD. Previous research has shown an approximately twice as high muscle activity in FSHD participants compared to healthy individuals for the Deltoid, Trapezius Descendens, and synergist Biceps muscles during reaching tasks [4]. The increased activity of selected shoulder muscles can be postulated to compensate for the loss of strength, with scapular mobilization possibly affected as a result. In healthy individuals, scapular mobilization and stability are necessary during humeral elevation, particularly above shoulder level [71-73]. At present however, the way in which scapular rotator and humeral elevator muscles are coordinated by FSHD individuals during daily tasks is still unclear. The extent of these alterations that are known to occur in other diseases affecting the shoulder, including stroke, multiple sclerosis, and shoulder impingement [74-79], indicate that the neuromuscular output can be affected by the disease.

Muscle synergy analysis can be used to reveal alterations in the coordination of groups of muscles. In healthy individuals the central nervous system activates muscles in groups, as a neural strategy to simplify the control of multiple degrees of freedom [80]. These group activations, commonly called muscle synergies, can be described by the relative contribution of each muscle (weights) during a common time-dependent activation command (coefficients) [17]. Muscle synergy analysis of the upper extremity in people post-stroke has revealed alterations in the shoulder muscle synergies during isometric force generation [20] and dynamic tasks [21]. A high similarity between affected and unaffected arm muscle synergies was shown in a variety of daily activities, together with the presence of compensatory strategies by Trapezius and Pectoralis muscles during reaching tasks [20, 22, 23]. In people with FSHD, however, it is unknown how muscle synergies change during the execution of upper extremity daily tasks. Understanding the neuromuscular output can help reveal how the disease-resulting changes in kinematics are underlined by muscular changes, with implications for the long-term management of the condition.

This study concentrates on planar humeral elevation tasks to understand the neuromuscular changes affecting the shoulder muscles, including muscles responsible for scapula rotation and stabilization, in people with FSHD compared to healthy individuals. We hypothesized that in people with FSHD the maximum activity of prime movers of humerus and scapula and of synergist muscles would be higher compared to healthy individuals. Secondly, we also hypothesized that muscle synergies would show alterations in people with FSHD, reflecting the increase in maximum activity, mainly in synergy weights. The second hypothesis was tested to investigate whether the known shoulder mobility limitations in people with FSHD would affect the muscle synergies.

3.2 METHODS

3.2.1 Participants

Eleven healthy control participants (5M/6F, 55±14ys, 175±7cm, 69±8kg, 11Right-Dominant) and eleven participants with FSHD (6M/5F, 54±15ys, 177±11cm, 78±21kg, 2LD/9Right-Dominant) were included in this study. Healthy participants were informed by advertisement flyers located at University Medical Center Groningen. People with FSHD were informed about the study through the Dutch Association for Neuromuscular Diseases (Spierziekten Nederland, Baarn, NL). Healthy and participants with FSHD were included in this study if they were aged between 18-75 years, able to read and understand Dutch, and able to give written informed consent. Additional criteria for people with FSHD were the ability to transfer from wheelchair to chair with side- and lower back-rest, and a Brooke scale score of 3 or 4. Healthy participants were excluded if they were diagnosed with pathologies that could interfere with the measurement results, had a presence of pain in the shoulder, a history of severe trauma of the shoulder within the previous two years (e.g. fracture, luxation). Participants with FSHD were excluded if they had comorbidities that could interfere with the measurement results, previous surgery on the right shoulder, extrinsic causes of shoulder pain, a history of severe trauma, or were unable to elevate the right arm above 30°. Age, gender, hand-dominance, body height, and body mass were also recorded. The central Medical Ethical Committee of University Medical Center Groningen approved the study (NL55711.042.15), which was carried out in accordance with the guidelines of the Helsinki protocol. Participants were informed about the procedure beforehand and provided written informed consent.

3.2.2 Movement tasks

The participants were positioned in a chair with a left side-rest and lower back-rest and with the seat height adjusted to achieve a knee flexion angle of 90°. Participants received detailed instructions prior to the execution of each task regarding the movement. For the shoulder abduction-adduction task (SAA), the right arm was first positioned downward with the elbow straight and the hand palm facing forward (Figure 1). The movement consisted of lifting the arm as far as possible in the coronal plane and bringing it back to the start position while keeping the trunk and elbow straight, with the hand palm facing forward. The shoulder flexion-extension task (SFE) was similarly executed but with the hand palm facing medially and the thumb pointing forward. One researcher mirrored each task at pace with the participant. Each task was repeated three times but not consecutively as the order of the tasks was randomized.

3.2.3 Measurement and processing

Kinematics of the trunk, chest, and right-sided upper extremity was recorded using the Optotrak 3020 system (Northern Digital Inc., Canada) [81]. Single markers were placed on bone landmarks and rigid bodies were placed on soft tissues on the lateral side of the upper and lower arm as shown in table 1. Humeral elevation was calculated from the recorded kinematics and expressed as joint angle between trunk and humerus where 0° represents the arm straight downward and 180° straight upward.

Surface electromyograms (EMG) of the right side muscles were recorded for the prime humeral elevator/depressors and scapular rotator muscles, i.e. medial Deltoid, Pectoralis Major clavicular head, Latissimus Dorsi, Trapezius Descendens, Trapezius Ascendens, and Serratus Anterior 5-6th rib, and the synergist muscles Biceps Brachii short head and Triceps Brachii long head. Data were captured at 2000Hz using the Delsys Trigno system (Delsys Inc., UK) [19]. Maximum voluntary contractions (MVCs) were recorded beforehand (appendix table 1). The recorded

EMG data were filtered with a 4th order Butterworth 20-450Hz bandpass and a 49-51Hz bandstop filter, rectified, smoothened with a 100ms moving window, normalized to the maximum amplitudes derived from all MVC and task recordings, and filtered with a 4th order Butterworth 5Hz low pass filter. The maximum task-specific muscle activity was extracted as highest normalized amplitude over all task repetitions. Time was normalized to 1001 samples for each repetition ranging from 0 to 100%.

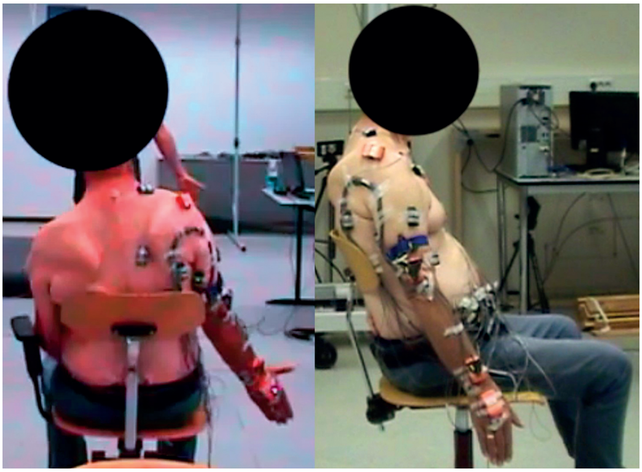

Figure 1. Experimental set up of a FSHD participant about to perform shoulder abduction-adduction (left) and flexion-extension (right).

Table 1. Single and rigid body markers

Marker #	Body location
1	Spinal process of 7th cervical vertebra
2	Jugular notch clavicle-sternum
3	Xiphoid process of sternum
4	Acromio-clavicular joint (left)
5	Acromio-clavicular joint (right)
6-8*	Lateral upper arm (right, 1/3 of acromion to lateral epicondyle)
9	Lateral epicondyle (right)
10	Medial epicondyle (right)
11-13*	Lateral lower arm (right, 1/2 of lateral epicondyle to styloid process of radius)
14	Styloid process of radius (right)
15	Styloid process of ulna (right)
16	Head of the 3rd metacarpal (right)

* Rigid body refers to a rigid cluster of three markers.

Kinematics and EMG recordings were executed consistently with one researcher placing the markers and electrodes and another research assessing the placement and data quality.

3.2.4 Muscle synergy extraction

EMG data were pooled per participant to contain equal samples of both tasks in a single matrix to investigate the shared synergies across humeral elevation planes. Muscle synergies were then extracted using Non-Negative Matrix Factorization (NNMF), which decomposed the matrix into 1 to 8 sets of components consisting of weights and coefficients [17]. These weights and coefficients were converted to a unit vector and represent normalized muscle activity (0-1). Additionally, for each set of components (synergy), the NNMF provided the percentage of variance accounted for of all muscles (VAF) and per individual muscle (VAFM). The minimum required number of synergies per participant were extracted using as thresholds VAF > 90% and VAFM > 75% [17]. The variance accounted for per task was calculated with respect to the reconstructed data (weights * coefficients) for each synergy. Coefficients were then averaged for pooled repetitions per task. Synergies were clustered within each group using an iterative process that matched weights in an ascending order based on Pearson's correlation coefficients. The muscle synergy extraction procedure was executed for two conditions. One condition included the complete motion and the second condition focused on the upward motion up to 60° humeral elevation.

3.2.5 Statistical analysis

Humeral elevation differences between groups were investigated using independent-samples Mann-Whitney U tests. To test the first hypothesis on whether EMG amplitudes of prime movers and synergist muscles would be higher in people with FSHD, the maximum muscle activities were compared using a non-parametric analysis of variance, with Task and Muscle as within-group factors and Group as between-group factor (R v3.5.0, The R Foundation for Statistical Computing, nparLD package) [82]. The Post-hoc tests were performed accordingly between groups using independent-samples Mann-Whitney U tests, and between tasks using related-samples Wilcoxon signed rank tests. Alpha levels were corrected for multiple comparisons and set at 0.025. Effect sizes were expressed as Cohen's d (very small: 0.00- 0.01, small: 0.01 - 0.20, medium: 0.20 - 0.50, large: 0.50 - 0.80, very large: 0.80 - 1.20, and huge: >1.20) [83]. Furthermore, the standard error of measurement (SEM) was calculated on the consistency of maximum muscle activity over repetitions for each group and consequently used to calculate standard deviations of mean group differences [84].

To test the second hypothesis on whether muscle synergies were altered or dissimilar in people with FSHD, Pearson's correlation coefficients were used to quantify synergy weight and zero-lag correlation coefficients to quantify synergy coefficient similarities (α: 0.025) [85]. Correlation coefficients values were calculated only for significantly similar synergy weights to minimize type I errors. Additionally, within-group similarity was calculated through the EMG cross-validation method [86], and Pearson correlations (r) for synergy weights only. Differences in within-group similarity from EMG cross-validations were tested with Fisher's least significant difference (LSD) post-hoc test with the number of muscle synergies as a factor (α: 0.025).

3.3 RESULTS

3.3.1 Kinematics

All participants successfully completed all tasks. The control group elevated the humerus significantly higher in SAA to 149±19° (N=22, Cohen's d:4.28, p<0.001) and in SFE to 141±17° (N=22, Cohen's d:3.09, p<0.001). The FSHD group's maximum humeral elevation was 70±18° and 83±20° during the SAA and SFE task, respectively.

3.3.2 Muscle activity

Maximum muscle activities were significantly different for Task (p<0.010), Muscle (p<0.001), Muscle*Task (p<0.001), Group*Muscle (p<0.001), and Group*Muscle*Task (p<0.001), but not for Group (p:0.248) or Group*Task (p:0.121). Post-hoc tests of the Group*Muscle*Task interaction effect revealed that maximum muscle activities were significantly different between groups (FSHD-control) for Biceps Brachii SFE: +25±2% (N=22, Cohen's d:1.38, p:0.013), Trapezius Ascendens SAA: -32±8% (N=22, Cohen's d:-1.45, p:0.004) and SFE: -41±6% (N=22, Cohen's d:-1.95, p:0.001), and Serratus Anterior SAA: -39±4% (N=22, Cohen's d:-1.72, p:0.002) (Figure 2). Within the control group there was a significant difference between tasks (SAA-SFE) for Trapezius Ascendens: -14±14% (N=22, Cohen's d:-0.74, p:0.005) and Latissimus Dorsi: -5±6% (N=22, Cohen's d:-0.25, p:0.024). Within the FSHD group significant differences between tasks (SAA-SFE) were found for Biceps Brachii: -12±15% (N=22, Cohen's d:-0.55, p:0.010), Trapezius Descendens: +21±25% (N=22, Cohen's d:0.79, p:0.024), Pectoralis Major: -13±16% (N=22, Cohen's d:-0.80, p:0.010), Serratus Anterior: -19±23% (N=22, Cohen's d:-0.83, p:0.014), and Latissimus Dorsi: -17±18% (N=22, Cohen's d:-0.90, p:0.019). The SEMs were 1.9% and 3.3% for the control and FSHD group, respectively.

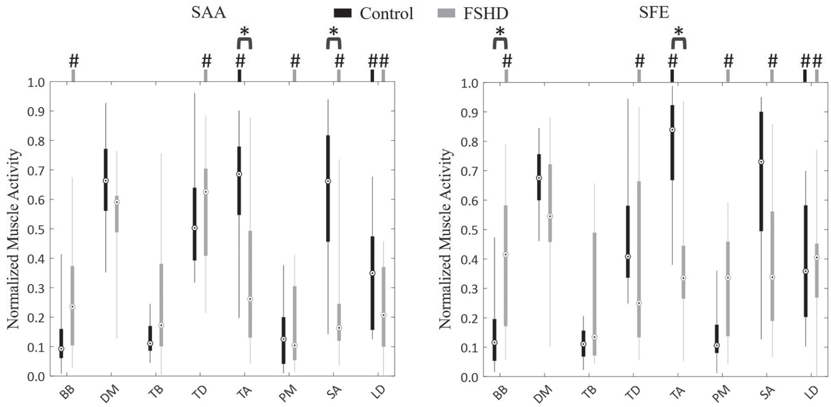

Figure 2. Boxplots of maximum muscle activity amplitudes of control (black) and FSHD group (grey) for the SAA (left) and SFE (right) tasks. (*: significant group differences; #: task differences; p<0.025). Boxes and whiskers indicate minimum, first quartile, median, third quartile, and maximum. BB: Biceps Brachii; DM: medial Deltoid; TB: Triceps Brachii; TD: Trapezius Descendens; TA: Trapezius Ascendens; PM: Pectoralis Major; SA: Serratus Anterior; LD: Latissimus Dorsi.

3.3.3 Muscle synergies

The number of synergies extracted were equally distributed between the two groups (Figure 3). In each group at least 90% of the variance was described with one synergy for two participants, two synergies for eight participants, and three synergies for one participant. The control and FSHD group's synergies were clustered into two sets each where FSHD participants were also investigated individually and compared to the clustered control synergies (Figures 4 and 5). Appendix figure 1 shows the participant-specific synergies.

Synergy #1 on average accounted for 74±19% variance for FSHD participants (controls: 87±9%) in the SAA task and 50±35% VAF (controls: 86±9%) in the SFE task. The VAF per task by synergy #2 was 29±12% for FSHD participants (controls: 15±3%) in the SAA task and 59±27% (controls: 15±6%) in the SFE task. Within-group similarities for synergy weights #1 and #2 were, respectively, for controls r:0.73±0.15 (N=55) and r: -0.06±0.37 (N=36), and for FSHD R:0.00±0.42 (N=55) and 0.08±0.56 (N=36). Correlation of synergy weights was not significant for any synergy combination between groups. On an individual level two FSHD participants (#6, #9) showed significant similar synergy weights where synergy #1 correlated with control synergy #2 (p:0.023, r:0.78 and p:0.001, r:-0.92 for participant #6 and

#9, respectively). Correlation coefficients values for the SAA and SFE tasks were respectively r:0.19 and 0.24 (FSHD #6, $p<0.001$ and $p<0.001$), and r:0.09 and 0.18 (FSHD #9, p:0.006 and $p<0.001$).

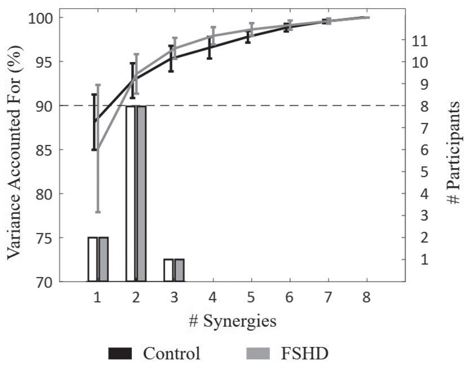

Figure 3. Variance accounted for as means and standard deviation (lines; left y-axis) and number of extracted synergies (bars; right y-axis) of the control (black) and FSHD group (grey). The dashed line indicates the 90% VAF threshold.

In the upward motion to 60° humeral elevation condition, at least 90% of the variance was described by two synergies for seven controls and seven FSHD participants, and three synergies for four controls and one FSHD participant. Three FSHD participants did not reach at least 60° in both tasks and were excluded for this condition. Control and FSHD participants' synergies were clustered into three sets each (Figure 6). Synergy #1 accounted for 63±11% variance for FSHD participants (controls: 62±17%) in the SAA task and 39±10% (controls: 45±16%) in the SFE task. For synergy #2 this was 37±10% and 56±16% (controls: 29±21%, 47±16%) in the SAA and SFE tasks respectively, and 6% and 41% (controls: 24±9%, 21±19%) for synergy #3. Synergy weights showed significant similarities between groups for synergy #1 (r:0.84, p:0.009) where correlation coefficients values showed r:0.98 ($p<0.001$) for both tasks.

EMG cross-validations showed that less variance was accounted for by other participants' complete synergy set than one's own set in the control ($p<0.001$) and FSHD group ($p<0.001$, figure 7). With the exception of controls' synergy #1, other participants' individual synergies accounted for less variance than the complete set ($p<0.001$). Upon further inspection, synergy #2 accounted for an additional 5±2% VAF in controls after which the criteria of >90% was met for 8 participants (Figure 3). In the upward to 60° humeral elevation condition all factors accounted for less variance than one's own synergy set ($p<0.025$).

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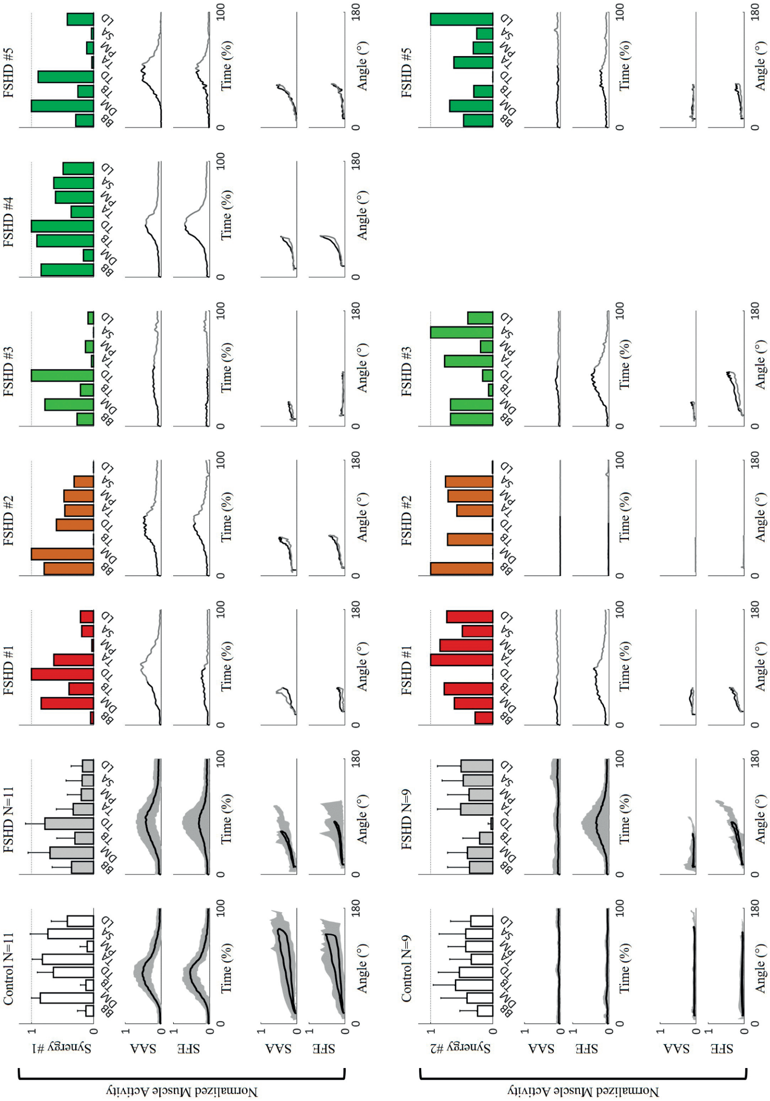

Figure 4. Muscle synergies no. 1 (top, N = 11) and no. 2 (bottom, N = 9) of the control group (black) and the FSHD group (grey) and participants 1-5 for the SAA and SFE tasks. The FSHD participants were ranked by averaged humeral elevation in ascending order from left to right (#: participant number). N equals the amount of participants within each clustered synergy. Clustered synergies are presented as mean (rectangles and black thicker line) with standard deviation (bars) or ±95% confidence interval (grey area). Individual synergy coefficients show upward (black line) and downward motion (grey line). Participants #1, 2, 3, and 5 have two synergies, and 4 has one synergy. BB: Biceps Brachii; DM: medial Deltoid; TB: Triceps Brachii; TD: Trapezius Descendens; TA: Trapezius Ascendens; PM: Pectoralis Major; SA: Serratus Anterior; LD: Latissimus Dorsi.

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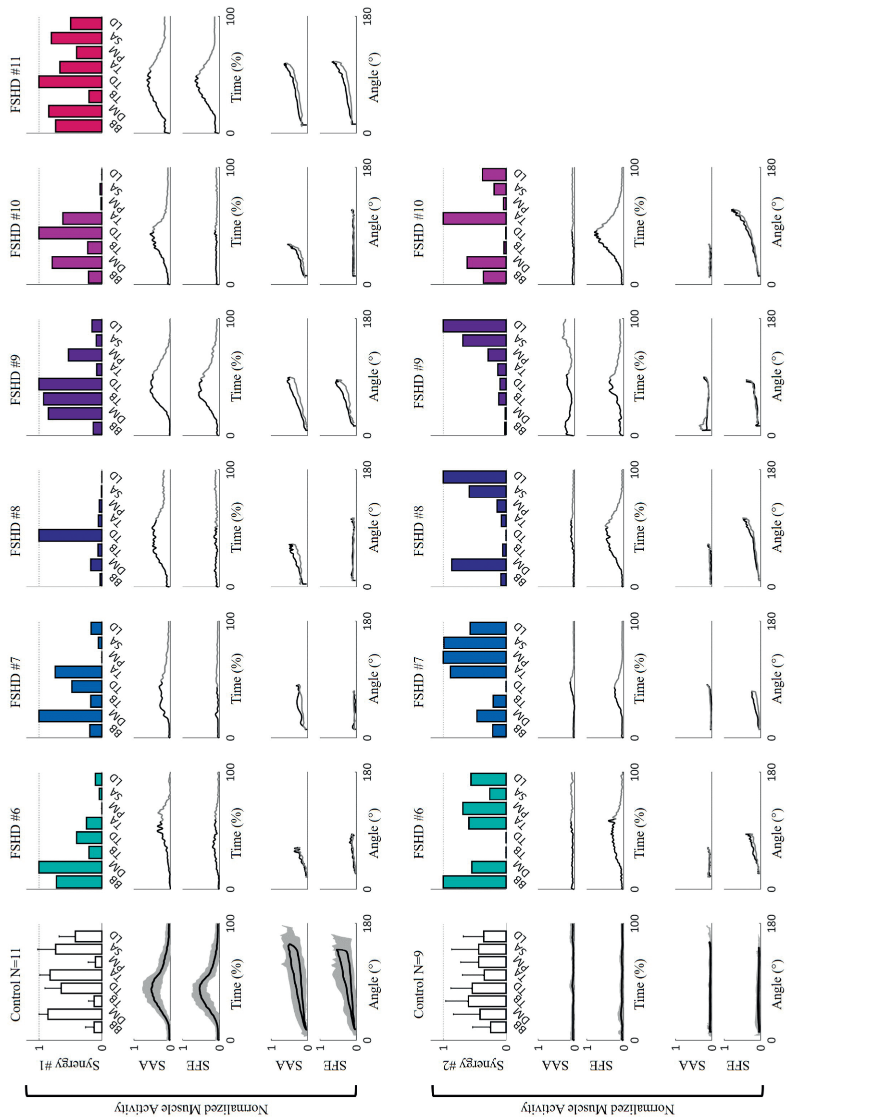


Figure 5. Continued from previous. Muscle synergies no. 1 (top, N = 11) and no. 2 (bottom, N = 9) for the remaining FSHD participants 6-11 for the SAA and SFE tasks. The FSHD participants were ranked by averaged humeral elevation in an ascending order from left to right (#: participant number). N equals the amount of participants within each clustered synergy. Clustered synergies are presented as a mean (rectangles and black thicker line) with standard deviations (bars) or ±95% confidence interval (grey area). Individual synergy coefficients show upward (black line) and downward motion (grey line). Participant #6 has three synergies (synergy #3 is presented in appendix fig. 8), 7-10 have two synergies, and 11 has one synergy. BB: Biceps Brachii; DM: medial Deltoid; TB: Triceps Brachii; TD: Trapezius Descendens; TA: Trapezius Ascendens; PM: Pectoralis Major; SA: Serratus Anterior; LD: Latissimus Dorsi.

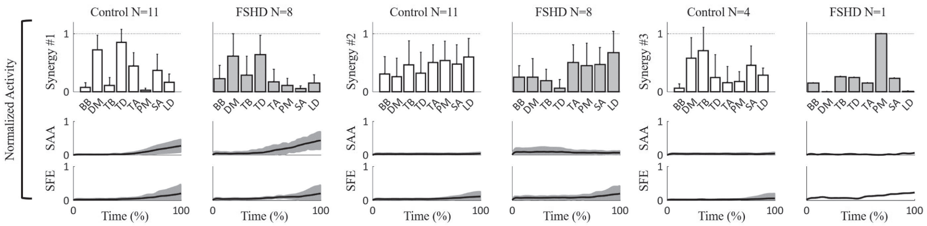


Figure 6. Muscle synergies no. 1 (left), no. 2 (middle), and no. 3 (right) of control group (black) and FSHD group (grey) for the SAA and SFE tasks cut up to 60° of humeral elevation. N equals the amount of participants within each clustered synergy. Clustered synergies are presented as a mean (black line) with standard deviation (bars) or ±95% confidence interval (grey area). BB: Biceps Brachii; DM: medial Deltoid; TB: Triceps Brachii; TD: Trapezius Descendens; TA: Trapezius Ascendens; PM: Pectoralis Major; SA: Serratus Anterior; LD: Latissimus Dorsi.

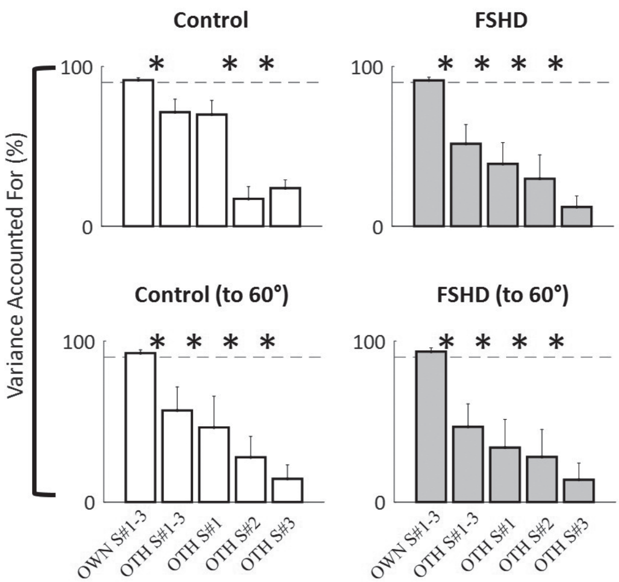


Figure 7. Variance accounted for as means and standard deviation from EMG cross-validation performed within the control (left) and FSHD group (right) for whole motion (top) and cut to 60(bottom). Bars show calculations using participants' own synergies (OWN), and from others (OTH) for a complete synergy set (S#1-3) and for individual synergies (S#1, S#2, S#3). Dashed line indicates the 90% VAF threshold. Significant differences were indicated by *.

3.4 DISCUSSION

The activities of eight superficial shoulder muscles were studied to investigate the changes in neuromuscular output in people with FSHD during humeral elevation. We hypothesized that the maximum activities of humeral elevator, scapular rotator and synergist muscles would be increased. This was confirmed for the Biceps Brachii (SFE task only). Contrary to what was expected the activity of the scapular rotators Trapezius Ascendens and Serratus Anterior (SAA task only) significantly decreased in people with FSHD. Additionally, it was hypothesized that the muscle synergies would be altered as a result of the impaired muscle functionality. Along this line, the synergies between groups were found to differ in importance for the complete motion: within group similarity indicated that controls mostly used one synergy for both tasks while the majority of the FSHD group required two task-specific synergies. From the comparable kinematic data and maximum muscle activities it can be concluded that while synergy weights were similar up to 60° humeral elevation, the Trapezius Ascendens and Serratus Anterior contributed on average less to humeral elevation in the FSHD than in the control group. The irregularity of variances accounted for per task by each synergy and the difference in maximum muscle activities, and synergy weights and coefficients, suggest the presence of participant-specific adaptation mechanisms.

The muscle activities of the control group for medial Deltoid (40-69%), Serratus Anterior (60-65%), Trapezius Ascendens (45-60%), Trapezius Descendens (35-55%), Latissimus Dorsi (10-23%), and Pectoralis Major (5-20%) during shoulder abduction-adduction or flexion-extension were consistent with other literature findings [4, 71, 72]. Maximum elevation angles in the control and FSHD group were also in line with a comparable study [4], while trends in increased activity of Trapezius Descendens and Pectoralis Major found by others [4] were not significant in this study. This could be ascribed to the large variability in muscle activation of people with FSHD [87], and partly to methodological differences in the MVCs protocol used for the Trapezius Descendens. In this study a strap over the shoulder was used to limit the participants' movements during the Trapezius Descendens MVCs recordings, while in others [4] the participants' shoulders were manually restrained.

The activities of the lower scapular rotator muscles during humeral elevation tasks in FSHD are presented for the first time in this study. The decreased activities of Trapezius Ascendens and Serratus Anterior muscles reveal that these scapular lateral rotators generated a lower force and thus a lower moment to rotate the scapula, a movement which is necessary during humeral elevation [73]. This insufficiency was confirmed by visual observations of very limited scapular rotation in the FSHD group. The decreased activity of these muscles appears to be a characteristic signature of the FSHD disease, which is in contrast with an increased activity of Trapezius Ascendens and Serratus Anterior found in shoulder impingement and post-stroke patients [74-79]. Ultimately, the inability to laterally rotate the scapula leads to a decrease in humeral elevation. This situation could produce unnecessary stress on the rotator cuff muscles, which provide a stabilizing function of the glenohumeral head and are preserved in FSHD individuals, based on MRI evidence [69, 70]. The increased synergist Biceps Brachii activity likely assisted in the stabilization of the humeral head and the elevation of the humerus within the decreased range of scapular motion [88]. However, a larger variability in muscle contributions did not reveal a clear relationship between the activity of lower scapular rotators or synergist muscles and the amount of humeral elevation.

At the level of intra-task differences between SAA and SFE, a significant increased activity in the FSHD group was found for the Serratus Anterior and Pectoralis Major while an increased activity trend occurred for the Trapezius Ascendens muscle. The higher activity of the Pectoralis Major is consistent with the greater abduction moment required during forward flexion. Furthermore, more scapulothoracic internal rotation is known to occur in healthy shoulders during shoulder flexion-extension than abduction-adduction [73], while external rotation of the scapula increases following Serratus Anterior fatigue [89]. A higher activity of the Trapezius Ascendens and Serratus Anterior during shoulder flexion-extension is therefore consistent with the requirements for more internal scapula rotation and joint stability.

In order to understand whether the coordinated activity, i.e. synergy weights, of selected muscles underlines possible compensatory strategies in the FSHD group, a muscle synergy analysis was carried out and presented here for the first time in this population. The synergies accounting for the highest proportion of the VAF (Figures 4 and 5) showed a changed coordinating action of humeral elevator and scapular rotator muscles. Specifically, synergy #1 for the control group was most likely responsible for glenohumeral elevation, scapula rotation and scapula stabilization, as exemplified by the main contributions of the Deltoid Medial, Trapezius Descendens and Ascendens, Serratus Anterior, and Latissimus Dorsi muscles. Synergy #1 for the FSHD group showed involvement of the Deltoid Medial and Trapezius Descendens and was therefore most likely responsible for glenohumeral elevation and scapula upward rotation. Contributions from the Trapezius Ascendens, Serratus Anterior, and Latissimus Dorsi muscles appeared diminished compared to the control group, reflecting the differences found in maximum muscle activity. The controls' second synergy was characterized by low muscle activation and follows from the methodological choice of accounting for >90% variance of all muscles. We postulate that this second synergy is a collection of short activation bursts (<20%) from different muscles, possibly to stabilize or facilitate the movement. Eight out of eleven FSHD participants used a second synergy with distinct coefficients for the SFE task. This second synergy was most likely responsible for scapula rotation and stabilization. This synergy also differs from the first in the contributions from Trapezius Ascendens and Serratus Anterior, reflecting the task-specific differences found in maximum muscle activity. Additionally, FSHD participants who applied this second synergy had higher humeral elevation angles. Similarly to what was found for the maximum activity, no clear relationships was present between the humeral elevation angles achieved and the amount of required synergies and/or involvement of scapula rotator/stabilizer muscles. The variety in muscle synergies compositions shows evidence that muscle control is less consistent in FSHD.

The synergy coefficients of similar synergy weights correlated poorly between groups. However, considering that the similarities were computed between two synergies of a high VAF proportion (FSHD) versus a clustered synergy of a low VAF proportion (control), it is questionable whether the comparison is representative of the change at a group level. Additionally, EMG cross-validations indicated a larger data similarity within the controls than the FSHD group, revealing a higher consistency in muscle activation in the former group. Future analysis should specifically focus on evaluation of within-group similarity of synergy weights and coefficients. In summary, coordination differences in FSHD appear to reflect the physiological changes of muscles due to the disease.

Based on the above findings it is evident that FSHD can lead to alterations in the coordination of muscle groups and lead to altered function and thus performance in some individuals. Muscle strengthening therapy, including scapular control, is sometimes considered part of the rehabilitation treatment for impingement and scapular winging [90-92]. Although scapular control therapy remains a debated topic [93], future research should explore whether this therapy could be effective in people with FSHD [87, 94]. Given the limited function of the scapular rotators, it is likely that therapeutic decisions should be made on an individual level, after careful assessment of the muscles' coordination using a methodology similar to the one proposed in this study.

3.4.1 Limitations

Muscle synergy analysis was shown to be sensitive to detect changes in motor output with respect to internal/external factors, however the technique has its limitations. The outcomes can be influenced by the choices made in EMG processing, NNMF settings, and threshold of VAFs [95-98]. For example, a lower VAF threshold would reduce the required number of synergies, possibly oversimplifying the motor output. To overcome this problem, this study uses two thresholds to ensure the variance of all muscles have been accounted for on a collective as well as a singular level [17]. In addition, the statistical approach was thorough and ensured that the limitations did not affect the conclusions.

The number of muscle synergies were inconsistent between participants and resulted in two clustered synergies of eleven and nine participants. However, this can be explained by individual characteristics, unrelated to disease effects [95, 96]. Furthermore, the total number of synergies were equal between the groups. Nonetheless, this could have resulted in the large within-group variances, specifically in muscle synergy weights, where a common coordinating activity is only evident for selected muscles [99]. The presented clustering method is suitable for simple movements as examined in this study, but arguably not when multiple synergies are needed, for example during more complex motions. Other cluster analysis methods can be used to pool synergies based on more distinct weights [22, 23] and are recommended in future research.

3.5 CONCLUSION

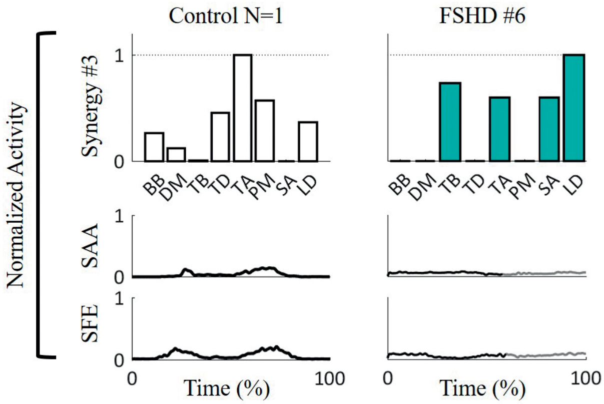
People with FSHD showed motor output alterations during humeral elevation, which were often movement- and participant-dependent. In general, the lower scapula rotators showed decreases in activity, with compensatory increase of a synergistic upper arm muscle. A group*muscle*task interaction effect was accompanied with increased activities of the lower scapula rotators, and synergistic chest and upper arm muscles during shoulder flexion-extension compared with abduction-adduction. The large group variances indicate that individual characteristics have a large influence on motor output. An assessment of the muscles' coordination is recommended to reveal individual synergies and to design evidence-based therapy for the management of the condition.

3.6 APPENDIX

Appendix table 1. Maximum voluntary contraction protocol

Muscle	Instructions
Biceps Brachii short head (BB)	Position: Upper arm is alongside the torso. Elbow is flexed at 90° and forearm temporarily supported by a researcher. A strap on the wrist prevents elbow flexion. Execution: Flex the elbow against the strap.
Medial Deltoid (DM)	Position: Upper arm is abducted at 60°. Elbow is flexed at 90° with hand palm downwards. A strap on the upper arm prevents abduction. Execution: Abduct the arm against the strap.
Triceps Brachii long head (TB)	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Extend the forearm against the resistance provided by a researcher.
Trapezius Descendens (TD)	Position: Upper arm is alongside the torso. Elbow is fully extended. A strap is placed above the shoulder and medial to the acromion. Execution: Pull the shoulder upwards against the strap.
Trapezius Ascendens (TA)	Position: Both upper arms are alongside the torso. Elbows are fully extended. Execution: Bend the trunk slightly forward and elevate the arms to form a straight line from the fingertips to the hips.
Pectoralis Major clavicular head (PM)	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Adduct the arm to the sagittal plane against the resistance provided by a researcher.
Serratus Anterior 5-6th rib (SA)	Position: Upper arm is alongside the torso. Elbow is fully flexed. Execution: Pull the shoulder down by pushing the elbow towards the hip against the resistance provided by a researcher.
Latissimus Dorsi (LD)	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Adduct the upper arm against the resistance provided by a researcher.

MVCs were executed while seated, with two repetitions and 2 minutes rest between repetitions.



Appendix figure 1. Participant-specific muscle synergy of a control participant (black, 9% VAF) and FSHD participant #6 (dark cyan, 19% VAF) for the SAA and SFE tasks. N equals the amount of participants within the clustered synergy. Individual synergy coefficients show upward (black line) and downward motion (grey line). Participant #6 has three synergies. BB: Biceps Brachii; DM: medial Deltoid; TB: Triceps Brachii; TD: Trapezius Descendens; TA: Trapezius Ascendens; PM: Pectoralis Major; SA: Serratus Anterior; LD: Latissimus Dorsi.

CHAPTER 4

The effects of Facioscapulohumeral Dystrophy and dynamic arm support on upper extremity muscle coordination in functional tasks

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4.1 INTRODUCTION

Facioscapulohumeral Dystrophy (FSHD) is considered one of the most prevalent neuromuscular disorders with an estimated two thousand affected individuals in The Netherlands in 2010 [33]. FSHD is characterized by progressive loss of muscle strength, mostly in the shoulder area, increased fatigue, pain, and joint stiffness [1, 34, 100-102]. People with FSHD have difficulties performing activities of daily life (ADL) and often show compensatory strategies requiring increased effort and energy [4]. Dynamic arm support devices compensate for gravity and consequently improve ADL performance for people with muscular weakness [6, 103]. However, Heide et al. a discontinuous use of dynamic arm support devices is reported in the majority of user studies [6], which implies a suboptimal use of these devices. A better understanding of how dynamic arm support devices influence body functions and ADL may contribute to their further development and increase usage rate [104].

Muscle weakness of the shoulder girdle significantly limits the ability of people with FSHD to perform independent ADL [10, 105]. Bergsma et al showed that eating, drinking, and reaching are severely limited in these persons, with ~42% experiencing extreme difficulties to reach forward at shoulder level and ~80% to reach over their head [34]. These limited activities are generally accompanied by increased muscle activities of biceps, deltoid, trapezius, and pectoralis muscles, which are ~3-5 times higher than in healthy individuals [4]. In addition, FSHD affects muscle coordination of the shoulder girdle during arm lifting, resulting in a reduced contribution to scapular upward rotation by the trapezius ascendens and serratus anterior up to 41% [58, 106]. Typically, a dynamic arm support provides an enhanced ability to reach and repeatedly lift the arm during ADL such as personal care, eating, and drinking [14, 15, 35, 103]. However, a dynamic arm support can also induce mechanical constraints resulting in longer movement time and altered smoothness [13, 15, 37, 42]. The support device may thus influence shoulder muscle coordination, potentially leading to a destabilizing effect on the glenohumeral joint. This can occur as the upward force imposed by the device may demand a greater effort by the glenohumeral joint muscles, thus also leading to long-term fatigue. Therefore, examining the effects of muscular weakness and the adaptations that may occur from the use of an arm support device during ADL is important to understand the long-term implications of such devices.

In the current study, we use muscle synergy analysis to quantify the changes in muscle coordination while using a dynamic arm support. Research has shown that the central nervous system controls groups of synergistic muscles to solve the motor redundancy problem [17]. Muscle synergy analysis simplifies the representation of muscle coordination patterns to a lower dimensional spatiotemporal output of synergistic contributions (weights) and activation patterns (coefficients) [17]. Four parameters are commonly used to quantify the variability and alterations in muscle coordination: 1) the number of muscle synergies required, 2) the variances accounted for per synergy, 3) synergy similarities between groups or conditions, and 4) synergy consistency within the same group or condition [13, 15, 20, 21, 24, 58, 74, 99, 107, 108].

It has been shown that dynamic arm support devices have little influence on the muscle coordination of healthy (older) participants, regardless of the level of support [13, 15, 24]. In people with FSHD, it can be speculated that an arm support would alter the selection of synergistic shoulder elevation muscles over time in a more pronounced way than in healthy persons. Moreover, overcoming the additional external force resulting from the gravity compensation device, could require altered synergies compared to healthy persons, due to the muscle weakness of the arm adductors in persons with FSHD. Movement performance, e.g. task duration and movement smoothness would consequently also be affected. However, muscle synergies in FSHD persons using a support remain unclear at present.

Altered muscle coordination patterns in persons with FSHD, following the use of an arm support device, may influence factors such as fatigue or susceptibility to injury, which are likely to influence usage rates. Knowledge of how muscular weakness and arm support devices influence muscle coordination and activation is needed for the continued development of such devices. Therefore, the aim of this study is to investigate the effect of muscular weakness in persons with FSHD, as well as the effect of a dynamic arm support on muscle coordination and activity during ADL tasks [104]. Furthermore, the effect of the dynamic arm support on the movement execution is quantified. Our primary hypothesis is that, muscle coordination when performing ADL without the arm support is less consistent within the FSHD group than within the healthy control group and is influenced by the type of ADL performed. Our secondary hypothesis is that using a dynamic arm support results in a more consistent muscle coordination, with a larger increase in consistency within the FSHD group compared to the healthy control group. Thirdly, we also hypothesize that using the arm support would lead to more similar synergies, i.e. muscle coordination would become more similar between the two groups.

4.2 MATERIAL AND METHODS

4.2.1 Participant characteristics and inclusion criteria

Data were collected from participants with FSHD and healthy controls in a larger study approved by the central Medical Ethical Committee of the University Medical Center Groningen (NL55711.042.15). The study was conducted in accordance with the guidelines of the Helsinki protocol. Participants were aged between 18-75 years, able to read and understand Dutch, and able to give written informed consent. Additionally, people with FSHD were included if they were able to transfer from a wheelchair to a chair (including with manual assistance), and had a Brooke scale score of 3 or 4. Healthy participants were excluded if they had any pathologies, shoulder pain, or a history of severe trauma of the shoulder <2yrs (e.g. fracture, luxation) that could interfere with the measurement results. Exclusions criteria for participants with FSHD were as follows: comorbidities that could interfere with the measurement results, previous surgery on the right shoulder, extrinsic causes of shoulder pain, a history of severe shoulder trauma, or an inability to elevate the right arm above 30°.

4.2.2 Tasks

The participants were seated in a chair with a left side-rest and lower back-rest and with the seat height set to achieve a knee flexion angle of 90°. Participants received detailed instructions regarding the movement before the execution of each task. Five tasks were chosen, according to the categories provided by Bergsma et al. 2017, [34], to reflect a selection of important ADL. Tasks were repeated three times in a randomized order. The tasks included 1) pushing and pulling (PP) an object, 2) simulated drinking with a cup of 200 grams (C2M), 3) simulated eating with a spoon (S2M), and 4) reaching towards a target at shoulder height on the ipsilateral side (ILR) and 5) on the contralateral side (CLR). Tasks PP, ILR, and CLR were performed at one shoulder width from the participant's midline to the respective side. Participants were allowed to rest for a few minutes between tasks and repetitions. All tasks and repetitions were first completed without the dynamic arm support and followed by a rest period of fifteen minutes. Successively, tasks were once more randomized and performed with the dynamic arm support. This sequence of task execution wo/w the device and incorporation of resting periods were to minimize fatigue and ensure protocol completion.

4.2.3 Dynamic arm support

The Gowing dynamic arm support (Focal Meditech BV, Tilburg, Netherlands) (Figure 1) provides spring-actuated passive support at the lower arm, where the spring tension is adjustable by motorized actuators [53]. The amount of support was personalized to simulate a gravity-free sensation and was constant within the reachable task workspace. Participants had no previous experience with a dynamic arm support device and were given up to 10 minutes of familiarization time prior to performing the tasks.

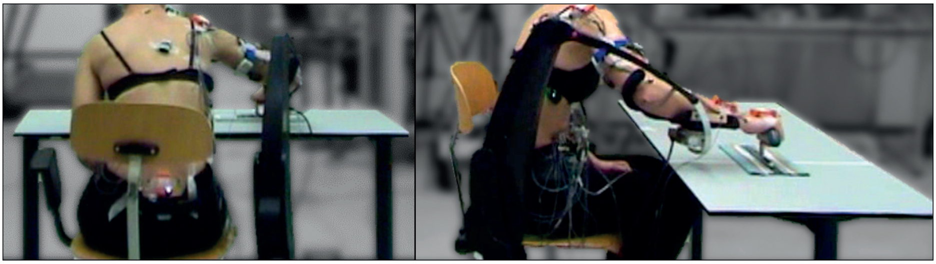


Figure 1. A participant with FSHD performing the push and pull task with the Gowing viewed from a posterior (left) and lateral (right) perspective.

4.2.4 Measurement and processing

Kinematics of the right hand, using an active marker placed on head of the 3rd metacarpal, were recorded at 100Hz using the Optotrak 3020 system and NDI First Principles application (Northern Digital Inc., Canada) [81] and used to calculate movement performance as in task duration, smoothness, and efficiency (see section 2.6 Kinematics). Surface electromyograms (EMG) were recorded for muscles on the right side, which included the prime humeral elevator/depressors and scapular rotator muscles, i.e. medial deltoid, pectoralis major clavicular head, latissimus dorsi, trapezius descendens, trapezius ascendens, and serratus anterior 5-6th rib, and the synergist muscles biceps brachii short head and triceps brachii long head. Data were captured at 2000Hz using the Delsys Trigno Wireless EMG system and EMGworks Acquisition application [19]. Skin was prepared and sensors were placed according to the Surface ElectromyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) guidelines [109].

Maximum voluntary contractions (MVCs) during isometric conditions were recorded beforehand (appendix table 1). The recorded EMG data were filtered with a 4th order Butterworth 20-450Hz bandpass and a 49-51Hz bandstop filter, rectified, smoothened with a 100ms moving window, normalized to the maximum amplitudes derived from all MVC and task recordings, and filtered with a 4th order Butterworth 5Hz low pass filter. The maximum task-specific muscle activity was extracted as highest normalized amplitude over all task repetitions. Time was normalized to 1001 samples for each repetition ranging from 0 to 100%.

In addition, force output of the shoulder elevators, humeral elevators, and elbow flexors were measured with a load cell, AST KAP-S/KAP-E Force Transducer [25]], at 100Hz during the MVC recordings to evaluate muscle strength in the two groups. The load cell was attached to the chair to minimize the burden on the participants in terms of transfers and time. One researcher provided instructions to ensure the correct position and execution (Appendix Table 1) of respective isometric contractions for shoulder elevation (during the trapezius descendens recording), humeral elevation (during the medial deltoid recording), and elbow flexion (during the biceps brachii recording). Participants were instructed and encouraged to contract maximally for five seconds, which was repeated after two minutes rest. The force output was visually checked and extracted as the maximum force during these five seconds of both repetitions.

4.2.5 Muscle synergy extraction

EMG data were pooled for task repetitions per individual in a single matrix to investigate the muscle synergies between the two groups and two support conditions within respective tasks. Muscle synergies were extracted using non-negative matrix factorization, which decomposed the matrix into 1 to 8 sets of components consisting of weights and coefficients [17]. The weights and coefficients were converted to a unit vector and represent normalized muscle activity (0-1). Furthermore, the non-negative matrix factorization provided the percentage of variance accounted for of all muscles and per individual muscle for each set of synergies. At least 90% of all muscles' and >75% of individual muscles' variance should be accounted for before a set of synergies was considered to represent muscle coordination. The synergies were then clustered based on the Pearson's correlation coefficients (r) calculated between all possible combinations of individual participants' synergy weights within respective group, support condition, and task [85, 108]. Clustered synergies, which represented the muscle coordination of a group, for a support condition and a task, were then ranked in an ascending order (MS1-4) based on the number of participants in each cluster.

Subsequently, in a leave-one-out process [99, 107] synergy consistency, which refers to correlations within clustered synergies, was calculated as the Pearson's correlation coefficients between the synergy weights of individual participants included in the cluster and the mean synergy weights of the cluster without that given participant. Furthermore, synergy similarity, which refers to correlations between clustered synergies, was calculated as the Pearson's correlation coefficients between the synergy weights of individual participants of one cluster and the mean synergy weights of another cluster within respective tasks [85]. For similarity calculations between groups, FSHD individuals were compared with the mean of controls, and between support conditions, individual synergy weights while supported were compared with the mean while unsupported. Furthermore, the synergy consistency and similarity calculations were restricted to 1) equally-ranked clustered synergies and 2) the first (MS1) and second (MS2) ranked synergies, since these account for the majority of the EMG variance, which were on average >50% and >33%, respectively.

4.2.6 Kinematics

Indicators of movement performance, as in task duration, smoothness, and efficiency, were calculated similarly as in Pirondini et al. 2016 [13]. Task duration was calculated as time in seconds between start and end of movement. Smoothness was calculated as the median jerk of the finger marker. Efficiency was calculated as the root mean square error between the trajectory of the finger marker and straight lines between start, target, and end, and normalized for length where i represents one sample and n represents the entire data set:

$$efficiency = \frac{\left(\frac{\sqrt{\sum_{i=1}^{n}\left(\left(finger\ trajectory_i - straight\ line_i\right)^2\right)}}{\sqrt{\sum_{i=1}^{n}\left(straight\ line_i{}^2\right)}}\right)}{n} \quad (1)$$

The start, target, and end positions were determined as the respective positions where velocity was closest to zero. Task duration, smoothness, and efficiency values towards zero represent a fast, smooth, and efficient movement, respectively.

4.2.7 Statistical analysis

Ten parameters were extracted to investigate the effect of FSHD and a dynamic arm support on muscle coordination with respect to movement performance (Table 1). Our first hypothesis is that synergies in the FSHD group are less consistent than in the control group. Our second hypothesis is that a dynamic arm support results in more consistency, thus the synergy consistency of supported tasks should be higher than unsupported tasks. To test these two hypotheses, a non-parametric analysis of variance [110] was performed on the synergy consistency of the first and second ranked synergies, with population as between group factor and support conditions and tasks as within group factor (α = 0.05). The Pearson's correlation coefficients were first transformed with the Fisher's z-transformation formula [111] to normalize the sampling distribution:

$$Fisher's\ z\text{-}transformation = 0.5 * \ln\left(\frac{(1+r)}{(1-r)}\right) \quad (2)$$

Table 1. Overview of study outcome parameters with input data, conversion method, and unit.

Biomechanical characteristic	Input data	Outcome parameter	Method	Unit
Muscle coordination	EMG	Number of synergies [2]	Non-negative matrix factorization	#
		Synergy weight consistency [1]	Pearson correlation coefficients within clustered synergies	r
		Synergy weight similarity [1]	Pearson correlation coefficients between clustered synergies	r
Muscle activity	EMG	Muscle activity [2]	Maximum amplitude	0-1 MVC
Movement performance	Kinematics	Task duration [2]		seconds
		Smoothness [2]	Jerk	mm/s^3
		Efficiency [2]	Root Mean Square Error	mm
Muscular weakness	Kinetics	Shoulder elevation strength [2]	Maximum force output	N
		Humeral elevation strength [2]	Maximum force output	N
		Elbow flexion strength [2]	Maximum force output	N

Numbers represent the primary (1) and secondary (2) outcome parameters. EMG: ElectroMyoGrams, MVC: Maximum Voluntary Contraction.

In addition, Cohen's d [83] was calculated from the Pearson's correlation coefficients of the groups' mean (M) and standard deviations for both groups (SD):

$$Cohen's\ d = [\frac{(M2-M1)}{SD1}, \frac{(M2-M1)}{SD2}] \quad (3)$$

The Cohen's d was then corrected with the unbiased d formula as Hedges' g [112], which includes sample sizes for the correction (df):

$$Hedges'g = Cohen's\ d * (1 - \left(\frac{3}{4*df-1}\right) \quad (4)$$

The Hedges' g (Hg) was interpreted as very small (0.00- 0.01), small (0.01 - 0.20), medium (0.20 - 0.50), large (0.50 - 0.80), very large (0.80 - 1.20), and huge (>1.20) [58, 83].

4

Post-hoc analyses were performed for significant effects in the analyses of variance, with the exception of task effects, as a Wilcoxon signed rank test for related samples and Wilcoxon rank sum test for unrelated samples. Tasks were pooled and alpha levels were corrected accordingly using the Bonferroni method and set to 0.01.

To test our third hypothesis, that muscle coordination would become more similar between the two groups under the influence of an arm support, differences in muscle synergy similarity were investigated with a Wilcoxon rank sum test after the Fisher's z-transformation (formula 2). First, we compared the similarities calculated between the unsupported FSHDs and unsupported controls with the similarities calculated between the supported FSHDs and supported controls to test the effect of support. Second, we compared the similarities calculated between the unsupported FSHDs and unsupported controls with the similarities calculated between the supported FSHDs and unsupported controls to test the interaction effect of support and FSHD. Comparisons were performed respectively on the first and second ranked synergies. The Hedges' g was calculated as a range of effect size. Tasks were pooled and alpha levels were corrected accordingly using the Bonferroni method and set to 0.01.

Additional analysis was performed on the secondary outcome measures (table 1). The Hedges' g was calculated as a range of effect size for all supplementary analyses. In all analyses, tasks were pooled and alpha levels were corrected accordingly using the Bonferroni method and set to 0.01.

First, the number of extracted synergies were compared between FSHD and control group for support conditions with a Wilcoxon rank sum test and subsequently between support conditions respectively for the FSHD and control group with a Wilcoxon signed rank test.

Second, to evaluate the limitations of muscular weakness, the maximum muscle activity during tasks was tested with a non-parametric analysis of variance with population as between group factor and support conditions and tasks as within group factors ($\alpha = 0.05$). Post-hoc analyses were performed for significant effects in the analyses of variance, except for task effects, as a Wilcoxon signed rank test for related samples and Wilcoxon rank sum test for unrelated samples.

Third, to investigate the effect on task performance, a non-parametric analysis of variance was performed on the task duration, jerk, and efficiency ($\alpha = 0.05$). Post-hoc analyses were performed for significant effects in the analyses of variance, except for task effects, as a Wilcoxon signed rank test for related samples and Wilcoxon rank sum test for unrelated samples.

Fourth, to quantify muscular weakness the maximum voluntary force output during shoulder elevation, humeral elevation, and elbow flexion were compared between populations with a Wilcoxon rank sum test ($\alpha = 0.05$).

4.3 RESULTS

4.3.1 Participant characteristics

Twelve healthy control participants (6M/6F, 55.5±13.4yrs, 1.76±0.08m, 72±14kg, 11 right- & 1 left-handed) and twelve participants with FSHD (6M/6F, 56.0±14.5yrs, 1.76±0.10m, 75±20kg, 9 right- & 3 left-handed) were included in this study.

4.3.2 Muscle synergies

Up to four synergies were extracted per task where more than 70% of the participants generally required two synergies to perform a task (appendix figure 1). Synergies were less on average for the FSHDs than for the controls (unsupported p: 0.002, Hg: -0.59 to -0.49, and supported p: 0.003, Hg: -0.57 to -0.50) (appendix figure 1). The number of extracted synergies could not be found to differ per participant between support conditions. The clustered synergy weights for the contralateral reaching task are shown in Figure 2, while weights and coefficients for other tasks are shown in the appendix (appendix figures 2-10).

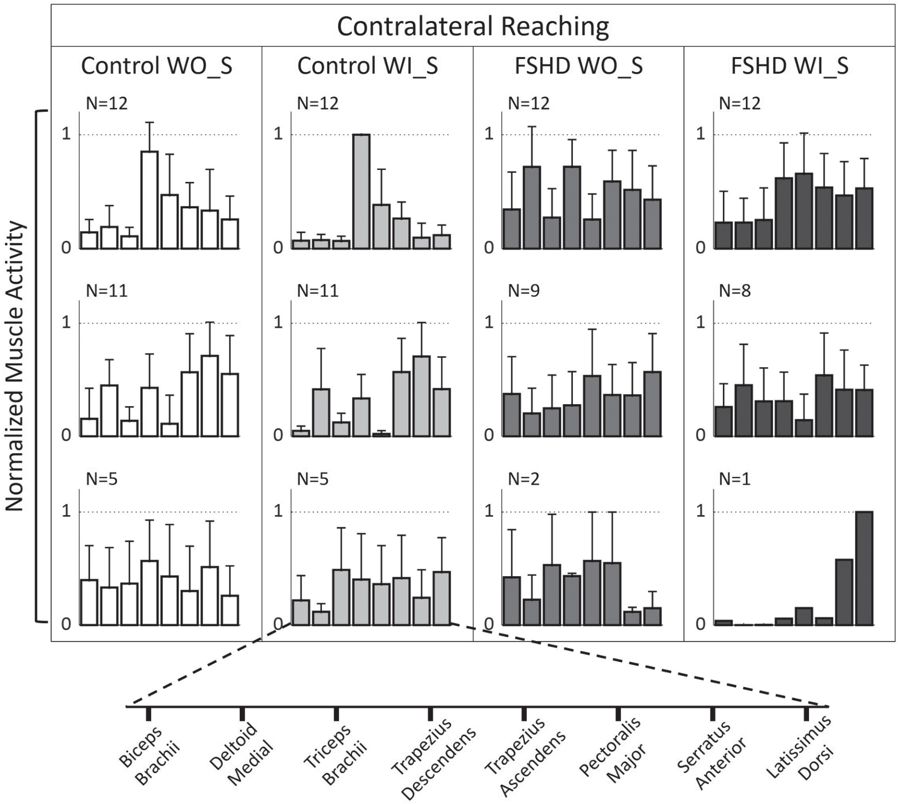


Figure 2. Clustered muscle synergy weights during contralateral reaching for control without support (white, WO_S), control with support (light gray, WI_S), FSHD without support (gray, WO_S), and FSHD with support (dark gray, WI_S) ranked horizontally in order of prevalence (N). Bars represent the mean amplitude and lines one standard deviation.

In the unsupported contralateral reaching task (Figure 2 and appendix figure 10), the controls' first ranked synergy involves scapular mobility and stabilization, mostly by the trapezius. In the second ranked synergy, similar functions are present, in particular upward scapular rotation by the trapezius descendens and serratus anterior and were more actively accompanied by up- and inward rotation and stability of the humerus by the deltoid, pectoralis, and latissimus dorsi, respectively. In FSHDs, the first ranked synergy resembles a merge of the controls' first and second ranked synergies. Their second ranked synergy represents a co-contraction that involves elbow flexion, scapular downward rotation, and humerus depression, mostly by the biceps, trapezius ascendens, and latissimus, respectively. In the supported contralateral reaching task, minor differences can be noted in controls while FSHDs present a shift in deltoid and trapezius ascendens contributions between the first and second ranked synergies.

4

4.3.3 Synergy consistency

A significant group effect for the weight consistency was found for MS1 and MS2, with FSHDs less consistent than controls in MS1 ($p<0.001$, task averaged Pearson's correlation coefficients r: -0.34, Hg: -1.48 to -0.98) and MS2 ($p<0.001$, r: -0.41, Hg: -1.12 to -1.08). In addition, a significant support effect was found for MS1, where consistency was generally higher in the supported tasks ($p<0.001$, r: +0.14, Hg: 0.41 to 0.47).

Furthermore, a significant group * support interaction effect was found for MS2, but not MS1 (p: 0.110), with FSHD less consistent than controls. For unsupported movements the difference in consistency between FSHDs and controls was r: -0.26 ($p<0.001$, Hg: -0.69 to -0.65) and for supported movements r: -0.54 ($p<0.001$, Hg: -1.61 to -1.60). Between supported FSHD and unsupported controls the difference was r: -0.39 ($p<0.001$, Hg: -1.15 to -0.96) and for unsupported FSHD and supported controls r: -0.42 ($p<0.001$, Hg: -1.24 to -1.08).

Moreover, a significant group * support * task interaction effect was found for MS1, while for MS2 the group * support interaction effect (p: 0.002), as explained above, and the support * task interaction effect (p: 0.009) were significant. Post-hoc analyses showed that MS1 of unsupported FSHDs was significantly less consistent than unsupported controls (Push and Pull, p: 0.002, r: -0.40, Hg: -1.64 to -1.31, Spoon to Mouth, p: 0.006, r: -0.36, Hg: -1.89 to -1.04) (Figure 3A), but there was no difference between supported FSHDs and unsupported controls. For MS2, there were no group differences while both were unsupported (Figure 3B), but supported FSHDs were significantly less consistent than unsupported controls (Push and Pull, $p<0.001$, r: -0.74, Hg: -3.55 to -2.32, Contralateral Reaching, p: 0.004, r: -0.39, Hg: -1.06 to -0.97). Furthermore, controls showed a significant increase in consistency from without support to with support in MS1 (Ipsilateral Reaching, $p<0.001$, r: +0.28, Hg: 1.03 to 4.92, Contralateral Reaching, p: 0.002, r: +0.25, Hg: 0.91 to 4.23) (Figure 3C) and in MS2 (Spoon to Mouth, p: 0.006, r: +0.40, Hg: 1.02 to 2.89, Ipsilateral Reaching, p: 0.002, r: +0.36, Hg: 0.83 to 1.21) (Figure 3D). There were no significant effects of support in the FSHD group. Additionally, significant task effects (MS1 and MS2) were found.

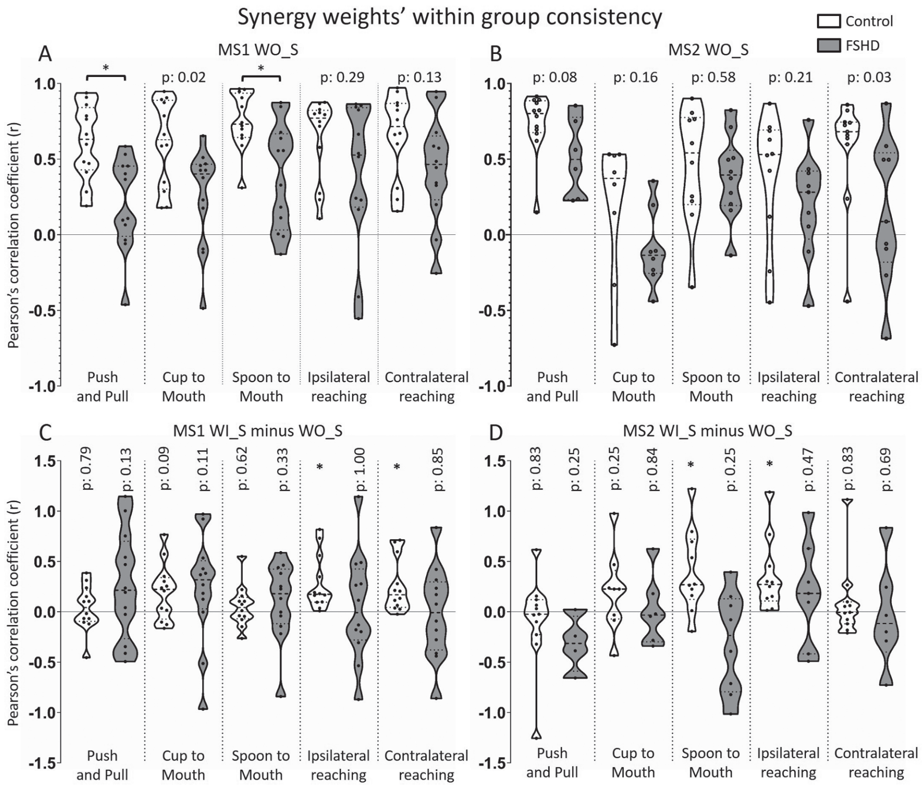


Figure 3. Synergy weights' within group consistency as Pearson's correlation coefficients (r) of MS1 (A, C) and MS2 (B, D) for without support (A, B) and with support minus without support conditions (C, D) presented for controls (white) and FSHDs (gray) as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. An asterisk indicates a significant difference between controls and FSHDs (A, B) or between the two support conditions for respective groups (C, D). WO_S: without support, WI_S: with support.

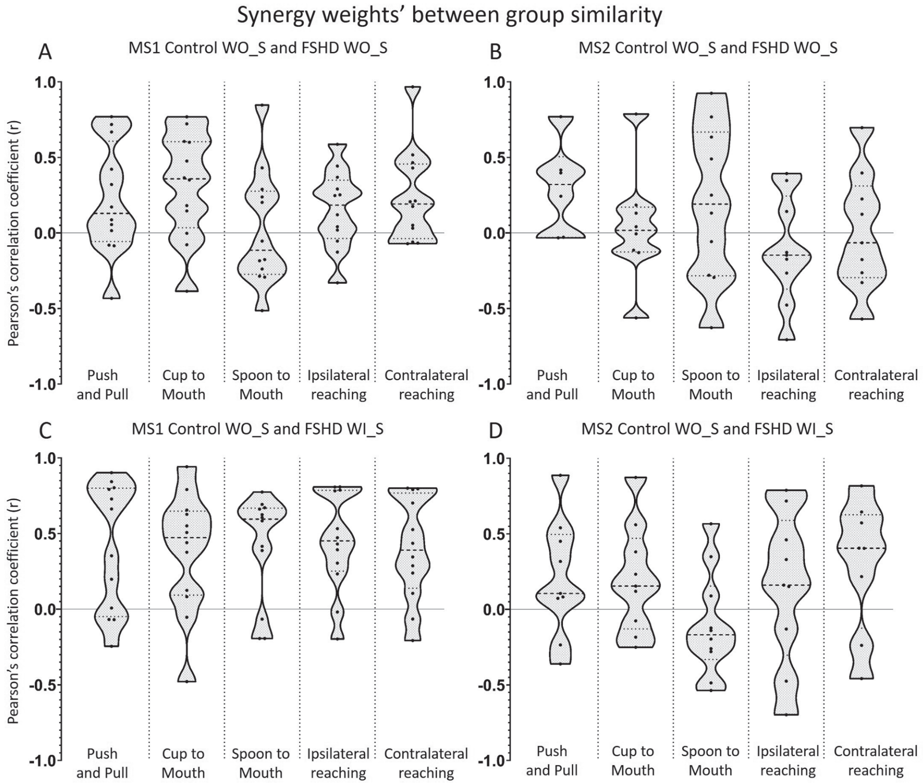


Figure 4. Synergy weights' between group similarity as Pearson's correlation coefficients (r) of MS1 (A, C) and MS2 (B, D) for without support in both groups (A, B) and without support control and with support FSHD (C, D) presented as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. WO_S: without support, WI_S: with support.

4.3.4 Synergy similarity

In unsupported movements it was found that the task averaged similarity between individual FSHDs and the mean of controls was r: 0.19 for MS1 and r: 0.07 for MS2 (Figure 4A-B). The similarity between the synergies without and with the support were r: 0.23 (MS1) and r: 0.00 (MS2) for FSHDs and r: 0.72 (MS1) and r: 0.52 (MS2) for controls (appendix figure 11). Furthermore, the similarity between FSHDs and controls significantly increased when FSHDs used a support while controls were unsupported for MS1 ($p<0.001$, r: +0.12, Hg: 0.59 to 0.65), but not for MS2 (p: 0.454, r: +0.03, Hg: 0.16 to 0.16) (Figure 4C-D). Finally, the similarity between FSHDs and controls significantly increased when both groups used a support compared with when both groups did not use a support for MS1 (p: 0.008, r: +0.12, Hg: 0.48 to 0.64), but not for MS2 (p: 0.409, r: +0.04, Hg: 0.17 to 0.19).

4.3.5 Maximum muscle activity

A significant group effect was found for the maximum muscle activities of the biceps, deltoid, pectoralis, and latissimus, while a significant support effect was found for biceps, triceps, serratus and latissimus. Finally, a significant group * support * task interaction effect was found for trapezius descendens, serratus, and latissimus (all $p<0.001$) with amplitudes of serratus and latissimus lower in controls in selected tasks with the support. Post-hoc analysis of the maximum muscle activity showed that the FSHDs had higher amplitudes than controls (all $p<0.001$) with biceps: +14% (Hg: 0.68 to 2.16), deltoid: +13% (Hg: 0.63 to 1.38), pectoralis: +13% (Hg: 0.68 to 1.47), and latissimus: +11% (Hg: 0.62 to 0.88) (appendix figures 12-13). Furthermore, activities during supported movements were lower for the biceps ($p<0.001$, -7%, Hg: -0.47 to -0.36), deltoid ($p<0.001$, -9%, Hg: -0.68 to -0.47), triceps (p: 0.003, -1%, Hg: -0.07 to -0.06), serratus ($p<0.001$, -6%, Hg: -0.54 to -0.37), and latissimus ($p<0.001$, -5%, Hg: -0.32 to -0.30). In addition, amplitudes of the serratus were significantly lower in controls due to support during all tasks except ipsilateral reaching (-15 to -5%, Hg: -1.55 to -0.43). Latissimus also showed significant lower muscle activity amplitudes (all $p<0.001$) in controls due to support during the spoon to mouth (-1%, Hg: -0.18 to -0.17) and contralateral reaching tasks (-12%, Hg: -0.97 to -0.60). In addition, all muscles, except for pectoralis and serratus, presented a significant task effect and the majority of muscles a significant support * task interaction effect.

4

4.3.6 Force output

The maximum voluntary force output was significantly lower in FSHDs than controls for shoulder elevation (p: 0.002, -166N and Hg: -1.82 to -1.21), humeral elevation (p: 0.032, -56N and Hg: -0.87 to -0.69), and elbow flexion (p: 0.004, -85N and Hg: -1.44 to -1.30), see also appendix table 2.

4.3.7 Movement performance

There was a significant support effect for task duration, efficiency, and jerk, and a significant group * support interaction effect for jerk, with longer task duration with support, reduced efficiency with support and reduced jerk with support in both groups (appendix figure 14). Additionally, there were significant task effects (task duration and jerk), significant group * task interaction effects (task duration and efficiency), and significant support * task interaction effects (efficiency and jerk) found.

4.4 DISCUSSION

4.4.1 Muscle coordination consistency and similarity

In this study, we investigated muscle coordination in persons with FSHD and healthy controls when performing ADL, without and with the use of a dynamic arm support device. Our first hypothesis was partially accepted, as without support muscle coordination was less consistent in FSHDs than controls for the first (MS1) and the second (MS2) ranked synergy. Moreover, while consistency was different per task within each group without support, it was not moderated by the type of task across groups since no significant group * task interaction was found. In addition, we partly confirmed our second hypothesis that a dynamic arm support resulted in a more consistent muscle coordination as this was found for controls (MS1 and MS2), but not for FSHDs. Furthermore, we partly confirmed the third hypothesis that synergies became more similar between the two groups when using an arm support for MS1, but not for MS2.

4.4.2 Muscular weakness in persons with FSHD

This is the first study to examine muscle coordination synergies in persons with FSHD during ADL. Our findings with regards to synergy weights during unsupported tasks are consistent with previous results in single joint arm elevation movements [58], revealing that muscle coordination in persons with FSHD remains heterogeneous during the ADL tasks used in this study. The nature of the unsupported task, that being whether it was close or away from the body, appears to influence the synergies' consistency of both the first (task effect) and the second ranked synergies (task effect and support * task interaction effect) within each group.

A clear-cut categorization of the synergies based on muscle function is not straightforward, but we made the following observations in control participants. During unsupported tasks, the first ranked synergy mostly involved the muscles responsible for elevation, rotation of the scapula, and arm adduction, while the second ranked synergy mostly involved those muscles responsible for scapula external rotation and arm abduction. Observation of the synergy weights in control participants reveals that, in unsupported tasks that are closer to the body, MS1 was characterized by a prominent involvement of the trapezius muscle, in tasks far away from the body, the deltoid was also involved. In MS2, the trapezius, serratus, and latissimus were largely involved during unsupported tasks that were closer to the body (cup and spoon to mouth), with the deltoid becoming additionally involved in unsupported tasks away from the body (reaching). In the push and pull task, which consisted of a reach and retrieval phase, the trapezius was not involved. The involvement of the trapezius and serratus during far away from the body tasks, where arm elevation was necessary to reach the target, is consistent with the functional anatomy. Both these muscles are in fact necessary to accomplish scapular lateral rotation [113].

In FSHD participants, a higher level of muscle co-contraction compared to controls was present for all muscle weights in both synergies during all unsupported tasks. This higher level of co-contraction in the FSHD weights was also accompanied by a higher variation in neural activation, as shown by the coefficients. Furthermore, higher maximum muscle activity was found for the biceps, deltoid, pectoralis, and latissimus dorsi compared to controls, which is in line with previous literature during unsupported tasks [4]. Despite these heterogeneous neuromuscular activations and the lower muscle strengths found in FSHDs, the movement performance indicators could not be shown to differ between the two groups. These findings indirectly highlight the existence of compensatory movement strategies in persons with FSHD that aid task completion but also lead to a greater muscle effort than controls.

4.4.3 The effects of dynamic arm support

The effects of a dynamic arm support on motor capacity, i.e. what a person can do in controlled settings, are reported for the first time in persons with FSHD. Knowledge of motor capacity and capability are important to assess how an arm support is used and ultimately to better understand the reasons a person may discontinue its use [57, 104].

When using the arm support, both groups displayed a reduction in maximum muscle activity of the biceps, deltoid, triceps, serratus, and latissimus. Yet, a more generalized co-activation was apparent in all muscles in the FSHD group. Internal consistency in this group was not significantly affected despite general alterations in muscle activity. The increased synergy similarity between the control and FSHD groups when using the support illustrates that the FSHD group did alter their synergies when assisted by the support device. The internal consistency, however, was lower in the FSHD than the control group and the group differences grew larger with the use of an arm support. These novel findings indicate that muscle coordination in persons with FHSD remains heterogeneous, which is likely the result of the individual-specific deficits in muscle strength.

Although the dynamic arm support facilitated arm elevation, the device also affected movement performance by restricting range of motion and increased movement duration in both groups. These findings are consistent with the existing literature in healthy adults and in stroke patients [13-15]. Future research should investigate the movement dynamics with the use of arm support devices, as it would be useful in persons with FSHD, to better understand how joint forces, moments, and powers are affected by the devices, in particular whether eccentric-concentric contractions are employed in response to the gravity compensation. Knowledge of these adaptations may have implications for a more efficient design of the arm support device and for the prevention of long-term neck-shoulder complaints, which are reported by more than 90% of adults with FSHD [102].

4.4.4 Limitations

The tasks were always completed first without device and then with device, which could have affected the results due to a higher chance of fatigue in the latter condition. However, resting periods, a randomization of the tasks, and repetitions were incorporated in both conditions to minimize fatigue and ensure that participants could complete the study.

The limited number of three repetitions per task, used in the current study, could have reduced the internal consistency in all cases, but would not have affected the number of synergies extracted [114]. Moreover, the number of repetition was experienced as very demanding by some FSHD participants and more repetitions would likely have resulted in discontinuation of the study.

The dynamic arm support, Gowing, imposed mechanical constraints that affected the performance of both groups. The elbow brace was experienced as a slight inconvenience during tasks away from the body, but did not hinder the participants in task execution. Participants could have compensated for the inconvenience, resulting in a more variable execution, but there were no indications noted in the outcome parameters.

4.4.5 Considerations for future research

Scapular kinematics and activity of deeper-layered muscles, such as the rhomboids, teres, and supra- and infraspinatus, should be considered in future research to better understand the effect of arm supports on scapular mobility and glenohumeral stabilization.

Future research should also consider the long-term effects of using a dynamic arm support device in a home environment to uncover potential benefits and disadvantages associated with regular, home use. The positive effects on motor capacity from the current study might also be reflected in long-term benefits in motor performance by experienced dynamic arm support users during repetitive tasks [103]. Negative effects, such as discomfort or inconvenience to perform certain ADL, may add to the evidence for discontinued use of the support device.

4.5 CONCLUSION

We found that muscle coordination is altered and less consistent in FSHDs compared with healthy controls. An arm support alleviated muscle efforts and affected muscle coordination in both populations by facilitating arm elevation. Consequently, the populations became more similar, yet, the internal consistency of FSHDs remained unaffected and lower than that of healthy controls. This is likely the result of the individual-specific deficits of muscle weakness and respective development of compensatory strategies for dealing with the compensation of gravity and movement constraints. The biomechanical consequences of using an arm support should be further investigated in people with FSHD on deeper-layered shoulder muscles and to evaluate potential long-term benefits and disadvantages.

4.6 APPENDIX

Appendix table 1. Maximum voluntary contraction protocol.

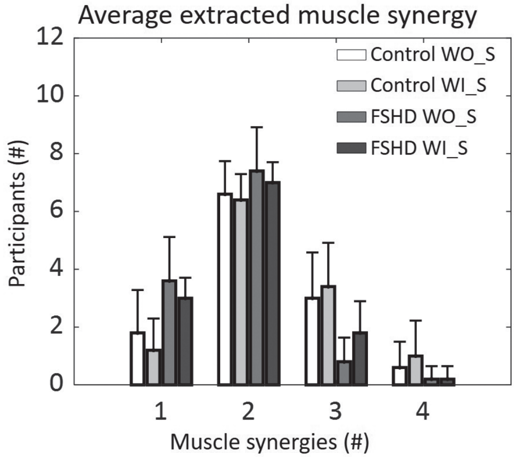
Muscle	Instructions
Biceps Brachii	Position: Upper arm is alongside the torso. Elbow is flexed at 90°. A strap on the wrist prevents elbow flexion. Execution: Flex the elbow against the strap.
Medial Deltoid	Position: Upper arm is abducted at 60°. Elbow is flexed at 90° with hand palm downwards. A strap on the upper arm prevents abduction. Execution: Abduct the arm against the strap.
Triceps Brachii	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Extend the forearm against the resistance provided by a researcher.
Trapezius Descendens	Position: Upper arm is alongside the torso. Elbow is fully extended. A strap is placed above the shoulder and medial to the acromion. Execution: Pull the shoulder upwards against the strap.
Trapezius Ascendens	Position: Both upper arms are alongside the torso. Elbows are fully extended. Execution: Bend the trunk slightly forward and elevate the arms to form a straight line from the fingertips to the hips.
Pectoralis Major	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Adduct the arm to the sagittal plane against the resistance provided by a researcher.
Serratus Anterior	Position: Upper arm is alongside the torso. Elbow is fully flexed. Execution: Pull the shoulder down by pushing the elbow towards the hip against the resistance provided by a researcher.
Latissimus Dorsi	Position: Upper arm is abducted at 90°. Elbow is flexed at 90° with hand palm downwards. Execution: Adduct the upper arm against the resistance provided by a researcher.

Maximum voluntary contractions were executed while seated, with two repetitions and 2 minutes rest between repetitions.

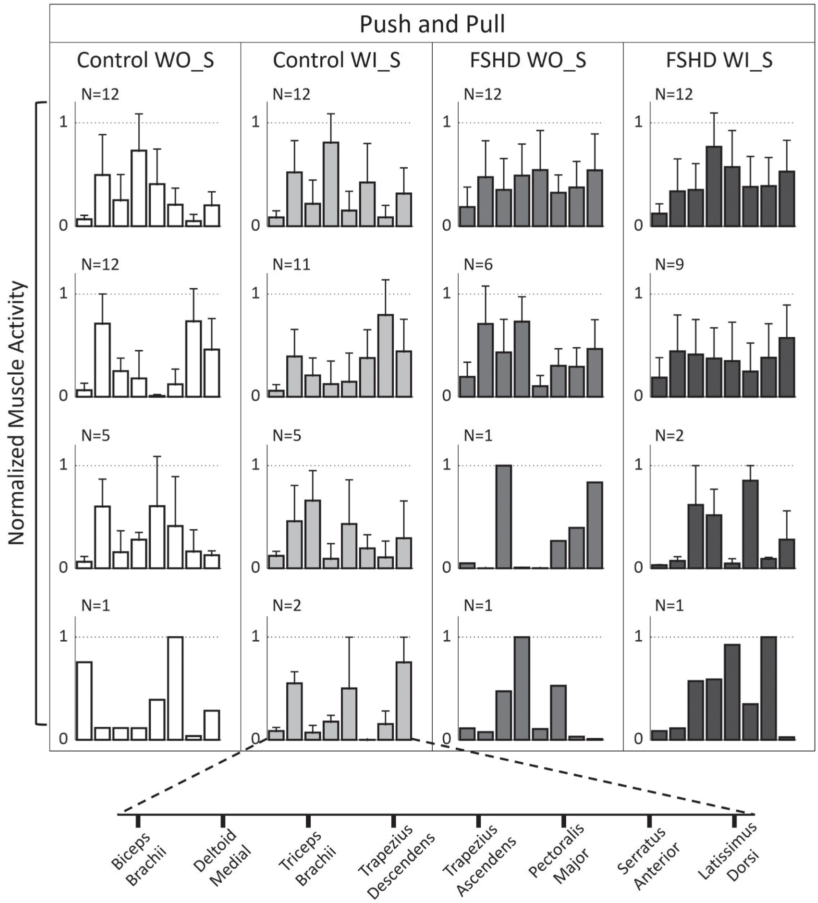
Appendix table 2. Maximum force output

Population	Shoulder elevation (N)	Humeral elevation (N)	Elbow flexion (N)
Control	408±128	174±76	224±61
FSHD	242±85	118±60	139±55

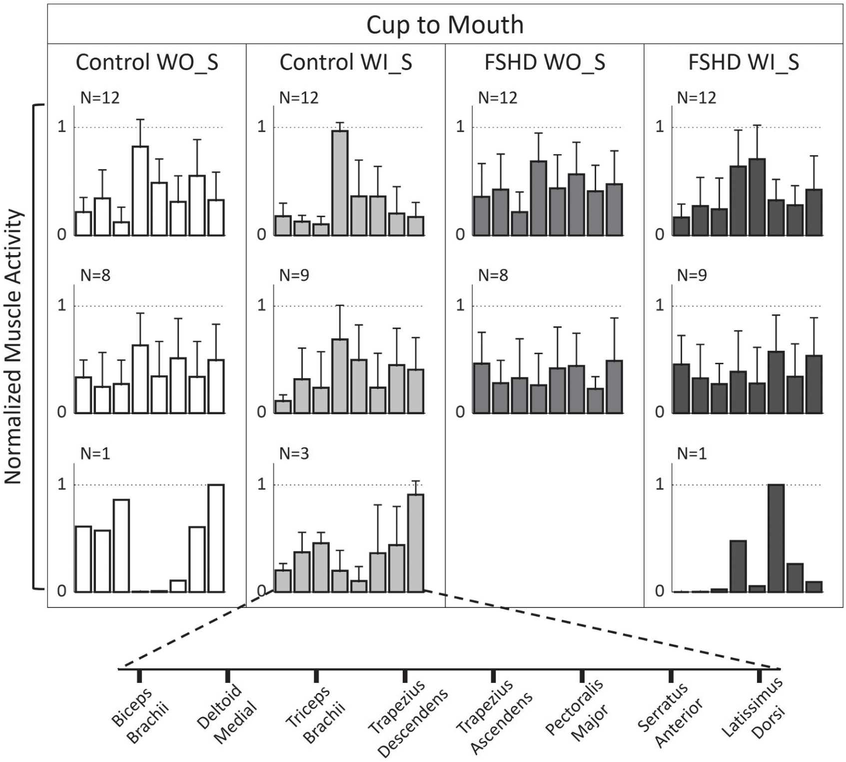
Maximum force output for shoulder elevation, humeral elevation, and elbow flexion of controls and FSHDs is presented as mean and one standard deviation.



Appendix figure 1. Amount of extracted muscle synergies for controls without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) for all tasks. Bars represent the mean and the vertical lines plus one standard deviation. WO_S: without support, WI_S: with support.



Appendix figure 2. Clustered muscle synergy weights during push and pull for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Bars represent the mean amplitude and lines one standard deviation. WO_S: without support, WI_S: with support.



Appendix figure 3. Clustered muscle synergy weights during cup to mouth for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Bars represent the mean amplitude and lines one standard deviation. WO_S: without support, WI_S: with support.

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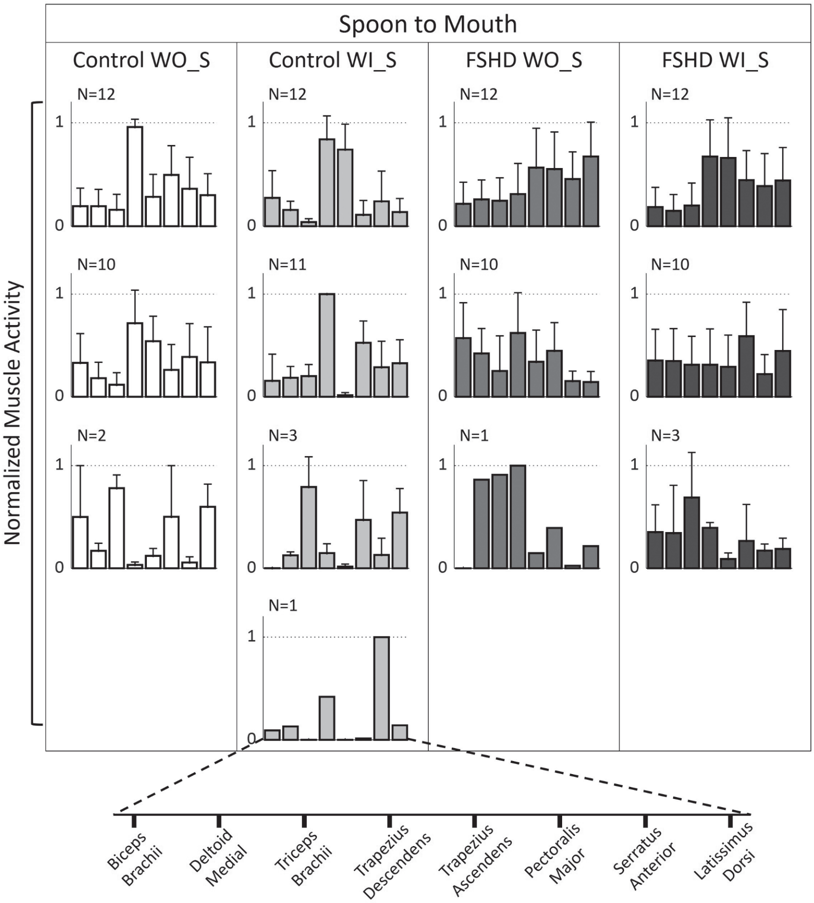
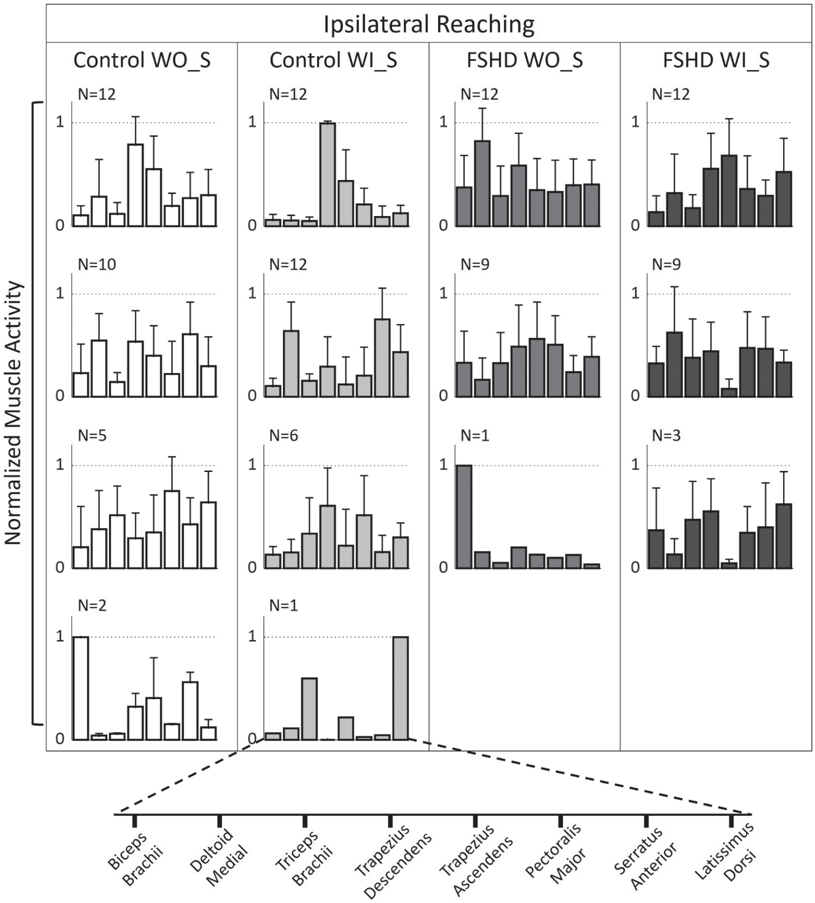


Appendix figure 4. Clustered muscle synergy weights during spoon to mouth for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Bars represent the mean amplitude and lines one standard deviation. WO_S: without support, WI_S: with support.



Appendix figure 5. Clustered muscle synergy weights during ipsilateral reaching for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Bars represent the mean amplitude and lines one standard deviation. WO_S: without support, WI_S: with support.

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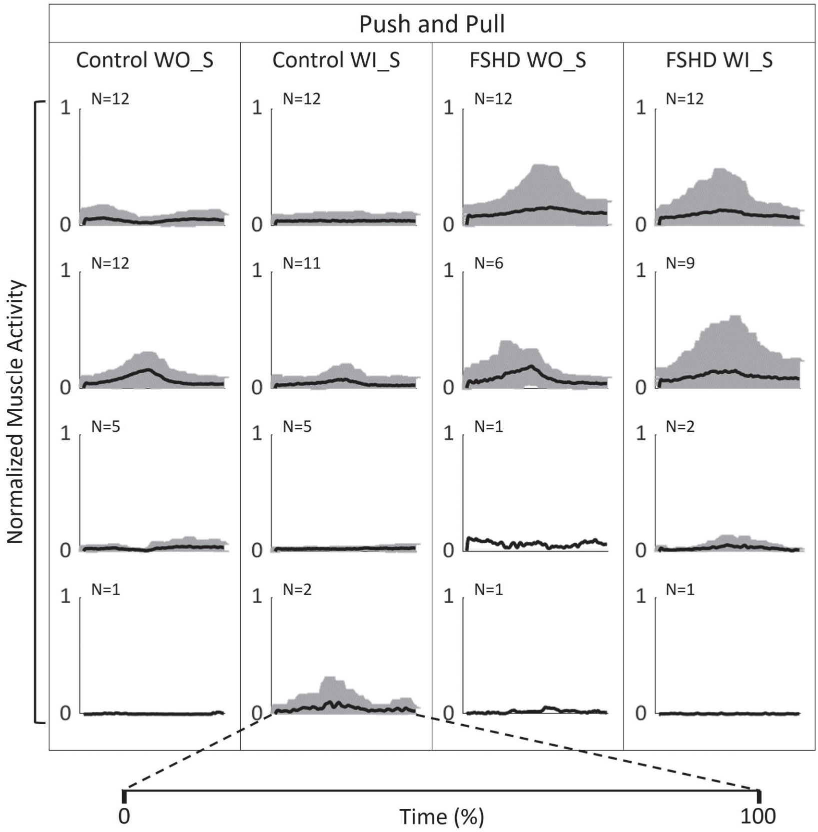
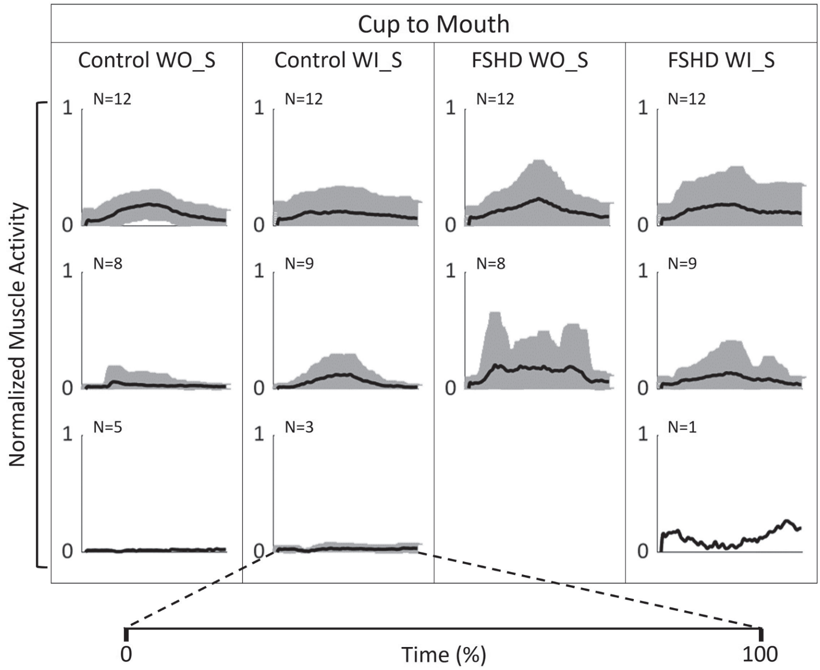


Appendix figure 6. Clustered muscle synergy coefficients during push and pull for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Thick black line represent the mean amplitude and gray area the 95% confidence interval. WO_S: without support, WI_S: with support.



Appendix figure 7. Clustered muscle synergy coefficients during cup to mouth for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Thick black line represent the mean amplitude and gray area the 95% confidence interval. WO_S: without support, WI_S: with support.

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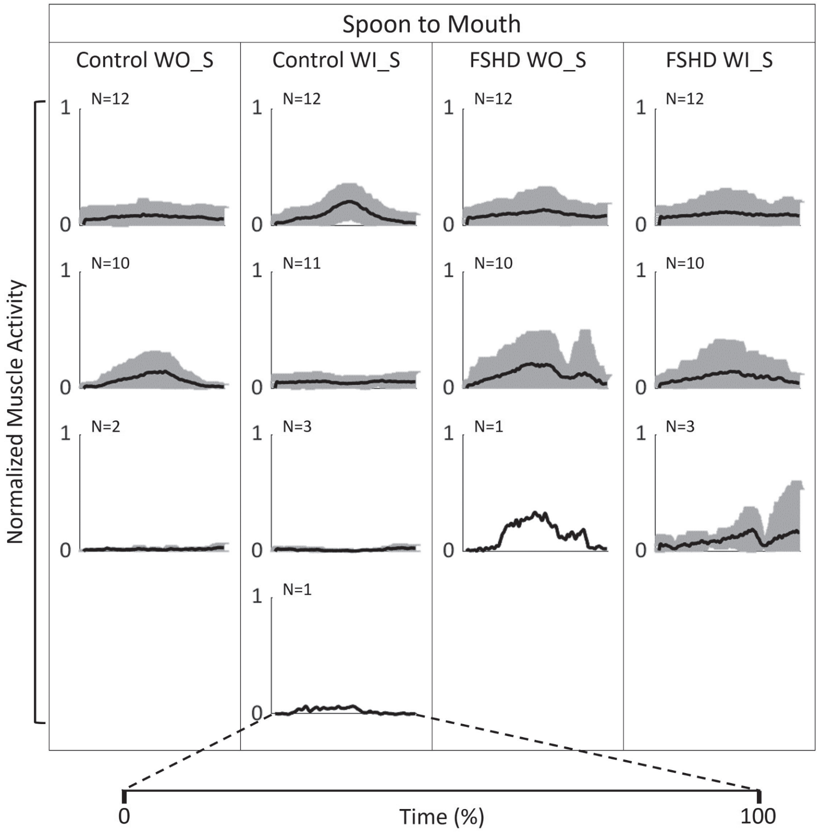
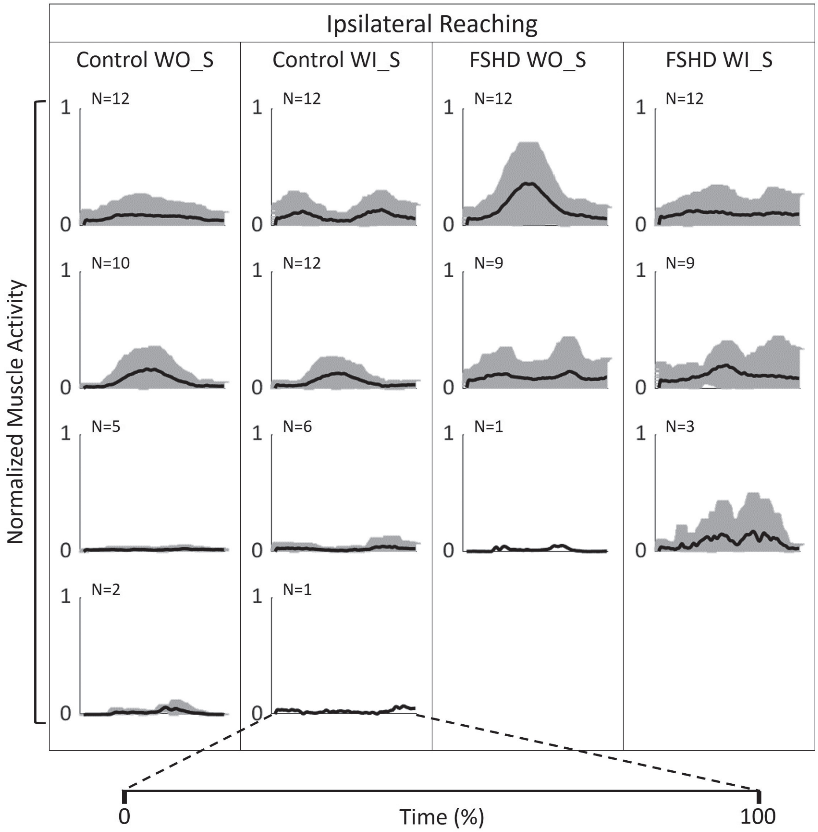
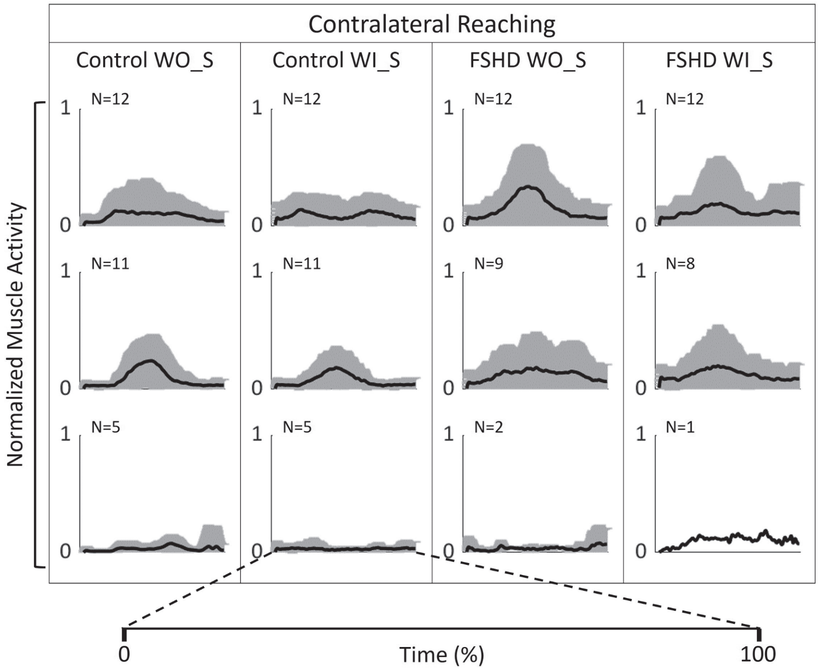


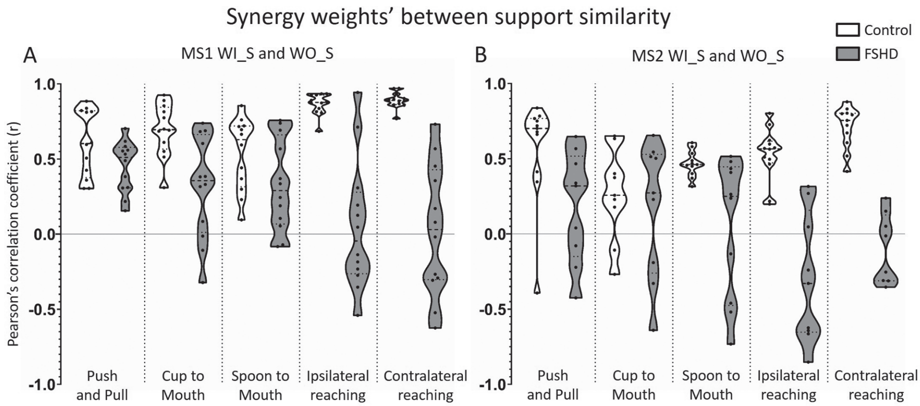
Appendix figure 8. Clustered muscle synergy coefficients during spoon to mouth for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Thick black line represent the mean amplitude and gray area the 95% confidence interval. WO_S: without support, WI_S: with support.



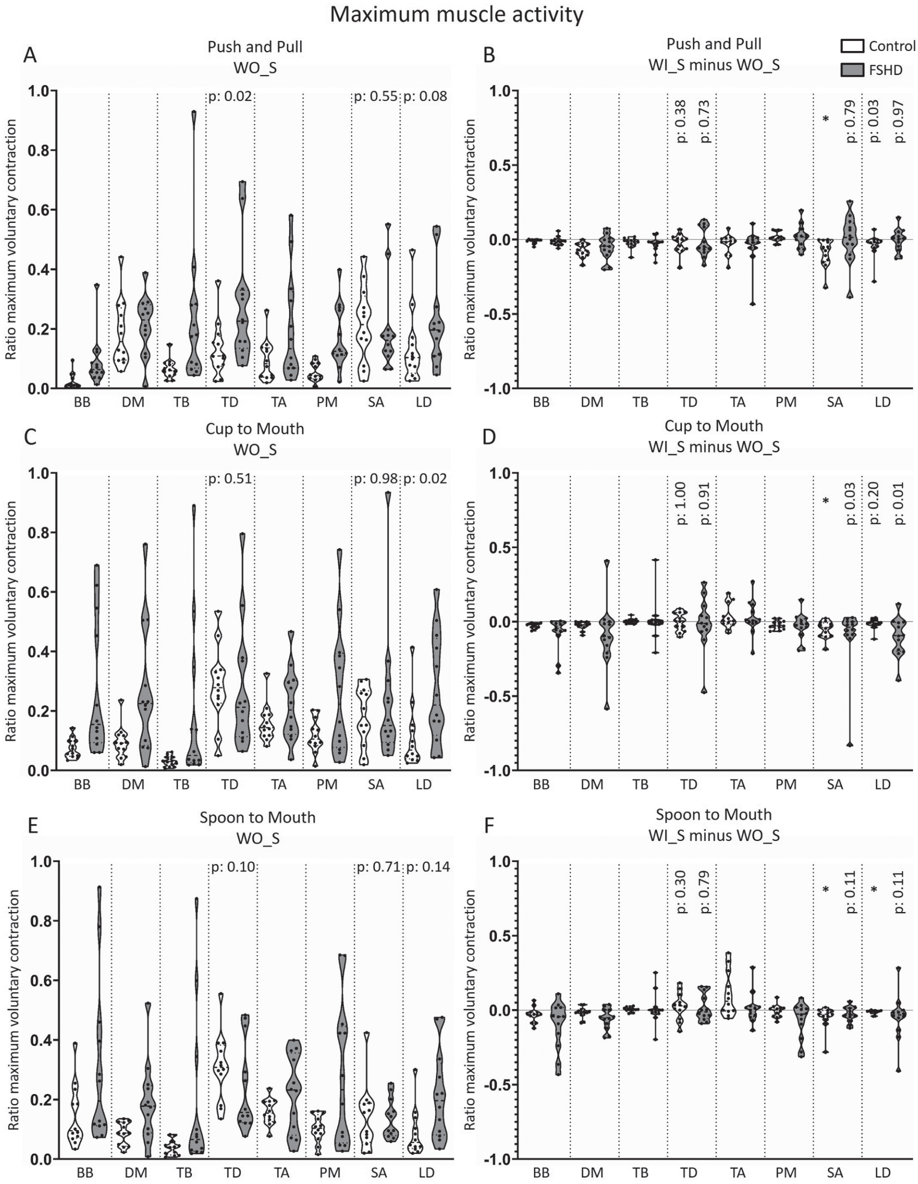
Appendix figure 9. Clustered muscle synergy coefficients during ipsilateral reaching for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Thick black line represent the mean amplitude and gray area the 95% confidence interval. WO_S: without support, WI_S: with support.



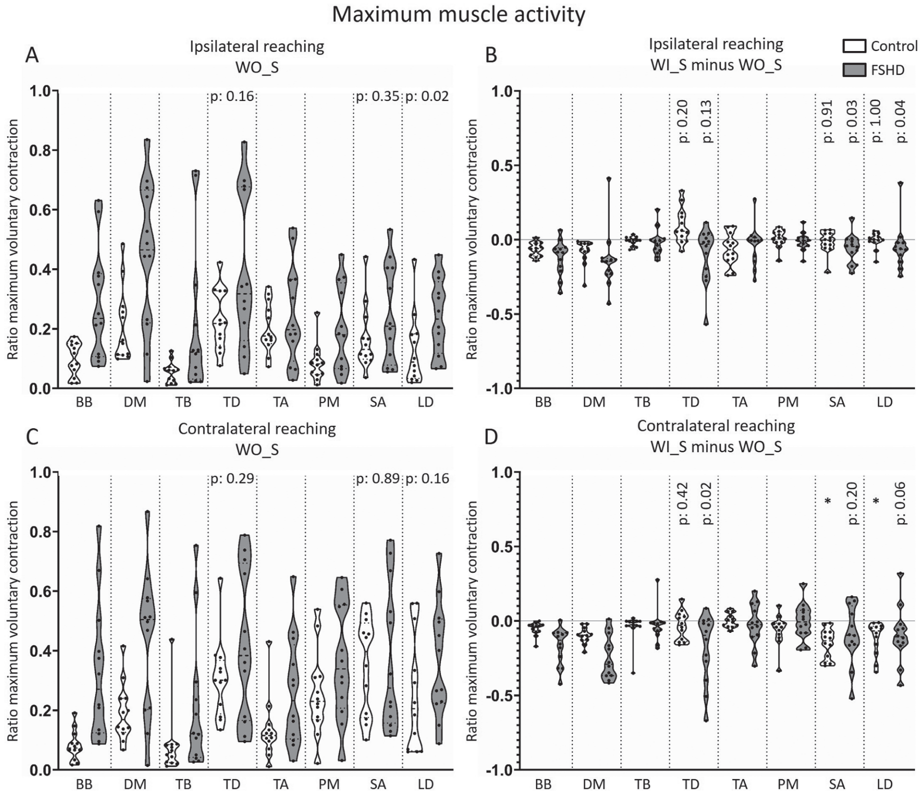
Appendix figure 10. Clustered muscle synergy coefficients during contralateral reaching for control without support (white), control with support (light gray), FSHD without support (gray), and FSHD with support (dark gray) ranked horizontally in order of prevalence (N). Thick black line represent the mean amplitude and gray area the 95% confidence interval. WO_S: without support, WI_S: with support.



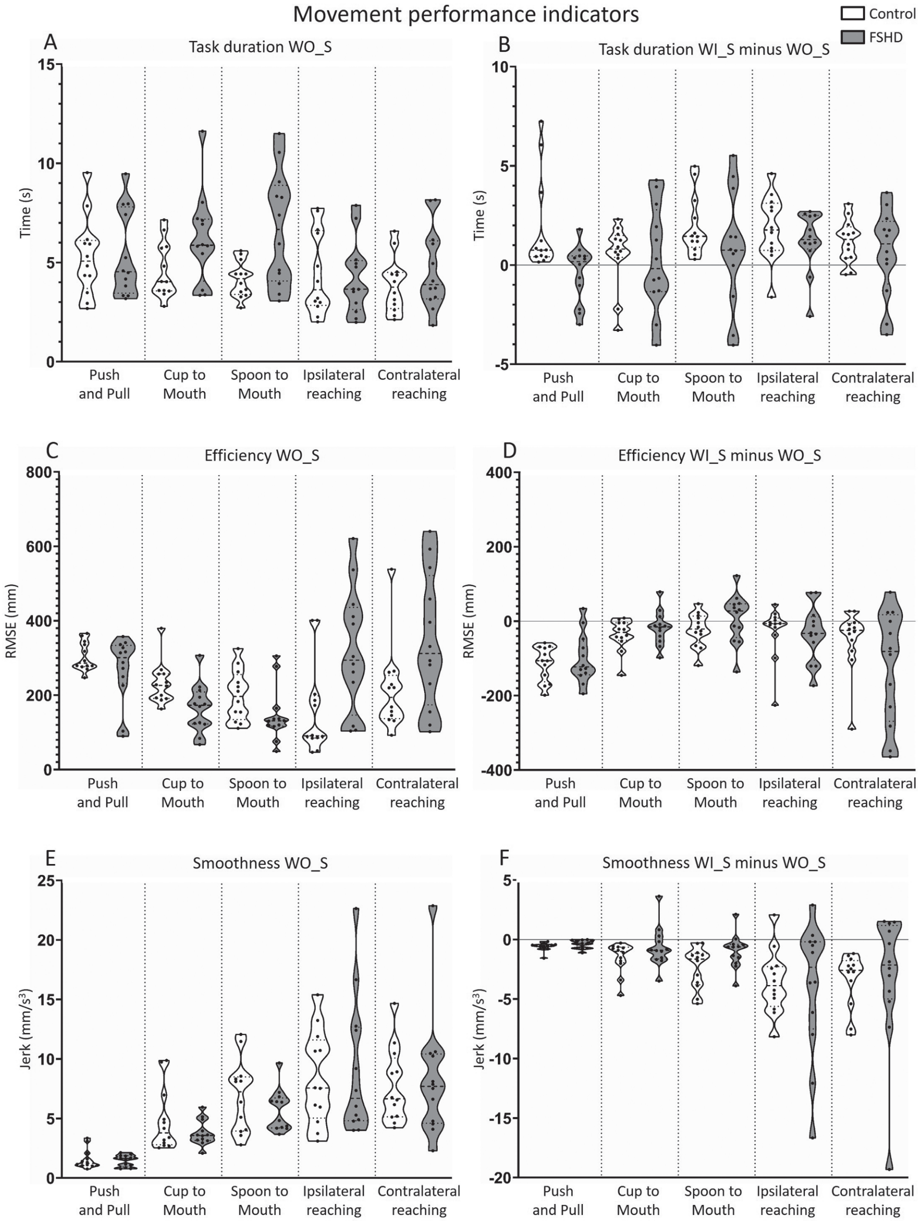
Appendix figure 11. Synergy weights' between support conditions similarity as Pearson's correlation coefficients (r) of MS1 (A) and MS2 (B) presented for controls (white) and FSHDs (gray) as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. WO_S: without support, WI_S: with support.



Appendix figure 12. Maximum muscle activity (ratio of MVC) of unsupported (A, C, E) and supported minus unsupported (B, D, F) tasks. The maximum muscle activities are presented for groups, Control (white) and FSHD (gray), as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. WO_S: without support, WI_S: with support, BB: Biceps Brachii, DM: Deltoid Medial, TB: Triceps Brachii, TD: Trapezius Descendens, TA: Trapezius Ascendens, PM: Pectoralis Major, SA: Serratus Anterior, and LD: Latissimus Dorsi.



Appendix figure 13. Maximum muscle activity (ratio of MVC) of unsupported (A, C) and supported minus unsupported (B, D) tasks. The maximum muscle activities are presented for groups, Control (white) and FSHD (gray), as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. WO_S: without support, WI_S: with support, BB: Biceps Brachii, DM: Deltoid Medial, TB: Triceps Brachii, TD: Trapezius Descendens, TA: Trapezius Ascendens, PM: Pectoralis Major, SA: Serratus Anterior, and LD: Latissimus Dorsi.



Appendix figure 14. Movement performance indicators, task duration (A, B), efficiency (C, D), and smoothness (E, F) of unsupported (A, C, E) and supported minus unsupported (B, D, F) tasks. The movement performance indicators are presented for groups, Control (white) and FSHD (gray), as truncated violin plots. The thick dotted line represents the median, the thin dotted lines the 25th and 75th percentiles, and dots the individuals. WO_S: without support, WI_S: with support.

CHAPTER 5

Daily life benefits and usage characteristics of dynamic arm supports in subjects with neuromuscular disorders

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5.1 INTRODUCTION

Neuromuscular disorders affect 153 persons per 100,000 in the Netherlands and 160 persons per 100,000 worldwide [1, 33]. One of the symptoms of neuromuscular disorders is muscular weakness, which is progressive in most cases and therefore increasingly limits upper extremity mobility and performance during activities of daily life (ADL). Approximately 7–24% of individuals with a neuromuscular disorder use dynamic arm supports (DAS) [8, 36], which provide gravity compensation and can improve mobility and quality of life [5, 7, 46, 115]. Specifically, they facilitate limb motion against gravity [14, 15, 35, 45, 116, 117], reduce efforts [14, 15, 35, 48, 115], and improve ADL performance [5, 12, 45, 116]; thus, supporting the user's overall activity and independence [48, 117]. Studies have shown that the intended benefits of DAS are not completely realized [6, 104], whereas most users seem satisfied with the DAS given to them, with continuous use reported up to 17 h per day [7]. However, over time most users no longer perceive these benefits and stop using the DAS altogether [6], which expectedly leads to a loss of function and reduced participation and quality of life. Experts believe that the disease progression makes it more difficult to operate the DAS; thus, contributing to the changed perception over time [104]. It is therefore important to understand the causes that lead to a DAS not being used any longer by first investigating the quality, or characteristics, of DAS usage.

Experts promote the integration of objective and subjective information on DAS usage [39, 104], to cover different components of the International Classification of Functioning, Disability, and Health (ICF) model [3]. Objective information, such as improved mobility and reduced efforts, mostly reflects device effectiveness in the body function ICF component. Subjective information, such as user needs, wishes, and experiences, mostly cover the impact of a DAS in the activity and participation components of the ICF as well as environmental and personal factors. Ideally, these aspects should be combined and monitored over time to detect temporal changes in daily life behaviour that would result in discontinuation.

Currently, monitoring methods in the field mainly rely on (subjective) self-reports, which are recommended to identify the reasons for DAS use [118]. Such methods are valuable to assess subjective factors, such as the perceived activity, benefits, and limitations, and possibly specific causes for discontinuation of use. However, they are also prone to bias, e.g., self-reported duration of DAS use was found to depend heavily on users' expectations of, and reliance on, the device [7, 116]. Similar bias was also present in studies evaluating the functional improvement where a patient's perceived gain was higher than the gain detectable through clinical scales [57]. The low level of specificity and the bias in self-reports makes it difficult to distil the most important aspects of DAS usage [118]. Therefore, we need to move beyond self-reporting if we want to understand the benefits and limitations of a dynamic arm support for specific user groups.

Accelerometer-based activity monitoring overcomes these inadequacies in self-reporting by objectively quantifying daily life upper extremity activity [28-32]. This approach has been successfully applied in various populations such as children with neurodevelopmental disorders, stroke survivors, and upper limb prosthesis users. In previous work [28], a multi-sensor network was used to classify upper and lower arm activities of children with Duchenne Muscular Dystrophy during ADL. The activity classifications, intensity, orientation, and frequency of arm elevations provided valuable insights into the daily activity levels, such as the timing, intensity, and duration of activities. Furthermore, the activity classifications correlated strongly with upper extremity functionality measured on a clinical scale (Brooke scale r: 0.73 ± 0.13), where less capable participants had lower activity levels and elevated their arms less frequently [28]. However, accelerometer-based activity monitoring has limited power to register and discriminate between postural ADL, such as while holding a telephone to the ear or typing on a keyboard. These activities, which are also important indicators for DAS use, should therefore not be neglected [7] and may be better captured via self-reports. It is evident that accelerometers and self-reports provide more detailed insight in the reasons for discontinued use and guide DAS development. However, this combined approach has still not been used to understand the usage characteristics of dynamic arm supports in persons with neuromuscular disorders.

The aim of the current study is to determine whether DAS produce quantifiable upper and lower arm mobility benefits that impact specific ADL. These benefits are derived from the duration, intensity, and frequency of reoccurring activities performed with and without a DAS. Furthermore, it will be investigated whether users also perceive these benefits, based on self-reporting assessments, and whether these benefits are consistent over time. An integrated activity-monitoring approach that exploits accelerometer sensor networks, in combination with self-reports, was adopted.

5.2 MATERIALS AND METHODS

5.2.1 Participants

Potential participants were informed about this study through digital flyers advertisements within the networks of Dutch Association for Neuromuscular Diseases (Spierziekten Nederland, Baarn, The Netherlands), Focal Meditech (Tilburg, The Netherlands), Maastricht University Medical Center+ (Maastricht, The Netherlands), and University Medical Center Groningen (Groningen, The Netherlands). Interested participants were pre-screened and included when older than 18 years, used a DAS at home, had a diagnosed disease resulting in muscular weakness, did not have other conditions that limited upper extremity movement (i.e., tremors), and could give written informed consent. Three researchers visited the participants at home to acquire the informed consent, provide a diary, and place activity sensors. The central Medical Ethical Committee of Maastricht University Medical Center+ approved the study (17-4-031.1), which was carried out in accordance with the guidelines of the Helsinki protocol.

Nine participants (4M:5F, 51 ± 14 yrs) diagnosed with a neuromuscular disorder were included in this study (Table 1). The participants used one or two DAS devices of the same type. Seven participants were entirely wheelchair-bound with the DAS mounted on the wheelchair. One participant (P3) was ambulant and had a chair-mounted DAS in the kitchen area. Another participant (P5) was ambulant but required a walker and had a wheelchair-mounted DAS. Four participants (P3, 5, 8, and 9) were monitored during three measurement periods, two (P4, 7) during two periods, and three (P1, 2, and 6) during one period of seven consecutive days. The monitored side for P3 and P8 deviated from the dominant side because P3 mostly used his left-sided device and P8 did not have access to her right-sided device due to a scheduled maintenance.

Participants collectively owned four different types of DAS; the Armon Edero [119], Dowing [120], Gowing [11], and Sling [121]. The Armon Edero, Dowing, and Sling are passive support devices with adjustable gravity compensation. The Armon Edero and Dowing use adjustable springs, whereas the Sling uses counterweights to support the weight of the arm. The Gowing is a hybrid device that provides spring-actuated passive support with an addition of motorized actuators to adjust the springs and provide active support. These devices are relatively easy to put on and off, allowing the user to switch between use and non-use with little effort.

Table 1. Participant characteristics.

	Gender (M/F)	Age (Y)	Diagnosis	Dynamic arm support	Dominant side	Monitored side	Total days
P1	F	61	Congenital Myopathy	Gowing	Equal	Left	3
P2	M	44	Amyotrophic Lateral Sclerosis	Gowing	Right	Right	6
P3	M	58	Progressive Spinal Muscular Atrophy	Dowing (bilateral)	Right	Left	21
P4	F	30	Limb Girdle Muscular Dystrophy	Armon Edero	Right	Right	12
P5	F	63	Desminopathy	Sling (bilateral)	Right	Right	16
P6	F	60	Limb Girdle Muscular Dystrophy	Dowing	Right	Right	4
P7	M	55	Limb Girdle Muscular Dystrophy	Gowing	Right	Right	13
P8	F	60	Spinal Muscular Atrophy type 3	Gowing (bilateral)	Right	Left	17
P9	M	27	Duchenne Muscular Dystrophy	Gowing (bilateral)	Right	Right	21

5.2.2 Activity Sensors

DAS use and upper extremity motions were monitored using activity sensors, Figure 1. The activity sensors (MOX, Maastricht Instruments, Maastricht, the Netherlands) were 3D accelerometers with inbuilt data loggers capable of at least seven days of recording with a sample rate of 25 Hz [26, 27]. Sensors were placed similarly to previous work [28]; on the lateral side of the upper arm (UA), on the lower arm at thewrist (LA), and on the device's base (DB), and in addition, on the supporting brace of the DAS in line with the wrist sensor (SB). The UA sensor provided information on the utilization of the shoulder joint, the LA sensor on the utilization of the shoulder and elbow joints combined, the LA and SB sensors combination on the use of the DAS, and the DB sensor on the transportation of the whole device. Participants were asked to wear the sensors continuously up to a maximum of 21 monitoring days divided into three periods of seven consecutive days with an interval of at least fourteen days between periods to monitor changes over time. Data analysis focused on a 24-h representation of activity. Therefore, days when sensors were worn for less than 24 h or when sensors had technical issues were excluded from the data analysis.

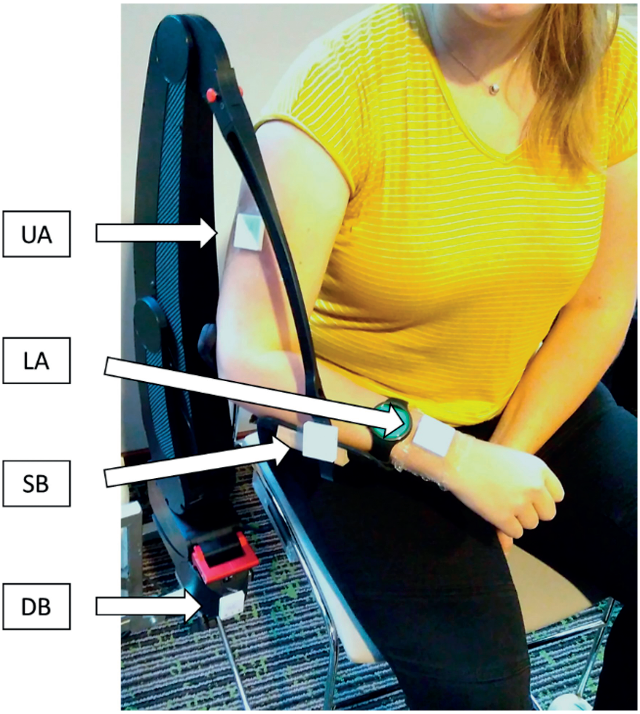


Figure 1. Measurement setup demonstration on a healthy volunteer. UA: Upper Arm, LA: Lower Arm, SB: Support Brace, and DB: Device Base.

5.2.3 Data Processing

Recorded accelerations were processed to provide information on the daily intensity (counts/s), defined as an integrated vector [122] and body segment orientation based on the gravity vector (pitch: 0–180 degrees) over time [28] (Figure 2, appendix figure 1). This information was further transformed to express the DAS usage characteristics in terms of duration, frequency, and activity levels (Table 2).

First, the tri-axial accelerations were filtered with a fourth order Butterworth 0.025 to 7.5 Hz band pass filter to calculate the intensity for each sensor (appendix figures 2–5). A threshold (1.125 counts/s) was determined during system calibrations to distinguish between still and motion based on the collective sensor noise level of a one-minute recording while sensors laid still and a one-minute recording of slow movements. Furthermore, we have verified the threshold with participant data and found the threshold to clearly distinguish between resting periods and activity bouts, also over longer periods of time. All intensity data were categorized per second as still or motion and corrected for non-stationary periods of the device's base as still. The lower arm sensor determined the periods of activity (motion). Within active periods, the support brace sensor determined the periods of DAS non-use (still) and use (motion). The periods of activity, non-use, and use were expressed as cumulative minutes per day. Furthermore, the intensity levels of non-use and use were extracted for the upper and lower arm and period occurrences of non-use and use for the lower arm only. The intensity levels were expressed as cumulative counts per day and the period occurrences (episodes) counted per day. The intensity data processing resulted in the outcome parameters periods of activity, non-use, and use, and the intensity levels and episodes of non-use and use of the upper and lower arm (Figure 2).

Second, accelerations of the axis in line with gravity, in a neutral body position, were filtered with an eight order Butterworth 2 Hz low pass filter to calculate the orientation from the arccosine per body-worn sensor (appendix figures 2–3). Orientation data were categorized for each sample (1/25 s) as low or high with a double threshold to filter threshold fluctuations. Samples had to be below the low threshold (UA: 40° and LA: 115°) or above the high threshold (UA: 50° and LA: 125°) to be classified as such and those

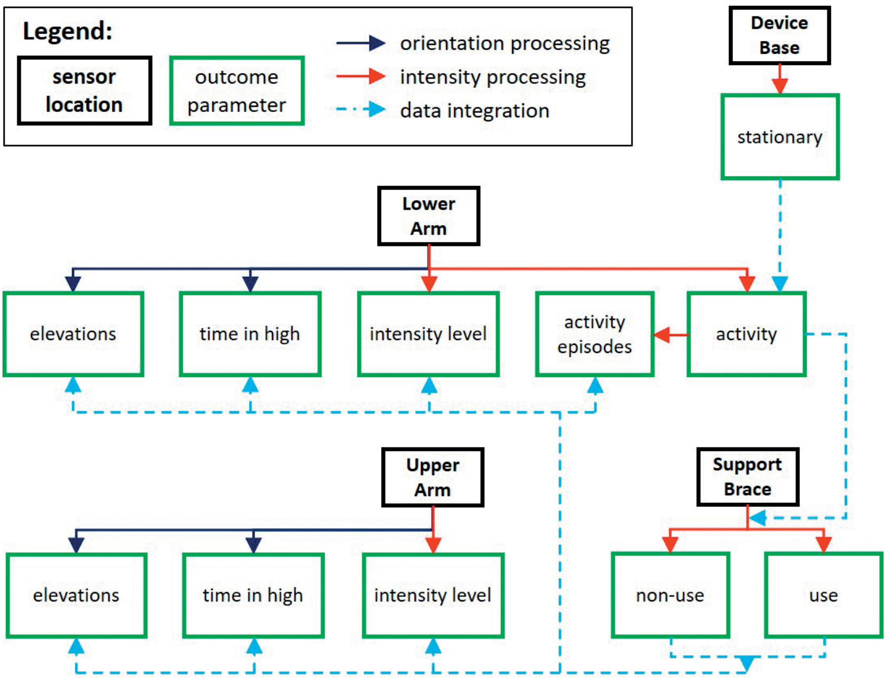


Figure 2. Multi-sensor network processing for orientation and intensity related parameters and integration of data between sensors.

between thresholds were placed in the same category as the precedent sample. The upper arm thresholds were chosen to differentiate between the low (<45°) and the middle and high orientations (>45°) of the upper arm, as used by van der Geest et al. 2019 [28], with a tolerance margin of ±5°. The lower arm thresholds represent the inclination, 120° with a tolerance margin of ±5°, where motions become more challenging with a DAS. This is because the application of the vertical force, normally on the entire lower arm, is being positioned towards the elbow with greater inclination and users therefore receive less support for elbow flexion. These thresholds were then verified during system calibrations for several ADL; simulated eating/drinking, reaching above shoulder level, and typing on a keyboard, at a slow, normal, and fast pace without and with a DAS. Time spent in high orientations were added for the upper and lower arm respectively and expressed as minutes per day. Furthermore, transfers from a low to high state were counted to represent the respective arm elevation frequency expressed as occurrences per day. The orientation data processing resulted in the outcome parameters time in a high orientation and arm elevation frequency, both during non-use and use of the upper and lower arm (Figure 2).

Finally, the parameters related to intensity levels, episodes, time in high orientation, and arm elevation were also normalized for DAS use and expressed as percentages of non-use and use, respectively (Table 2).

Table 2. Overview of monitored outcome parameters that represent dynamic arm supports (DAS) usage characteristics.

Primary	Secondary		
Time absolute (min/day)	**Time** absolute (min/day) normalized (% activity)	**Activity levels** absolute (counts/day) normalized (counts/min)	**Frequency** absolute (#/day) normalized (#/min)
activity use non-use	UA use in high LA use in high UA non-use in high LA non-use in high	UA use intensity LA use intensity UA non-use intensity LA non-use intensity	UA use elevations LA use elevations UA non-use elevations LA non-use elevations LA use episodes LA non-use episodes

5.2.4 Self-Reports

Diaries were used to extract self-reported ADL that are considered reoccurring within a day or week, such as eating/drinking, self-care, and work. Furthermore, the participant was asked retrospectively to answer five questions per monitoring period concerning DAS benefits and limitations. The questions were “describe motions or activities that (1) were unsuccessful without DAS, (2) were unsuccessful with

DAS, 3) required increased effort with DAS, (4) were only possible without DAS, and (5) were only possible with DAS". In addition, we retrospectively inquired about the participants' (1) perceived daily activity, (2) perceived daily use, (3) perceived benefit from the DAS on a 0 to 100 scale (0 = "I don't use it at all" and 100 = "I use it continuously"), and (4) the ratio between left and right arm involvement in their activities on a 0 to 100 scale. Answers were collected after initial data quality analysis, two to four weeks after all monitoring periods, to minimize the influence on participants' awareness of DAS use and ADL performance.

5.2.5 Data Synthesis

Monitored daily activity levels were divided in primary and secondary outcome parameters (without and with DAS) based on the data processing sequence. Primary outcome parameters were quantified as the time spend active and secondary as (1) the time spend with the arm elevated, (2) the frequency of activities episodes and (3) arm elevation, and (4) the intensity levels of the activities. Device benefits were quantified as the effect sizes of secondary outcome parameters during DAS use compared with non-use. The perceived averaged daily use, collected once, was tested for significant difference with the monitored daily use (primary outcome parameter). The daily collected perceived and monitored device benefits were compared on the similarity of self-reported activity benefits and the effect sizes of daily activity levels (secondary outcome parameters).

5.2.6 Statistical Analysis

The perceived daily DAS use was compared to the averaged monitored equivalent using a paired sample Wilcoxon signed rank test (SPSS) [123]. Changes in monitored activity and DAS use over the monitoring periods were investigated with a non-paired sample Mann–Whitney U test of period combinations 1–2, 2–3, and 1–3 within each subject where possible. Alpha levels were set at 0.025.

Cohen's d effect sizes were calculated within subjects for the secondary outcome parameters (absolute and normalized) using the formula:

$$\mathrm{d} = \frac{(\mathrm{m1} - \mathrm{m2})}{\sqrt{\mathrm{s1}^2 + \mathrm{s2}^2 - (2 * \mathrm{r} * \mathrm{s1} * \mathrm{s2})}} \tag{1}$$

Where m1 and m2 represent the means during use and non-use respectively, and the s1 and s2 the standard deviations during use and non-use, respectively. Pearson's correlation coefficient (r) was calculated between use and non-use over the processed days. The effect sizes were calculated for each participant and as a group for the secondary outcome parameters, time spend in high, intensity levels, and elevations for the upper and lower arm sensors, and for episodes of the lower arm sensor. Effect sizes were ranged as small: 0.20–0.50, medium: 0.50–0.80, and large: >0.80 [83]. Parameters with a group effect size of medium and above (>0.50) were considered a mobility benefit resulting from DAS use. Changes in effect sizes over the periods were investigated with a paired sample Wilcoxon signed-rank test of period combinations 1–2, 2–3, and 1–3, where possible. Alpha levels were set at 0.025.

5.3 RESULTS

The benefit of DAS use was quantified by comparing use and non-use periods of secondary daily activity levels. Effect size of DAS use was medium to large for normalized elevations of the upper (Cohen's d: 0.6, n = 3) and lower arm (Cohen's d: 1.0, n = 4), and large for normalized episodes of the lower arm (Cohen's d: 1.7, n = 8) (appendix table 1). Other normalized parameters did not show a medium or above group effect and absolute values showed negative effects. The effect sizes over the three periods were not significantly different (appendix figures 6–9). Daily activity levels were reported in the appendix (appendix figures 10–13).

Table 3. Summary of the self-reported reoccurring, facilitated, and limiting activities.

Reoccurring activities	Only possible with device	Only possible without device
Eating/drinking (P1-5,7-9) Personal hygiene (P1-5,7-9) Computer activities (P1-5,7-9) Household chores and cooking (P1,4,5,7,8) Touch head (P7,8,9)	Eating/drinking (P4,5,8,9) Extended time computer work (P5,8) Touching head (P7,8) Personal hygiene (P8)	Driving a car (P4) Motions beyond the ambulant chair (P5) Typing (P7)
	Unsuccessful with device	**Unsuccessful without device**
	Proper pronation/supination (P3) Maintaining arm within DAS (P3,4) Reaching below waist level (P3-5) Personal hygiene (P4) Folding laundry (P5) Support of wrist during hand to mouth (P5) Opening door handle (P7) Washing hand (P9)	Personal hygiene (P2) Reaching above shoulder (P3,4,5,7) Household chores/cooking (P4) Shake hands (P5) Flush toilet (P5) Opening door handle (P7) Eating/drinking (P7) Almost everything (P9)

Reoccurring activities were mostly related to self-care (eating/drinking, hygiene, and household chores) and computer activities (Table 3). The consensus from self-reports was that the DAS facilitated these activities and those involving reaching above shoulder level. Activities that involved forearm rotation, wrist motion, or large motions were mostly limited with a DAS. The averaged DAS benefits were rated as 83 ± 15% (appendix table 2). Furthermore, participants indicated that their monitored limb was used 60 ± 10% for their daily activities.

Participants perceived their averaged daily DAS use as more than monitored (p: 0.015) (Figure 3). Accelerometer data indicated that participants were on average active for roughly 561 ± 149 min a day of which 94 ± 77 min, or 18 ± 15%, was with a DAS (Figure 3). In contrast, participants reported an average of 430 ± 280 min being active a day of which 283 ± 212 min, or 74 ± 31%, with a DAS (appendix table 2). Furthermore, participants showed consistent activity and DAS use over time, except for P8 where the first period was lower ($p < 0.010$) compared to the latter two (Figure 4). The participant had stated in the diary to be ill and not very active in that period.

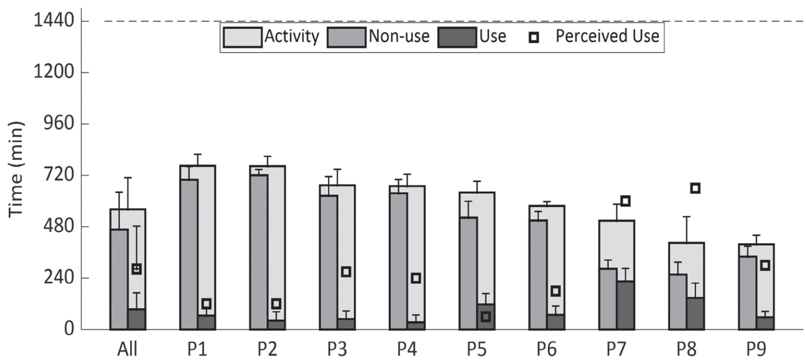


Figure 3. Averaged monitored daily activity, non-use, and use expressed in minutes as bars and perceived use as squares, both with one standard deviation. The dotted line represents a complete day. Participants were sorted on activity time in a descending order for visualization purposes.

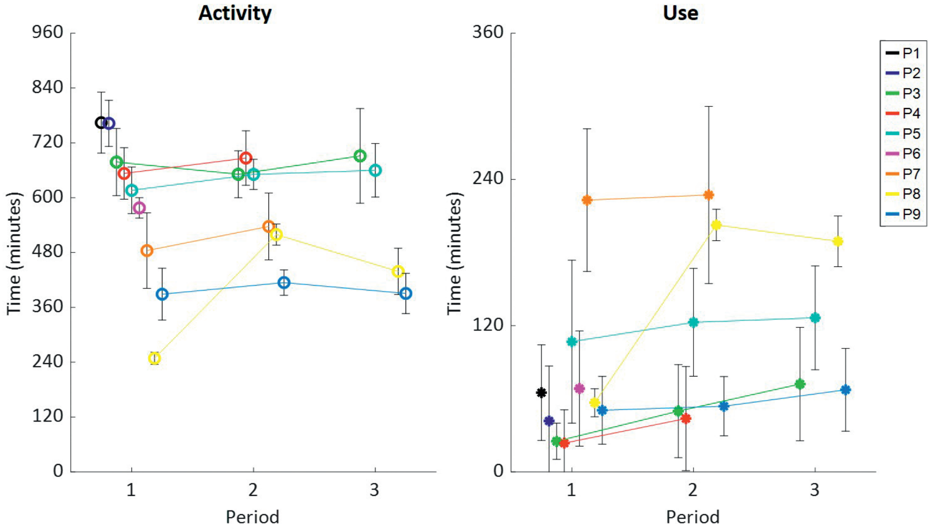


Figure 4. Activity (left; o) and use (right; *) per individual (colors) over the three periods. Participants were sorted on averaged activity time of all periods in a descending order for visualization purposes.

5.4 DISCUSSION

This is, to the best of our knowledge, the first study that included a multi-sensor network of accelerometers in combination with self-reported activity to monitor the usage characteristics of dynamic arm supports (four different types) in people with neuromuscular disorders in a home environment. The primary results were that 1) the DAS facilitated motions against gravity and enhanced the occurrence of these activities and 2) that in our study population the objectively measured use of the DAS did not change over a two-month period. Furthermore, it was found that self-reports seriously (3-fold) overestimated the time spent using the arm support. The self-reports did yield more detailed information on the

circumstances when the arm support was beneficial such as prolonged computer work, household chores, and personal hygiene. In addition, it gave detailed information on the conditions in which the arm support was considered to limit activities, such as in activities involving wrist movements, forearm rotation, and large motions of the arm. This study clearly shows the benefits of a combined approach in quantifying the benefits and limitations of dynamic arm supports for activities in daily life, although several aspects need to be considered to optimize the approach.

The integration of accelerometers with self-report measures in a home environment provided a quantifiable comparison of mobility benefits to describe facilitated ADL. As expected, repetitive motions, especially against gravity, were facilitated by DAS and perceived as beneficial in reoccurring ADL, eating/drinking and touching their head. However, several perceived ADL benefits or limitations were not identifiable from accelerometer data because the context was too general (household chores and personal hygiene) or they concerned aspects which were not measured (wrist movements and device range of motion). Furthermore, the heterogeneous effect sizes across participants and parameters favours the focus on participant-specific rather than general mobility benefits. Therefore, device effectiveness would be best determined on an individual basis. This should be organized according with what a person can do (motor capability), what a person wants to do (needs and wishes), and what a person does in daily life (motor performance) [104]. However, common benefits, such as activity frequency and arm elevation, could provide initial indications of device effectiveness for essential activities, such as eating/drinking [48, 124].

The ability to monitor changes in the use of the dynamic arm support over time could potentially help clinicians and developers optimize the system for the user [39, 104]. The two-month monitoring period used in this study was too short to capture discontinuation or even reveal large changes in usage, although one participant showed a large increase in activity from week 1 to weeks 2 and 3, which via the self-reports could be traced to a recovery from illness. Gradual changes in use could therefore provide first indications on disease progression, depending on the expected progression rate. The monitoring period should therefore be aligned with the expected progression rate, which for some neuromuscular disorders could expand to several months or years [125]. Essentially, a DAS should facilitate the use of muscles and independence in activities of daily life as important aspects of health, physically and mentally. Therefore, a better device match is preferred, which depends on the user's changing capabilities, needs, and wishes. DAS developers could use these longitudinal assessments to optimize device benefits and clinicians would be able to alter therapy or DAS type in individual cases so as to limit future discontinuation [104].

However, future research should reduce the physical and mental burden by minimizing the required number of worn sensors and diary input [28]. It was found that for some participants wearing two sensors and keeping a diary for seven consecutive days was a heavy burden, both physically as mentally. Solutions that use a single sensor at the wrist and use experience sampling of the satisfaction via an app, might overcome these issues an offer a feasible and participant-friendly option to obtain this valuable information.

Studies on whole body activity monitoring show that participants tend to overestimate their activity levels and underestimate their sedentary/resting episodes [126-128]. As expected, participants in the current study also mostly overestimated their use compared to those monitored with the multi-sensor network. However, the self-reported daily activity levels, mobility benefits, and satisfaction were comparable to those in other self-reported studies [6, 7, 115]. The monitored data showed that participants in this study spent about as much time being physically active as unimpaired people [29]. Furthermore, they lifted their arms about as often (28.9 ± 15.4 per hour) as children with Duchenne [28]. It should be realized that objective and subjective outcomes relate to different ICF domains. The multi-sensor network measures mostly the body functionality, while self-reports naturally tend to cover activity and participation [6, 48, 104]. Even though the ICF model components interact with each other, the measurements methods may not reflect the same aspects of DAS usage characteristics. To clarify, DAS users may have utilized the device for postural support or might have had breaks during activities that were experienced as a continuous activity but not monitored as such by the multi-sensor network. For example, arm motion is not required when typing and reading during computer work or chewing during eating, but these activities can still be considered as using the DAS. This could explain the differences in DAS use to some extent although not the observed daily difference of 3 hours. Furthermore, the averaged monitored use is considerably low (<20%) and varies greatly between and within participants. It is therefore unclear whether self-reports are sufficient to reflect the daily use or if postural activities present such a large aspect of DAS use. However, this would have a limited effect for within-participant comparisons, which are more important for future directions, such as determining individual device effectiveness and performing longitudinal measurements [6, 39, 104].

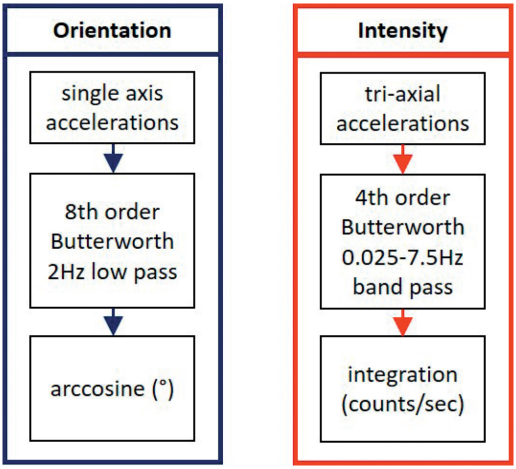
As expected, larger effect sizes were generally more prominent in the lower arm than the upper arm, as the lower arm is primarily supported and motion in the upper arm influences the first. Furthermore, effect sizes varied greatly between participants which could reflect environmental and personal factors, such as device-specific facilitations or otherwise undoable ADL and disease-specific affected muscle regions. In muscular dystrophy, such as Duchenne and limb girdle, proximal regions of the upper extremity are affected first which limits upper arm elevations [2, 36]. In addition, ADL such as eating and drinking can be performed with various contributions from the shoulder and elbow joint and may not depend on the ability to utilize both. Therefore, investigating the upper extremity as separate segments might reveal benefits corresponding to specific environmental and personal factors. However, the upper arm sensor would be redundant when investigating the collective mobility benefits reported in this study.

This study provided several significant contributions to a better understanding of DAS usage characteristics. First, the complementary information from objective and subjective measures provided better insights in DAS use and benefits, specifically about overestimation in self-reporting and sensitivity to energy loss. Second, mobility benefits were up to now not yet quantified in a home environment. Consequently, this study highlighted common benefits and the need for individual assessment. Third, multiple assessments up to a two-month period provided first-time evidence of actual use and revealed longitudinal consistency of daily activities. However, there were some limitations present in this study. First, the majority of participants did not complete the proposed 21 days, due to the imposed physical and mental burden (n = 5) or technical issues (n = 3), due to a limited battery life and noise (data clipping and repetitive single sample acceleration peaks). Sensor and diary reductions might lower the burden and thus allow a larger inclusion of participants and recordings. Second, the current study reports the mobility benefits after the DAS has been integrated in daily life and might not fully reflect the intended life-changing benefits. Furthermore, participant-specific benefits were not investigated, as residual arm movement and disease progression were not the target of this study. However, the current method could be used to monitor the DAS usage characteristics pre and post DAS integration in daily life to investigate the realization of intended benefits for several purposes such as device development and user-device optimization [104].

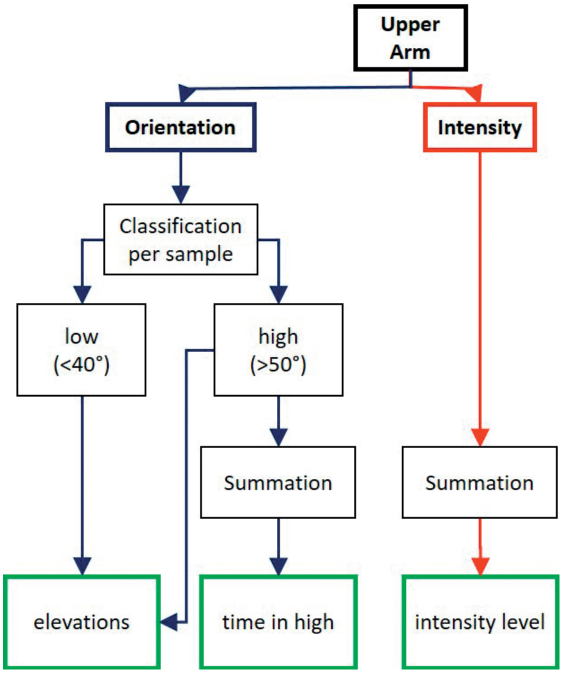
5.5 CONCLUSIONS

This study showed that the integration of a multi-sensor network and self-reports gives additive information on the use and benefits of dynamic arm supports. It has been shown that movements performed at home were executed more frequently with an arm support and that participants benefitted from the devices in important tasks such as eating and drinking and touching their head. Further simplification and integration of the assessments to address the relevant ICF domains is necessary to translate users' needs and wishes to mobility benefits and determine device effectiveness.

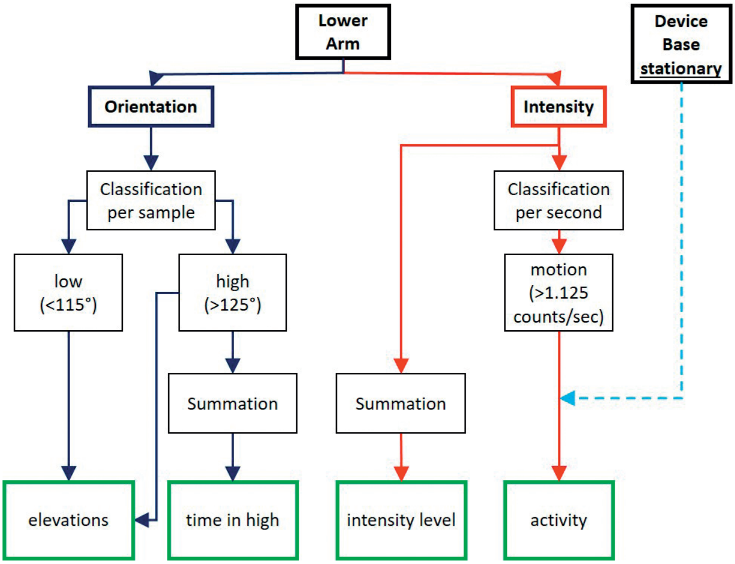
5.6 APPENDIX



Appendix figure 1. Overview of orientation and intensity data processing.



Appendix figure 2. Overview of upper arm sensor data processing.



Appendix figure 3. Overview of lower arm sensor data processing.

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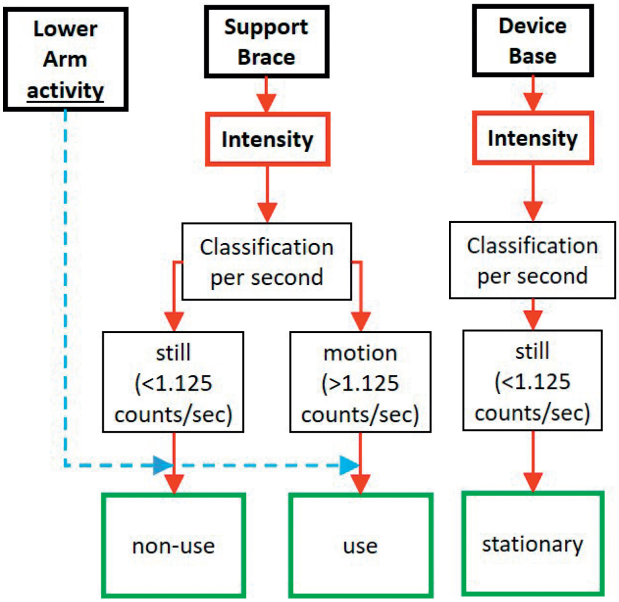


Appendix figure 4. Overview of support brace sensor data processing.

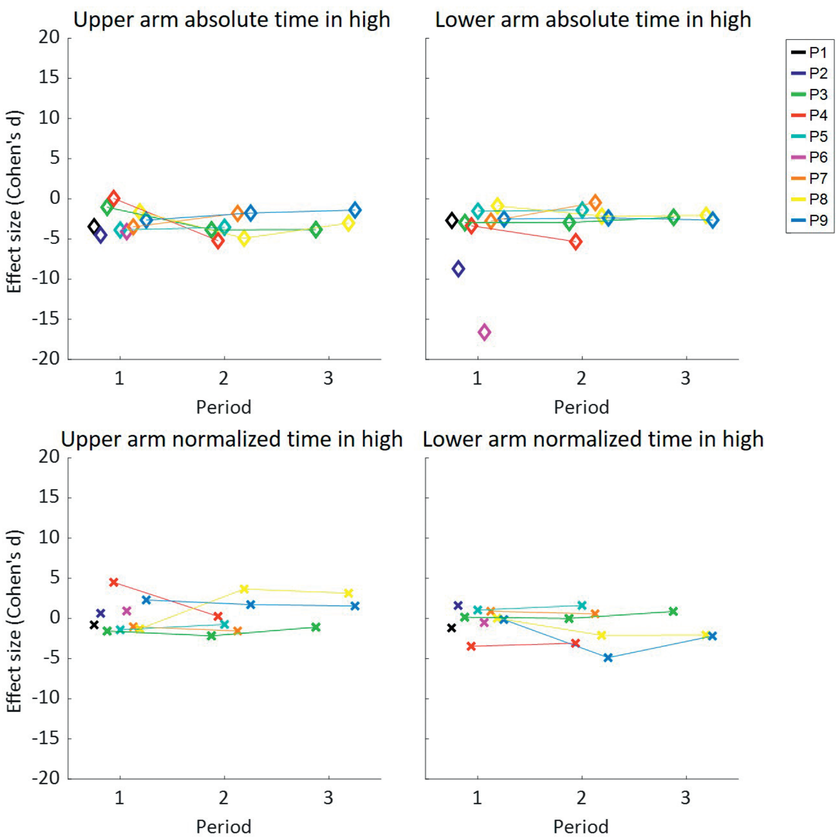
Appendix figure 5. Overview of device base sensor data processing.

Appendix table 1. Effect sizes (Cohen's d) for absolute (abs) and normalized (norm) parameters. UA: Upper Arm; LA: Lower Arm.

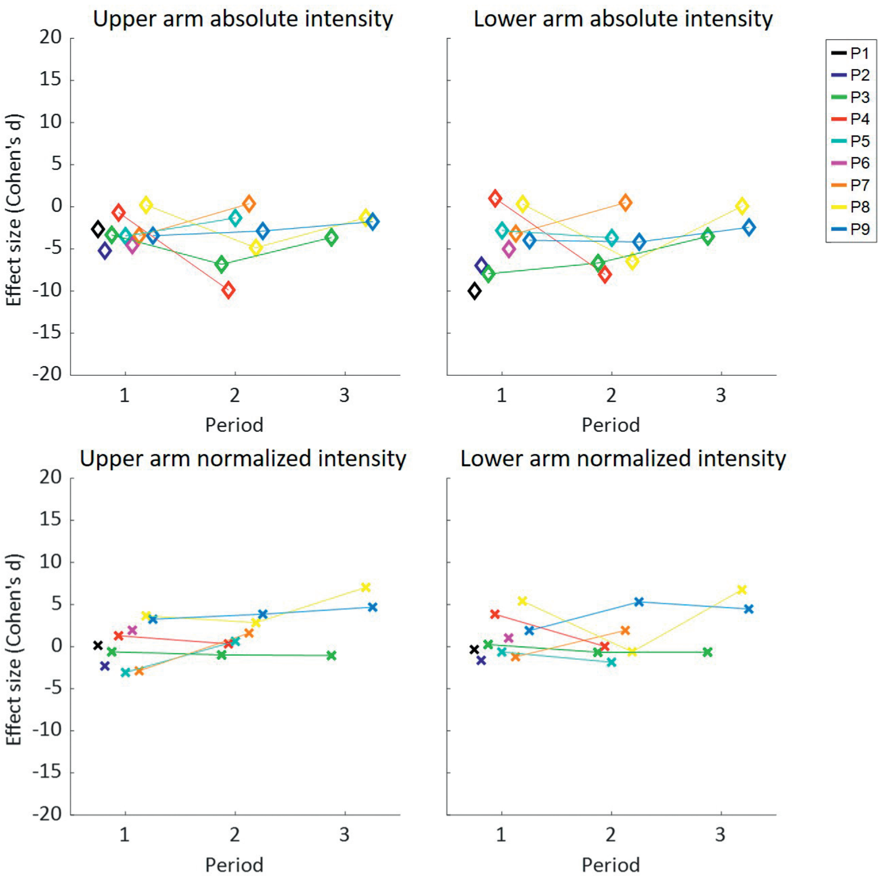
	Time in High UA		**Time in High LA**		**Intensity UA**		**Intensity LA**		**Elevations UA**		**Elevations LA**		**Episodes LA**	
	abs	norm	abs	norm	abs	norm	abs	norm	abs	norm	abs	norm	abs	norm
All	-1.9	-0.1	-1.6	0.0	-1.2	0.2	-1.3	0.1	-1.1	0.6	-0.9	1.0	-1.9	1.7
P1	-10.5	-16.3	-5.9	0.2	-6.5	1.0	-7.8	-0.1	-4.0	-0.1	-4.8	4.0	-5.7	4.1
P2	-3.5	-0.8	-2.7	-1.2	-2.7	0.1	-10.0	-0.3	-3.6	0.4	-2.9	0.2	-7.3	2.5
P3	-1.6	-1.6	-2.8	0.3	-4.2	-0.9	-5.0	-0.3	-3.0	0.2	-2.5	0.3	-3.8	0.8
P4	-4.7	0.5	-7.5	-1.0	-6.0	1.0	-6.3	0.6	-3.3	0.8	-6.0	0.1	-4.8	0.5
P5	-3.4	-1.2	-1.9	1.0	-2.4	-0.2	-3.2	-1.0	-2.2	-0.2	-1.2	1.5	-1.9	3.6
P6	-4.5	0.6	-8.7	1.6	-5.2	-2.3	-7.0	-1.6	-3.5	-0.5	-3.4	2.1	-2.8	3.7
P7	-1.6	-1.5	-0.7	0.2	0.3	2.1	0.4	2.6	1.2	2.3	1.8	4.3	-1.5	-0.5
P8	-1.3	2.3	-1.8	-2.2	-1.2	2.1	-0.6	0.4	-2.1	-1.3	-2.3	-1.5	-0.6	1.6
P9	-1.8	1.8	-1.6	-0.8	-2.5	3.8	-3.3	3.1	1.5	2.0	-2.6	0.2	-2.7	1.9

Appendix table 2. Participants' self-reported activity, DAS use, benefit, and limb side involvement.

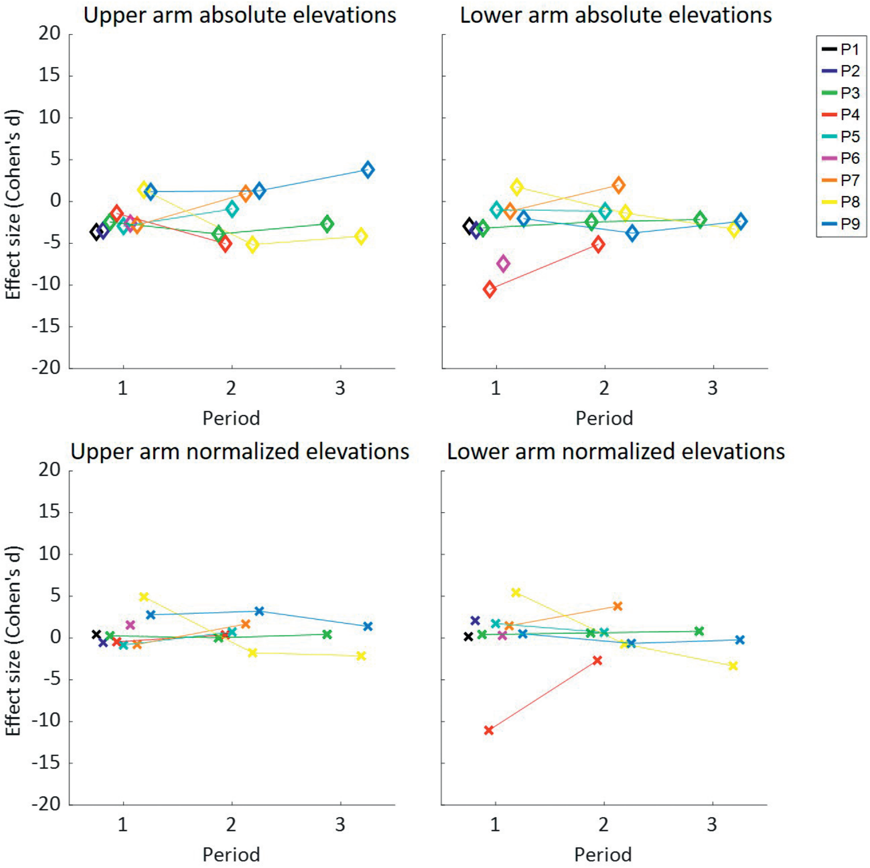
	Perceived activity (min)	Perceived use (min)	Perceived DAS use (%)	Perceived DAS benefit scale (%)	Ratio left-right arm activity
P1	120	120	100	80	50/50
P2	300	120	40	80	33/67
P3	270	270	100	90	70/30
P4	720	240	33	60	33/67
P5	60	60	100	67	40/60
P6	300	180	60	70	30/70
P7	600	600	100	100	40/60
P8	660	660	100	100	40/60
P9	840	300	36	100	40/60



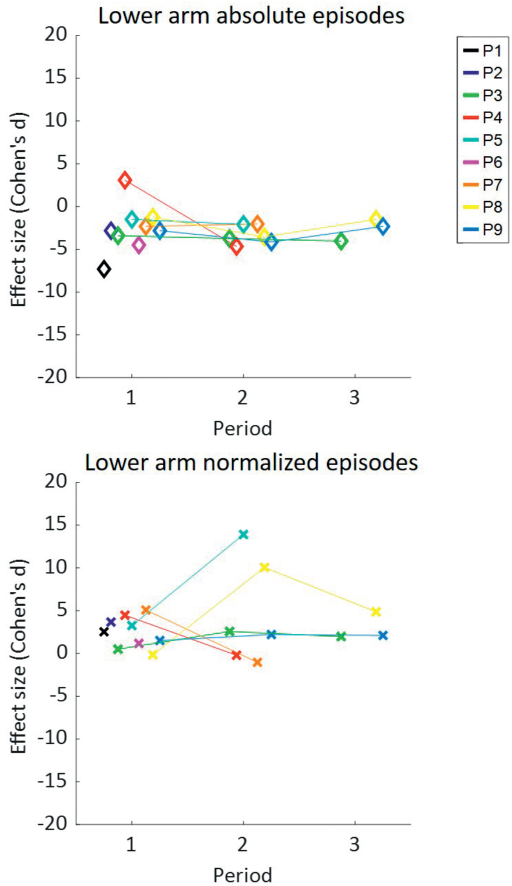
Appendix figure 6. Effect sizes of upper (left) and lower arm (right) time in high absolute (top; diamond) and normalized (bottom; x) per participant (colors) over the three periods. Participants were sorted on averaged activity time of all periods in a descending order for visualization purposes.



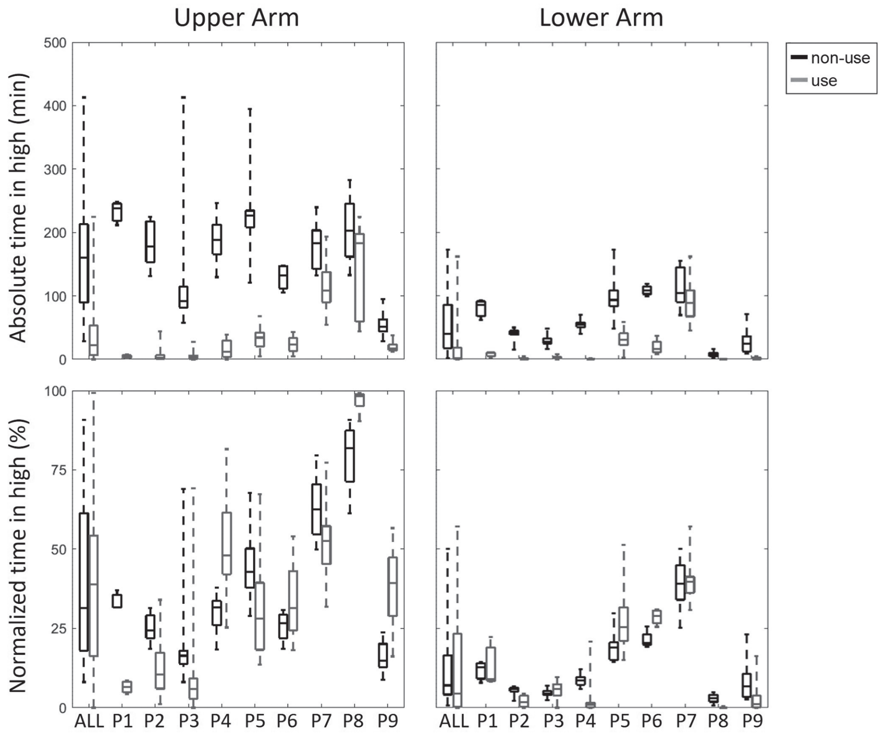
Appendix figure 7. Effect sizes of upper (left) and lower arm (right) intensity absolute (top; diamond) and normalized (bottom; x) per participant (colors) over the three periods. Participants were sorted on averaged activity time of all periods in a descending order for visualization purposes.



Appendix figure 8. Effect sizes of upper (left) and lower arm (right) elevations absolute (top; diamond) and normalized (bottom; x) per participant (colors) over the three periods. Participants were sorted on averaged activity time of all periods in a descending order for visualization purposes.



Appendix figure 9. Effect sizes of lower arm episodes absolute (top; diamond) and normalized (bottom; x) per participant (colors) over the three periods. Participants were sorted on averaged activity time of all periods in a descending order for visualization purposes.



Appendix figure 10. Daily time in high for the upper (left) and lower arm (right) expressed in absolute (top) and normalized (bottom) values during DAS non-use (black) and use (grey). Boxplots represent 25th and 75th percentiles, the median is the box bar, and the whiskers represent the min and max.

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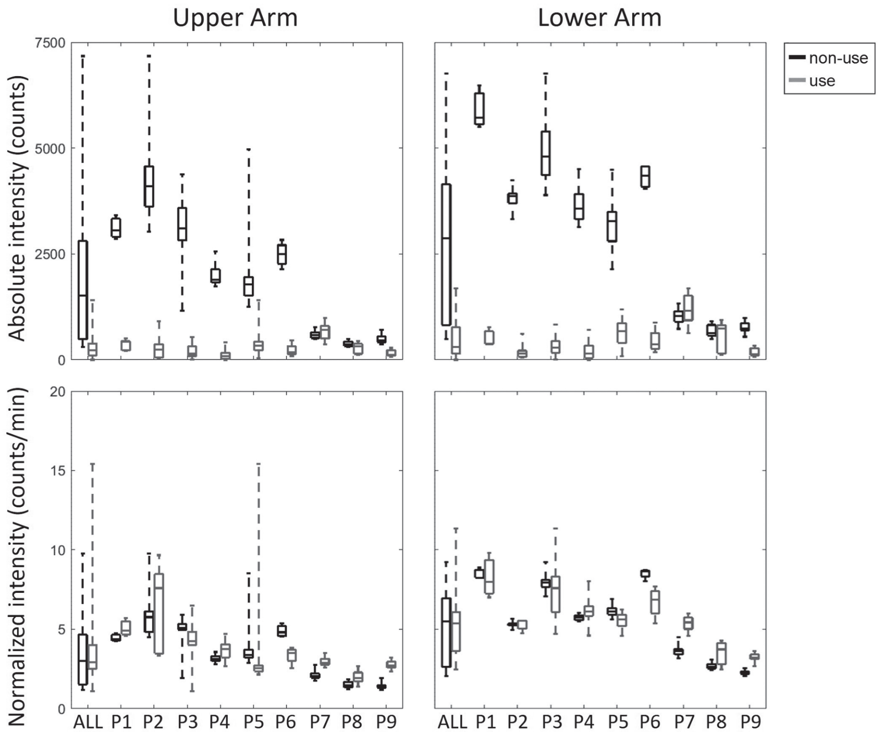
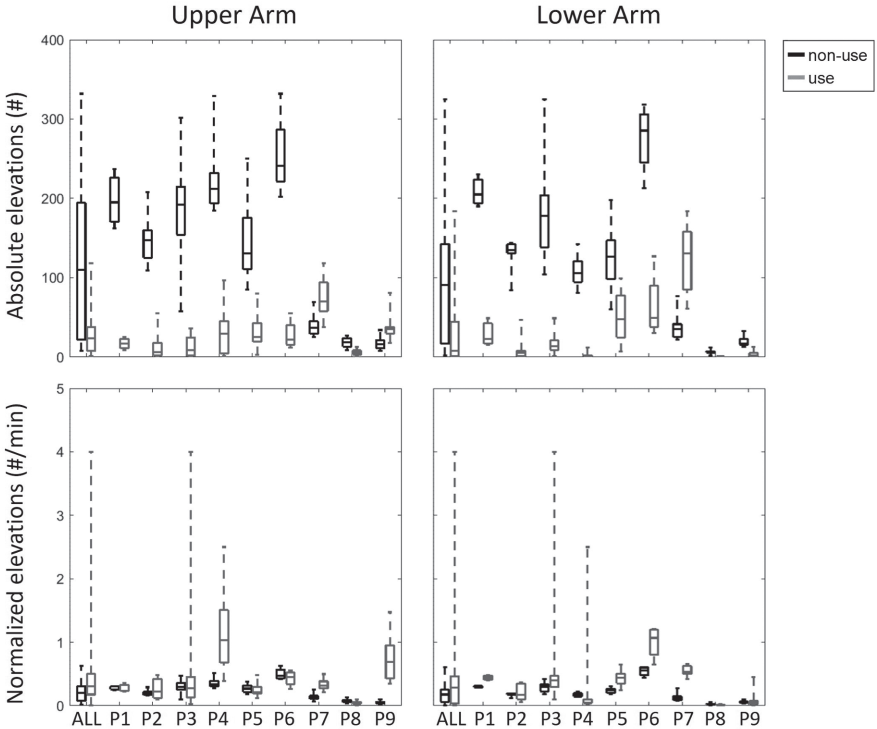
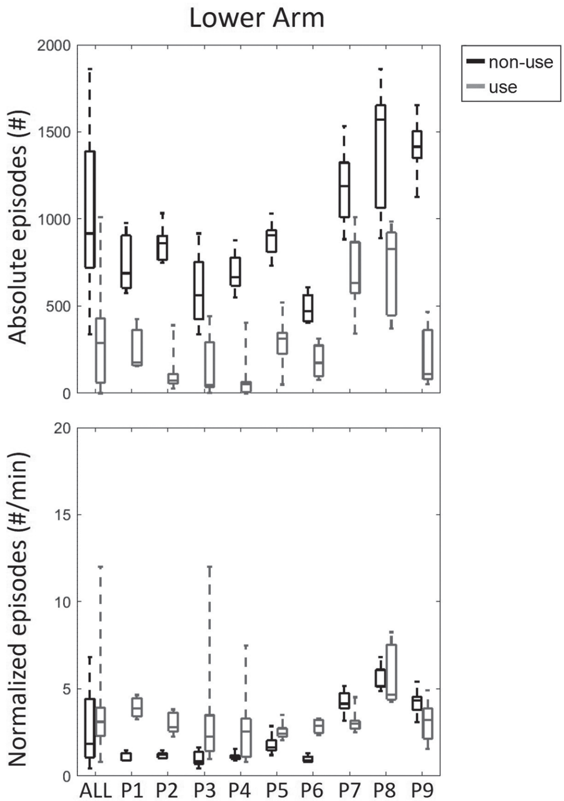


Appendix figure 11. Daily intensity levels for the upper (left) and lower arm (right) expressed in absolute (top) and normalized (bottom) values during DAS non-use (black) and use (grey). Boxplots represent 25th and 75th percentiles, the median is the box bar, and the whiskers represent the min and max.



Appendix figure 12. Daily arm elevations for the upper (left) and lower arm (right) expressed in absolute (top) and normalized (bottom) values during DAS non-use (black) and use (grey). Boxplots represent 25th and 75th percentiles, the median is the box bar, and the whiskers represent the min and max.



Appendix figure 13. Daily activity episodes for the lower arm expressed in absolute (top) and normalized (bottom) values during DAS non-use (black) and use (grey). Boxplots represent 25th and 75th percentiles, the median is the box bar, and the whiskers represent the min and max.

CHAPTER 6

General discussion

The aim of this thesis is to better understand how people with NMDs interact with dynamic arm supports to establish the factors that contribute to discontinued use. We started with a review of the literature and collected expert opinions (Chapter 2), we then performed an evaluation of the neuromechanical adaptations of people with NMDs (Chapter 3) and of the user-device interaction (Chapter 4) in a controlled environment, and lastly we evaluated the behavioural adaptations of user-device interaction in a home environment (Chapter 5).

This thesis describes several novel insights into the user-device interaction that may aid the development of dynamic arm supports. First, chapter 2 presents an overview and research recommendations of user-device interaction, synthesized from a scoping review with expert-based discussions, formulated in a novel framework of ICF components and contextual constructs. This study provides the essential clarity on present knowledge and its gaps so as to establish the contributing factors for discontinued use. Noteworthy is that the user-device interaction has different meanings and values to the different parties (users, developers, clinicians, and researchers) involved. The function of support, or the meaning and value of a device, depends greatly on the context where the device is used, rather than just the prospective improvements on upper extremity functionality. Second, chapter 3 is the first study to illustrate the neuromechanical adaptations of people with FSHD as muscle synergies. Therein, we found that even in simple shoulder joint tasks muscle synergy alterations are heterogeneous, which indicate that even small differences in task execution and individual characteristics have a large influence on motor control and output. Third, chapter 4 is the first study to investigate the user-device interaction of people with FSHD during tasks that mimic daily life activities. Its findings show that muscle synergy alterations in people with FSHD remain heterogeneous during tasks that are more complex. Furthermore, the dynamic arm support used in this study, Gowing, did not greatly affect the muscle synergies of people with FSHD as their similarity to those of healthy controls increased slightly, but their internal consistency remained lower than those of healthy controls. However, movement performance indicators were like those of healthy controls, without and with a dynamic arm support, indicating that the altered neuromuscular activations were effective for task execution. Finally, chapter 5 is novel for a) its approach to evaluate the behavioural adaptations of user-device interaction in b) a home environment over c) a relatively long period of time (up to three weeks). We showed that the support device facilitated motions against gravity and supported the occurrences of such motions, which did not change over time. Furthermore, using the device was beneficial for prolonged computer work, household chores, and personal hygiene, yet it also limited activities involving wrist movements, forearm rotation, and large motions of the arm.

6.1 NEUROMECHANICAL AND BEHAVIOURAL ADAPTATIONS OF PEOPLE WITH NMDS AND USER-DEVICE INTERACTION

People with FSHDs showed neuromechanical alterations, investigated as muscle synergies, compared to healthy controls. Generally, they displayed a more simplified muscle coordination, requiring fewer muscle synergies than healthy controls, to accomplish the same task. Yet, they also displayed more unique muscle coordination patterns, as the internal consistency of muscle synergies of FSHDs was lower than those of healthy controls. In addition, muscle synergies showed moderate similarities between FSHDs and healthy controls. While performing unsupported tasks, we showed that in FSHDs mainly the lower scapular rotators contributed less to the main synergy, while glenohumeral elevators contributed more than in healthy controls (chapters 3 and 4). Furthermore, the amount of lower scapular rotators' contribution while unsupported was dependent on the plane of elevation, where shoulder flexion/extension showed more muscle activity than shoulder abduction/adduction. This was also apparent in the muscle synergy alterations, where most participants with FSHD required distinct synergies while healthy controls utilized mostly the same coordination for both planes (chapter 3). In unsupported ADL, the alterations were less apparent due to the complexity of the tasks and the heterogeneity of muscle coordination in FSHDs (chapter 4). In general, persons with FSHD displayed a higher level of muscle co-contraction and higher neural activation than healthy controls.

The dynamic arm support effectively altered muscle synergies (chapter 4) and increased overall reachable workspace (chapter 2) and movement repeatability (chapter 5), and decreased movement speed (chapter 4). While, in contrast with healthy controls, the support used in chapter 4, Gowing, did not have a general effect on the muscle synergies in FSHDs, these synergies became more similar between the two populations. This suggests that muscle coordination in FHSDs could be made more similar to that of healthy person and could potentially be used to augment the benefits from the arm support. For example, improved strength and control of the scapula could increase the reachable workspace and allow for a higher movement repeatability [129] [130].

At home, the dynamic arm support improves the ability to independently perform self-care activities such as eating/drinking, doing household chores, and brushing teeth. Most users perceived these benefits in their daily life activities (chapter 5). However, the device has a limited range of motion in side- and downward directions and introduces collisions with the environment, causing disadvantages in performing ADL, such as eating/drinking and tooth brushing, if the environment is not suited to accommodate a device (chapter 2 and 5). These findings are in line with those from Gandolla et al. 2020 and Heide et al. 2016, indicating that the device provides benefits in the body function, activity, and participation components, but these benefits heavily depend on the environmental and personal factors [7, 57]. Apart from a few indications of mild discomfort reported in these studies and this thesis, there were no generalizable negative effects found in the musculoskeletal system in terms of disease progression, pain, and discomfort. In fact, Jansen et al. 2013 and 2015, showed that upper limb training with a dynamic arm support could delay functional deterioration in boys with Duchenne muscular dystrophy [56, 131]. Furthermore, Gandolla et al. 2020, emphasized that as the disability progresses, the benefits that a device can provide also change [57]. Therefore, proper augmentation of upper extremity functionality using dynamic arm support could increase and prolong quality of life.

Gandolla et al. 2020, also stated in their systematic review that perceived benefits are often higher than the functional gains identified through objective measures [57]. This stresses the fact that device development should consider the needs and capabilities of the user together with the environment to ensure proper application in daily life settings. However, this also means that the neuromechanical adaptations, as consequences of technological advances in dynamic arm supports, need to be assessed with respect to user capabilities and behavioural adaptations. This aspect is examined from the user-device interaction point of view in this thesis. So far, the translation from the neuromechanical consequences or functional gain to user benefits and behavioural adaptations, e.g. needs and capabilities, has been proven to be quite limited (Chapter 2 and 5). Moreover, due to the heterogeneity in capabilities and movement styles of the users, the user-device interaction from a select sample is difficult to interpret and generalize (Chapters 3, 4, and 5). Therefore, we propose a method to ensure transparency of individual results while providing meaningful implications for a broader user audience. To facilitate transparency and translation in user-device interaction, three findings of this thesis will be further discussed: 1) framework for assessing neuromechanics in people with NMDs and respective user-device interactions, 2) evidence and interactions within this framework, and 3) recommendations for future research. The latter will also be further addressed in the next chapter on valorization.

6.2 FRAMEWORK FOR ASSESSING NEUROMECHANICAL AND BEHAVIOURAL ADAPTATIONS IN PEOPLE WITH NMDS AND USER-DEVICE INTERACTIONS

We created a framework from the ICF model components (body functions, activity, and participations) and three contextual constructs (motor capacity, capability, and performance), to describe current evidence and identify knowledge gaps that limit dynamic arm support development. This approach allowed us to show what participants can do in a very controlled (motor capacity) and home environment (motor capability), and what they actually do on a daily basis (motor performance) depending on their impairments (body functions), limitations and restrictions (activity and participation), and other factors (personal and environmental). The above components and constructs are deeply connected, which improves the ability to translate effects between different combinations of components and constructs. For example, people with a neuromuscular disorder affecting the upper extremity have impaired muscular strength that limits their daily life activities involving food consumption and personal hygiene. The impaired muscular strength causes limitations in spatial and temporal kinematics and a disruption of coordinated muscle activations. For motor capacity, this means that participants had a smaller reachable workspace (chapters 2-4), movements were slower (chapter 2, 4), and control of the shoulder girdle was likely compromised (chapter 3 and 4). For motor capability and performance, the ability to do activities in a home environment are also limited (chapter 5). For example, lifting the hand towards the mouth repeatedly, and thus eating, will not be performed independently on a regular basis. Thus, an impaired upper extremity might limit the ability to eat at home and restrict social eating such as going to a restaurant, where a caregiver would be required. The reduction in physical activity and practice of independent eating will cause an accelerated decline and negatively affect the physical and mental condition, which is bound to worsen given the disease progressive nature. The idea is that independent eating (or similar activities) can be reinstated and the effects of disease progression slowed down by using a support device. However, categorizing the knowledge and expertise in a tabulation of ICF model components by contextual constructs revealed that studies seldom cover multiple combinations of the ICF model components and contextual constructs. Other researchers who evaluated user-device interaction effects in various settings [6, 12, 57, 118] derived similar conclusions, i.e. that translation between controlled and home settings are limited. The prominent reasons were that 1) the large variety of factors contributing to user-device interaction requires multiple time-intensive investigations of the different factors, with appropriate methods to unravel the correlation between factors, 2) time-continuity in these investigations is lacking, and 3) evidence in home settings is severely limited, which hampers translating the evidence to its intended environment. To overcome these barriers in transparency and translation of evidence it is imperative to find common ground between tabulated areas of the ICF model components by contextual constructs.

To find common ground, the measurement techniques used in this thesis were applied with their respective strengths within the structure of ICF model components and contextual constructs. For example, for body functions, we investigated motor control through kinematics using 3D motion capture, kinetics using load cells, muscle activity using electromyography, and muscle coordination using non-negative matrix factorization of electromyography data in a lab setting. This setup requires hours of preparation, constant surveillance of data quality, and it physically restricts the participant to the range of the equipment (i.e. cables or camera's field of vision). Therefore, investigating motor control is predominantly performed in a controlled setting (motor capacity). Furthermore, the interpretation of muscle coordination of the upper extremity has proven difficult, especially in this population, despite applying it in a controlled setting. The most notable difficulties were due to the a) heterogeneity presented in FSHD, b) many biomechanical degrees of freedom of the shoulder, c) variations in performance of activities of daily life, and d) individual responses in user-device interactions. People with FSHD seem to apply unique strategies reflecting individual-specific muscular weakness and loss of mobility. In addition, due to the many degrees of freedom the execution of activities of daily life can be potentially performed with a large movement variation. Finally, the gravity compensating force and added inertia of the support device causes individual-specific compensational strategies. Therefore, expanding a setup and analysis designed to investigate muscle coordination for motor capacity, so as to investigate motor performance (daily use) would require considerably more work on the experimental set-up.

For activity and participation, we used a network of accelerometers and a diary to investigate physical activity and mobility benefits at home (motor performance). This was relatively easy to set up, but the measurements were hard to interpret since context of recorded activity was very limited. While the diary did provide some contextual information in domains such as participation to social life, the indicated use of the device, based on diary records, was overestimated by up to three-fold compared to the accelerometer measures. A combination of objective and subjective measurement methods is recommended to understand and quantify mobility benefits during activities of daily life. Therefore, investigating motor capability in series to capacity and/or performance is a logical intermediate step. However, this requires equipment suitable for easy transportation and setup, customizable to a home environment. Along those lines, Heide et al. 2017 used inertial sensors to investigate the motor capacity as kinematics, an approach that could also potentially be used for motor capability. However, there they encountered similar limitations as we did, such as technical issues with the sensors, environmental and personal factors influencing ADL performance, and the users' inability to perform certain ADL without/with the device which led to missing data [12]. Therefore, technological advances and pragmatic choices regarding ADL and environmental and personal factors are necessary to deepen the understanding of the factors contributing to discontinued use.

6.3 RECOMMENDATIONS FOR FUTURE RESEARCH

Despite the relatively large evidence of user-device interaction displaying user benefits in daily use, discontinued use remains a problem, which cannot be completely explained by current evidence. Further research is needed with focus on collecting evidence connecting two or more combinations of ICF components and contextual constructs. For our recommendations, we highlight two important remaining knowledge gaps and include experience from preliminary evidence.

One of the most important yet largest gaps remaining is understanding the user-device interaction with respect to disease progression. Long-term studies are rare, although highly relevant since the discontinued use is hypothesized to be directly linked to disease progression. Currently, the evidence does not provide information on the full range of the impairment severity and lacks longitudinal measurements. Another gap is the evidence of performance in a home environment setting, as currently there are no studies identified in any of the ICF components for motor capability and three for motor performance. From our pilot testing and other studies [12, 28, 57, 118], the investigations seem to be limited by the current state of technology, complexity of daily use compared with standardized protocols, and complexity of environmental and personal factors. A recommended investigation would include unobtrusive wearables, which can measure the motor capacity, motor capability, and motor performance for an ICF component (sub) category, i.e. joint kinematics, as well as include information on the environmental and personal factors

of a larger number of dynamic arm support users, ranging in impairment severity. Furthermore, the needs and goals and perceived benefits and limitations should be established with respect to the ICF component (sub) category.
Beside the content presented in this thesis, we also conducted some pilot projects that were important to establish recommendations for future research. These projects included the use of 1) computerized musculoskeletal models [44, 132], 2) force and torque interactions between user and device, 3) reachable workspace analysis, 4) an inertial measurement unit (IMU) network, 5) surface electromyography (sEMG) for daily life setting, and 6) adaptable movement profiles for personalized support. Computerized musculoskeletal models seem promising to probe difficult to measure parameters, test cause-effect relationships, and test hypothetical changes [44]. However, key components, such as bone segment motions (i.e. scapula), individual muscle strengths, and compensatory motions and muscle activities are currently not understood well enough to create a representative model for muscular dystrophy. Therefore, an extension of current knowledge on body functionality and impairments (chapters 2, 3, and 4) is required. Along those lines, we recommend exploring the possibilities of muscle synergy analysis and improving robustness of data interpretation by isolating and creating various combinations of the four factors for complexity in muscle synergy analysis: the heterogeneity presented in FSHD, biomechanics of the shoulder, performance of activities of daily life, and user-device interaction. The intended use of measuring applied forces and torques between user and device was to create an adaptable movement profile for personalized support. Based on the user-device kinetics, the device could predict the user's intention (i.e. moving towards the left) and assist actively. However, in preliminary analysis we could not distinguish kinetic profiles for specific ADL with sufficient certainty. Most likely, the user-device kinetics are influenced by many components (i.e. impairments and environmental and personal factors), which complicated their interpretations. Future research could incorporate machine learning [133] to accurately distinguish kinetic profiles and personalize support for reoccurring ADL. Furthermore, unobtrusive wearable sensors were investigated to measure joint kinematics (IMU network) and muscle activity (sEMG) in a daily life setting. Then, the utilization of reachable workspace, defined as the areas a person can and does reach, combined with the required motor control and output could be investigated as an extension to chapters 3, 4, and 5. Unfortunately, current technology did not ensure unobtrusiveness and sufficient data quality and quantity. Technological advances are required in reducing dimensional size (Heide et al. 2017 [12] and chapter 5), increasing battery longevity (chapter 5), available sEMG channels from currently two to preferably >8 (chapter 3 and 4), and improving usability in a home environment (Heide et al. 2017 [12]).

6.4 CONCLUSIONS

To further improve our understanding of how people with neuromuscular disorders interact with dynamic arm supports and the contributing factors for discontinued use, future research needs to focus on three aspects. First, knowledge on body functionality and impairments needs to be extended. Currently, it is shown that people with FSHD are heterogeneous in their muscle coordination adaptations to muscular weakness and use of a dynamic arm support. However, it remains unclear how these adaptations change over time due to disease and experience with the support device. The cross-sectional observation of inexperienced participants showed that muscle coordination in people with FSHD has the plasticity to become more similar to that of healthy controls. Therefore, it is recommended to investigate how to exploit this plasticity, within the limits of a person's motor capabilities, for therapeutic means and to augment the support benefits.

Second, daily life use of dynamic arm supports need to be monitored over the long term to include disease progression and experience with the device. Currently, knowledge is limited to cross-sectional observations or to feedback regarding discontinued use. The exact reasons for the latter are difficult to extract due to, among others, the complexity in quantifying the user's capabilities to perform activities of daily life and to understand how the device can augment the performance of the user. Thus, it is important to monitor and connect the neuromechanical and behavioural adaptations to daily life use in a home environment, preferably when receiving the dynamic arm support.

Third, a framework is required that facilitates the translation between what users can do and want to do and how a support device may assist in various environments. Technological advances and a design encompassing the ICF model and contextual constructs are required to gather this evidence. Therefore, pragmatic choices and intermediate steps within the proposed framework are necessary to continue this journey.

CHAPTER 7

Valorisation

The project was part of the STW-perspective Symbionics program (project 13523)[134] and we collaborated with various stakeholders to create value from the generated knowledge in this PhD project. They were mostly from a muscular dystrophy patient community, support developer, clinical, rehabilitation, and research setting. This chapter will elaborate on the collective value focusing on the evidence of knowledge of muscular dystrophy and user-device interactions and the established framework to describe them.

7.1 EVIDENCE OF DISEASE AND USER-DEVICE INTERACTIONS

From the evidence found in this work, it is clear that the manifestation of a neuromuscular disorder is heterogeneous and so is the respective interaction between user and support device. More specifically, the coordination of muscle activities of people with Facioscapulohumeral dystrophy (FSHD) was found to be more unique, yet simplified, than those of healthy participants. Likely, their muscle coordination was adapted as result of individual-specific weaknesses present in the muscle groups that mobilize the shoulder, upper arm, and scapula. On top of these unique variations of muscular weakness, the versatility of the shoulder's motions adds another layer of complexity in coordinating muscle activities. It is therefore clear that future research and treatment should focus on individuals-specific alterations of muscle coordination in people with Facioscapulohumeral dystrophy and likely in other muscular dystrophy types. Meanwhile, the use of a dynamic arm support did not generally influence FSHD people's muscle coordination, while this was true for healthy participants.

However, muscle coordination did become more similar between populations using the device. With training, it might be possible to fine-tune the user-device interaction and augment the benefits received from the support. Although users could undergo specific training to operate the device more efficiently, it also seems plausible that the user can extend the support's benefits by improving the muscle coordination to a more stable scapula. The lack of scapular motion and stability limits, among others, the available strength in motions against gravity and consequently the reachable workspace. Muscle activity-based biofeedback is a possible therapeutic approach to achieve this goal. This approach should be applied, evaluated, and adjusted based on the progressive nature of muscular dystrophy.

An alternative approach would be to adapt the device to the capabilities and needs of its user. At home, the dynamic arm support was most used to consume food and to perform personal hygiene tasks, household chores, and computer activities. Users indicated that the device provided no support for or even limited activities involving wrist movements, forearm rotation, and large motions of the arm. Unfortunately, the consequences for daily use, indicated in chapter 5 as 74% of all activities by the user but only as 18% by the activity monitors, remain unclear. Thus, efficient adjustments to the device require a better understanding of the disease and user-device interaction, as well as an understanding of the contributing factors for the discontinued use of the device. Collaboration between stakeholders is necessary to fill these gaps. By exploiting each stakeholder's strength, the patient community can formulate the users' needs, estimate the users' capabilities, and formulate the biomechanical adjustments of the dynamic arm support device to fulfil these needs. Consequently, a common ground of shared knowledge and environment is required to facilitate this collaboration.

7.2 FRAMEWORK FOR DISEASE AND USER-DEVICE INTERACTIONS

Previously, some of the stakeholders commonly used the ICF model components, body functions, activity and participation, and personal and environmental factors, to describe the characteristics of neuromuscular disorders and user-device interactions. To properly understand the context of these characteristics and interactions, additional information such as the setting in which they were established and consequences for their daily lives are required. However, from literature and collaborations with stakeholders it became clear that there was no structure yet in existence that would facilitate this understanding. Therefore, a framework was created from the ICF model and incorporated three contextual constructs, which explains what people can do in controlled (motor capacity) and home environment (motor capability), and what they do daily (motor performance). As a result, the gaps in knowledge became quite clear. For example, stakeholders expressed the necessity to understand what users do with the dynamic arm support, what the beneficiaries are, how that affects body functions and activity and participation, and the interactions thereof. Yet, current evidence was found to be more focussed on the controlled environment and greatly lacking in the home environment. After evaluation of the available evidence, it is established that gathering evidence in the home environment is very complex and only recently gaining traction. Therefore, this framework enables the stakeholders to tackle this and similar issues by identifying the collective strengths and remaining barriers in various framework areas.

7.3 TAKE HOME MESSAGES

The capabilities, needs, and interactions with a dynamic arm support should always be attuned to an individual's capabilities. This means that the device should be fitted to the user and adjustments should be made in due time, for example, following disease progression. However, the possibility to train the user to extend the support benefits should also be investigated. Furthermore, the proposed framework for disease and user-device interactions should be used to fill in remaining knowledge gaps stated in chapter 2, such as the connection between user's capabilities and performance of activities in daily life and the adaptations in user-device interaction over time and due to disease progression. Ultimately, we should strive towards the collection of evidence in a home environment and towards monitoring disease progression. To this aim, collaboration among stakeholders is necessary for a better understanding of disease and user-device interactions and the consequent improvements in quality of life.

7

CHAPTER 8

Summary

8.1 ENGLISH SUMMARY

Upper extremity strength and mobility is impaired in people with neuromuscular disorders. As a result, people with neuromuscular disorders generally have limitations in eating/drinking and performing personal hygiene activities, which leads to restrictions in daily life. A dynamic arm support can help to alleviate some parts of these barriers. Consequently, people using an arm support are able to regain some motor capabilities and independence. However, the satisfaction level with the support device is generally low with a relatively high discontinuation rate over longer periods of time, which indicates that improvements are necessary.

This thesis focussed on the consequences of upper extremity muscular weakness in neuromuscular disorders and the interaction with a dynamic arm support. First, we combined recent evidence with current expert opinions to create a structured overview and facilitate future research (Chapter 2). Subsequently, the following chapters addressed a few of the identified research gaps. To this effect, the impact of muscular weakness on body functions and activity limitations was investigated in a controlled environment to provide a better understanding of the interactions between these components (Chapter 3). Furthermore, the impact of a dynamic arm support, Gowing, was investigated in that same environment in addition to the previous interactions (Chapter 4). Then, the impact of a dynamic arm support at the users' homes was investigated over the course of a few weeks to collect evidence in a real-life environment (Chapter 5). Finally, these chapters were followed by a general discussion (Chapter 6) and a valorisation (Chapter 7) to facilitate collaboration amongst stakeholders, such as arm support users, developers, clinicians, and researchers.

8.1.1 A framework to better understand disease and user-device interactions

The structured overview and established framework (chapter 2) provided evidence of the impact dynamic arm supports had on body functions and daily life activities of people with neuromuscular disorders. This evidence was synthesized from current literature and interviews with stakeholders from a patient community, support developers, clinical, rehabilitation, and research settings. The evidence included eight published articles, two non-peer reviewed articles, and fifteen stakeholders. The resulting framework facilitates the ability to describe what people are able to do in a very controlled (motor capacity) and in a home environment (motor capability), and what they actually do on a daily basis (motor performance) depending on their impairments (body functions), limitations and restrictions (activity and participation), and personal and environmental factors. We found that current literature mostly investigated the motor capacity of muscle function, joint mobility, and upper body functionality, and a few studies also addressed the impact on activity and participation. In addition, experts considered knowledge on device utilisation in the daily environment as important. Evidence showed that people with a neuromuscular disorder had a smaller reachable workspace (chapters 2-4), their movements were slower (chapter 2, 4), and their control of the shoulder girdle was compromised (chapter 3 and 4) compared with healthy participants and activities performed in a controlled environment. Similarly, their ability to perform these tasks at home was limited, as seen for example, from a reduced frequency of lifting their hand to mouth to independently eat and drink without a dynamic arm support (chapter 5). The dynamic arm support increased the overall reachable workspace (chapter 2 and pilot testing) and the movement repeatability at home (chapter 5), altered the motor control (chapter 4), and decreased movement speed (chapter 4). Furthermore, most users perceived benefits from using a dynamic arm support in their daily life activities (chapters 2 and 5). However, the limited range of motion in the other directions than against gravity and its increased risk for collisions with the environment, may cause disadvantages in performing daily life activities if the environment is not suited to accommodate the device (chapter 2 and 5).

8.1.2 Motor control alterations due to disease and interactions with a dynamic arm support

People with Facioscapulohumeral dystrophy (FSHD) have progressive loss of muscle strength, mostly in the shoulder area, and consequently scapular winging, joint instability, and a decline in upper extremity functionality. This leads to compensatory strategies that require increased effort and difficulties when performing activities of daily life. Therefore, in chapter 3 and 4, we investigated the motor control of FSHD with novel techniques to verify certain aspects of their body functions and the impact of a dynamic arm support. We measured the muscle activity of eight shoulder muscles and kinematics of the upper extremity of fourteen participants with FSHDs and fourteen healthy controls performing several tasks. Motor control was investigated as muscle synergies extracted via non-negative matrix factorization of electromyography data. Kinematic data was used to extract shoulder joint motion and task performance. First, we focused on select body functions and motor capacity in chapter 3 by isolating motion from the shoulder girdle in two planes, frontal and sagittal. Second, we integrated tasks that mimic activities of daily life and required multi-joint motion in chapter 4. Third, the impact of a dynamic arm support was investigated by repeating these tasks with a support device in chapter 4.

Chapter 3 revealed that motor control is altered in FSHD compared to healthy controls during upper arm elevation in the frontal and sagittal plane. In general, two muscle synergies were sufficient in both populations. The first synergy accounted for the vast majority (50-74%) of muscle activity variance in both populations and planes of motion and the second synergy for most of the remaining variance. Furthermore, the lower scapular rotators contributed less and compensatory activity was found for muscles surrounding the glenohumeral joint in FHSDs. The alterations in FSHDs were different for the planes of motion suggesting that motor control of the lower scapular rotators affected the scapulohumeral rhythm. Overall, participants with FSHD displayed less muscle activity and reached less high in the frontal than the sagittal plane and in both planes less than healthy controls.

Chapter 4 showed the impact of a dynamic arm support, Gowing, in FSHD with respect to healthy for five tasks that mimic activities of daily life. These tasks were 1) pushing and pulling an object, 2) simulated drinking with a cup of 200 grams, 3) simulated eating with a spoon, and 4) reaching towards a target at shoulder height on the ipsilateral side and 5) on the contralateral side. We used Pearson's correlation coefficient (r) to express the consistency, correlations within populations, and similarity, correlations between populations, of motor control under influence of disease and the interaction with a dynamic arm support. Furthermore, task performance was extracted from kinematic data as task duration, smoothness, and efficiency. First, up to four synergies were extracted where >70% of the participants generally required two synergies to perform a task. The number of extracted synergies were not significantly different between support conditions for each population. The first and second ranked synergies in terms of variance accounted for were considered for further analysis. Second, when comparing populations without the use of a dynamic arm support, the motor control was found to be altered and less consistent in FSHDs than in healthy controls (first r: -0.34 and second synergy r: -0.41). Third, when introducing the dynamic arm support, both populations showed alleviated muscle efforts, but only controls showed to have an affected motor control from the facilitated arm elevation (r: +0.25 to +0.40). Fourth, similarity between populations increased (r: +0.12), yet for FSHDs, the internal consistency remained unaffected and lower than that of healthy controls.
To summarize, in chapters 3 and 4 we found that motor control is altered, less consistent, and less affected by the arm support, Gowing, in FSHDs compared with healthy controls. Furthermore, these alterations appeared to be affected by plane of motion and activity of daily life. Also, the large group variances indicate that individual characteristics, such as individual-specific deficits of muscle weakness and respective development of compensatory strategies, have a large

influence on motor control. Therefore, an assessment of the muscles' coordination is recommended to reveal individual synergies and to design evidence-based therapy for the management of the condition. Finally, the biomechanical consequences of using an arm support should be further investigated in people with FSHD on deeper-layered shoulder muscles and to evaluate long-term benefits.

8.1.3 Daily life benefits and usage characteristics of dynamic arm supports

The impact of a dynamic arm support in a free-living environment was addressed in chapter 5 in order to establish a link between body functions and activity and participation. Participants with a neuromuscular disorder that used a dynamic arm support were monitored in daily life through activity trackers and self-reports. We integrated environmental and personal factors, as the perceived benefits of the devices, and long-term measurements by monitoring the use for up to three weeks. The integration of the multi-sensor network and self-reports provided context to dynamic arm support use and their daily life benefits. These benefits were experienced mainly during activities involving movement against gravity. Furthermore, the measured use did not change over time. However, self-reports overestimated the actual use by up to three-fold compared to the activity tracker's measures. A combination of objective and subjective methods is recommended for meaningful and quantifiable mobility benefits during activities of daily life. However, the assessment methods need to be simplified to reduce the burden on the participant. Furthermore, to determine the mobility benefits and device effectiveness, integration of relevant International Classification of Functioning, disability, and health (ICF) components is necessary.

8.1.4 Take home messages

Several lessons were learned during this research project. First, to facilitate a better understanding of how people with a neuromuscular disorder interact with dynamic arm supports the framework proposed above should be used. Thereby, it is important to incorporate what users can do and want to do, reflecting their capabilities and needs in various environments, respectively. Second, the evidence in this framework should be expanded, including the influence of personal and environmental factors when developing and deploying a device. Third, short-term and long-term measurements should be included to monitor adaptations over time. Adaptations include disease progression, but also the effect of training with a device and other changes in all framework combinations. Fourth, include user satisfaction as guidance to evaluate the device effectiveness. In line with users' capabilities and needs, the end goal is to enhance the quality of life experience by the user.

Dynamic arm supports have the potential to do so, yet the current platform shaped by the efforts of stakeholders seems to not be sufficiently supported to fulfil this goal in the long term. Therefore, collaborations across expert fields, such as arm support users, developers, clinicians, and researchers, are necessary to create a better understanding of the disease and user-device interaction.

8.2 NEDERLANDSE SAMENVATTING

De kracht en mobiliteit van de bovenste extremiteit is verminderd bij mensen met neuromusculaire aandoeningen. Als gevolg hiervan hebben mensen met neuromusculaire aandoeningen over het algemeen beperkingen bij het eten/drinken en het uitvoeren van persoonlijke hygiënische activiteiten, wat leidt tot beperkingen in het dagelijks leven. Een dynamische armondersteuning kan helpen om sommige van deze beperkingen te verlichten. Daardoor kunnen mensen die een armondersteuning gebruiken weer wat motoriek en zelfstandigheid terugkrijgen. De tevredenheid is echter over het algemeen laag met een relatief hoog percentage van stopzettingen over een langere periode, wat erop wijst dat verbeteringen noodzakelijk zijn.

Dit thesis richtte zich op de gevolgen van spierzwakte van de bovenste extremiteit bij neuromusculaire aandoeningen op lichaamsfuncties en dagelijkse activiteiten en de interactie met een dynamische armondersteuning. Ten eerste hebben we recent bewijs uit de literatuur gecombineerd met actuele meningen van experts om een gestructureerd overzicht te creëren en toekomstige onderzoeksvoorstellen te vergemakkelijken (hoofdstuk 2). Deze onderzoeksrichtingen werden vervolgens in de opvolgende hoofdstukken behandeld om enkele van de geïdentificeerde hiaten in kennis aan te pakken. Ten tweede werd de impact van spierzwakte op lichaamsfuncties en dagelijkste activiteiten onderzocht in een gecontroleerde omgeving om een beter begrip te krijgen van de interacties tussen deze componenten (hoofdstuk 3). Ten derde werd de impact van een dynamische armondersteuning, Gowing, in diezelfde omgeving onderzocht naast de eerdere interacties (hoofdstuk 4). Ten vierde werd in de loop van een paar weken de impact van een dynamische armondersteuning onderzocht in de dagelijkse leefomgeving van de gebruikers om van kennis over een gecontroleerde naar een thuisomgeving over te hevelen (hoofdstuk 5). Ten slotte werden deze hoofdstukken gevolgd door een algemene discussie (hoofdstuk 6) en de valorisatie (hoofdstuk 7) om de samenwerking tussen belanghebbenden, zoals gebruikers van armondersteuningen, ontwikkelaars, clinici, en onderzoekers, te vergemakkelijken.

8.2.1 Een kader om ziekte en gebruiker-apparaat interacties beter te begrijpen

Het gestructureerde overzicht en het vastgestelde kader (hoofdstuk 2) leverden bewijs van de impact van dynamische armsteunen op lichaamsfuncties en dagelijkse levensactiviteiten van mensen met neuromusculaire aandoeningen. Dit bewijs werd gesynthetiseerd uit de huidige literatuur en interviews met belanghebbenden uit de patiënten gemeenschap, ontwikkelaars van armondersteuningen, klinische, revalidatie en onderzoek omgevingen. Het bewijsmateriaal omvatte acht gepubliceerde artikelen, twee niet-peer-reviewed artikelen en vijftien belanghebbenden. Het resulterende raamwerk maakt het mogelijk te beschrijven wat mensen kunnen doen in een zeer gecontroleerde (motorische capaciteit) en in een thuisomgeving (motorische vermogen), en wat ze daadwerkelijk dagelijks doen (motorische prestatie), afhankelijk van hun beperkingen (lichaamsfuncties), barrières en restricties (activiteit en participatie), en persoonlijke en omgevingsfactoren. Wij vonden dat de huidige literatuur vooral de motorische capaciteit van spierfunctie, gewrichtsmobiliteit en functionaliteit van het bovenlichaam onderzocht, en een paar studies ook in gingen op de impact op activiteit en participatie. Daarnaast vonden experts kennis over het gebruik van hulpmiddelen in de dagelijkse omgeving belangrijk. Uit de gegevens bleek dat mensen met een neuromusculaire aandoening een kleinere bereikbare werkruimte hadden (hoofdstuk 2-4), dat hun bewegingen trager waren (hoofdstuk 2, 4) en dat hun controle over de schoudergordel aangetast was (hoofdstuk 3 en 4) in vergelijking met gezonde deelnemers en activiteiten in een gecontroleerde omgeving. Ook was hun vermogen om deze taken thuis uit te voeren beperkt, zoals blijkt uit de verminderde frequentie waarmee zij hun hand naar de mond brengen om zelfstandig te eten en te drinken zonder dynamische armondersteuning (hoofdstuk 5). De dynamische armondersteuning vergrootte de totale bereikbare werkruimte (hoofdstuk 2 en proeftesten), de herhaalbaarheid van bewegingen thuis (hoofdstuk 5), veranderde de motorische controle (hoofdstuk 4) en verminderde de bewegingssnelheid (hoofdstuk 4). Bovendien beleefden de meeste gebruikers voordelen van het gebruik van een dynamische armondersteuning bij hun dagelijkse activiteiten (hoofdstuk 2 en 5). Het beperkte bewegingsbereik in de andere richtingen dan tegen de zwaartekracht in en het verhoogde risico op botsingen met de omgeving kunnen echter nadelen veroorzaken bij het uitvoeren van activiteiten in het dagelijks leven als de omgeving niet geschikt is voor het hulpmiddel (hoofdstuk 2 en 5).

8.2.2. Veranderingen in motorische controle ten gevolge van ziekte en interacties met een dynamische armondersteuning

Mensen met Facioscapulohumerale dystrofie (FSHD) hebben een progressief verlies van spierkracht, vooral in het schoudergebied, en als gevolg daarvan een afstaand schouderblad, gewrichtsinstabiliteit en een afname van de functionaliteit van de bovenste extremiteit. Dit leidt tot compensatiestrategieën die meer inspanning vergen en het moeilijk maken om activiteiten van het dagelijks leven uit te voeren. Daarom hebben we in hoofdstuk 3 en 4 de motorische controle van FSHD onderzocht met nieuwe technieken om bepaalde aspecten van hun lichaamsfuncties en de impact van een dynamische armondersteuning na te gaan. We maten de spieractiviteit van acht schouderspieren en de kinematica van de bovenste extremiteit van veertien deelnemers met FSHD en veertien gezonde controles die verschillende taken uitvoerden. Motorische controle werd onderzocht als spiersynergiën geëxtraheerd via niet-negatieve matrix factorisatie van elektromyografische data. Kinematische gegevens werden gebruikt om de beweging van het schoudergewricht en de taakuitvoering te extraheren. Eerst richtten we ons op geselecteerde lichaamsfuncties en motorische capaciteit in hoofdstuk 3 door beweging van de schoudergordel in twee vlakken, frontaal en sagittaal, te isoleren. Ten tweede integreerden wij in hoofdstuk 4 taken die activiteiten van het dagelijks leven nabootsen en beweging van meerdere gewrichten vereisen. Ten derde werd het effect van een dynamische armondersteuning onderzocht door deze taken te herhalen met een ondersteuningsapparaat in hoofdstuk 4.

Hoofdstuk 3 liet zien dat de motorische controle veranderd is bij FSHD in vergelijking met gezonde controles tijdens bovenarmheffing in het frontale en sagittale vlak. In het algemeen waren twee spiersynergiën voldoende in beide populaties. De eerste synergie was verantwoordelijk voor de overgrote meerderheid (50-74%) van de variantie in spieractiviteit van beide populaties en bewegingsvlakken en de tweede synergie voor het grootste deel van de resterende variantie. Verder droegen de onderste schouderblad rotators minder bij en werd compenserende activiteit gevonden voor spieren rond het glenohumerale gewricht in FHSDs. De veranderingen in FSHDs waren verschillend voor de bewegingsvlakken wat suggereert dat motorische controle van de onderste schouderblad rotators het scapulohumerale ritme beïnvloedde. Over het algemeen vertoonden deelnemers met FSHD minder spieractiviteit en reikten ze minder hoog in het frontale dan het sagittale vlak en in beide vlakken minder dan gezonde controles.

Hoofdstuk 4 toonde het effect van een dynamische armondersteuning, Gowing, bij FSHD ten opzichte van die bij gezonde controles voor vijf taken die activiteiten van het dagelijks leven nabootsen. Deze taken waren 1) duwen en trekken aan een voorwerp, 2) gesimuleerd drinken met een beker van 200 gram, 3) gesimuleerd eten met een lepel, en 4) reiken naar een doel op schouderhoogte aan dezelfde zijde en 5) aan de tegengestelde zijde. Wij gebruikten Pearsons correlatiecoëfficiënt (r) om de consistentie, correlaties binnen populaties, en de overeenkomst, correlaties tussen populaties, van de spiersynergiën onder invloed van de ziekte en de interactie met een dynamische armondersteuning uit te drukken. Daarnaast werden uit de kinematische gegevens taakprestaties geëxtraheerd als taakduur, soepelheid en efficiëntie. Ten eerste werden maximaal vier synergiën geëxtraheerd, waarbij >70% van de deelnemers doorgaans twee synergiën nodig hadden om een taak uit te voeren. Het aantal geëxtraheerde synergiën verschilde niet significant tussen de ondersteuningsvoorwaarden voor elke populatie. De als eerste en tweede gerangschikte synergiën in termen van verklaarde variantie werden vervolgens meegenomen voor verdere analyse. Ten tweede, bij het vergelijken van populaties zonder het gebruik van een dynamische armondersteuning, bleken de spiersynergiën bij FSHDs veranderd en minder consistent te zijn dan bij gezonde controles (eerste r: -0,34 en tweede synergie r: -0,41). Ten derde, bij invoering van de dynamische armondersteuning vertoonden beide populaties verminderde spierinspanningen, maar alleen de controles vertoonden beïnvloede spiersynergiën door de gefaciliteerde armheffing (r: +0,25 tot +0,40). Ten vierde nam de overeenkomst tussen populaties toe (r: +0,12), maar voor FSHDs bleef de interne consistentie onaangetast en lager dan die van gezonde controles.

Samenvattend vonden we in hoofdstuk 3 en 4 dat de motorische controle, onderzocht als spiersynergiën, veranderd is, minder consistent is en minder beïnvloed wordt door armondersteuning, Gowing, bij FSHDs in vergelijking met gezonde controles. Verder bleken deze veranderingen beïnvloed te worden door het bewegingsvlak en de activiteit van het dagelijks leven. Ook wijzen de grote groepsvariaties erop dat individuele kenmerken, zoals individu-specifieke tekorten van spierzwakte en respectieve ontwikkeling van compensatiestrategieën, een grote invloed te hebben op de motorische controle. Daarom wordt een beoordeling van de coördinatie van de spieren aanbevolen om individuele spiersynergiën bloot te leggen en kennis gedreven therapie te ontwerpen voor het beheer van de aandoening. Ten slotte moeten de biomechanische gevolgen van het gebruik van een armondersteuning bij mensen met FSHD op de dieper gelegen schouderspieren verder worden onderzocht en de voordelen op lange termijn worden geëvalueerd.

8.2.3 Voordelen voor het dagelijks leven en gebruikskenmerken van dynamische armsteunen

De impact van een dynamische armondersteuning in een vrije leefomgeving is werd behandeld in hoofdstuk 5 om een verband te leggen tussen lichaamsfuncties en activiteit en participatie. Deelnemers met een neuromusculaire aandoening die een dynamische armondersteuning gebruikten, werden in het dagelijks leven gevolgd door middel van activiteitenmonitors en zelfrapportages. We integreerden omgevings- en persoonlijke factoren, zoals de waargenomen voordelen van de armondersteuningen, en de lange termijn metingen door het gebruik tot drie weken lang te volgen. De integratie van het meerdere sensor netwerk en de zelfrapportage bood context voor het gebruik van dynamische armondersteuningen en de voordelen ervan in het dagelijks leven. Deze voordelen werden vooral ervaren tijdens activiteiten waarbij tegen de zwaartekracht in werd bewogen. Bovendien veranderde het gemeten gebruik niet in de loop van de tijd. Zelfrapportages overschatten echter het werkelijke gebruik tot driemaal zo hoog als de metingen van de activiteitenmonitor. Een combinatie van objectieve en subjectieve methoden wordt aanbevolen voor zinvolle en kwantificeerbare mobiliteitsvoordelen tijdens activiteiten van het dagelijks leven. De beoordelingsmethoden moeten echter worden vereenvoudigd om de belasting voor de deelnemer te verminderen. Bovendien is voor het bepalen van de mobiliteitsvoordelen en de doeltreffendheid van de armondersteuning integratie van relevante ICF-componenten (International Classification of Functioning, disability, and health) noodzakelijk.

8.2.4 Belangrijkste boodschappen

Tijdens dit onderzoeksproject zijn verschillende lessen geleerd. Ten eerste moet het hierboven voorgestelde kader worden gebruikt om beter te begrijpen hoe mensen met een neuromusculaire aandoening omgaan met dynamische armondersteuningen. Daarbij is het van belang te incorporeren wat gebruikers kunnen en willen doen, als afspiegeling van respectievelijk hun capaciteiten en behoeften in verschillende omgevingen. Ten tweede moet het bewijsmateriaal in dit kader worden uitgebreid, met inbegrip van de invloed van persoonlijke en omgevingsfactoren bij de ontwikkeling en invoering van een hulpmiddel. Ten derde moeten kort termijn en lange termijn metingen worden opgenomen om aanpassingen in de tijd te volgen. Aanpassingen omvatten ziekteprogressie, maar ook het effect van training met een armondersteuning en andere veranderingen in alle combinaties van het kader. Ten vierde, de tevredenheid van de gebruiker moet opgenomen worden als leidraad om de doeltreffendheid van de armondersteuning te evalueren. In overeenkomst met de mogelijkheden en behoeften van de gebruikers is het einddoel om de levenskwaliteit die de gebruiker ervaart te verbeteren. Dynamische armondersteuningen hebben het potentieel om dat te doen, maar het huidige platform, dat door de inspanningen van belanghebbenden tot stand is gekomen, lijkt onvoldoende ondersteund om dit doel op lange termijn te bereiken. Daarom is samenwerking tussen deskundigen, zoals gebruikers van armondersteuningen, ontwikkelaars, clinici en onderzoekers, nodig om een beter inzicht te krijgen in de ziekte en de interactie tussen gebruiker en armondersteuning.

CHAPTER 9

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CHAPTER 10

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CHAPTER 11

Curriculum Vitae

ABOUT THE AUTHOR

Johannes Maria Nicolaas (Hans) Essers was born on the 11th of January 1987 in Maastricht, the Netherlands. After graduating from secondary school (Porta Mosana, Maastricht) he partook in the then relatively new bachelor program Biometrie at Hogeschool Zuyd Heerlen between 2005 and 2009. This program focusses on engineering and using applications for human related purposes in a clinical, scientific, and athletic setting. Due to his passion for research, he consequently followed the master program Human Movement Sciences at University Maastricht between 2009 and 2010. For his master thesis, he created a digital goniometer by himself that bypasses the rotational point by using simple materials and techniques and validated the device with the help of physiotherapy students and guidance of Alessio and Kenneth. Afterwards, he started as a research assistant at the department of Human Movement Sciences (now Nutrition and Movement Sciences, NMS) at University Maastricht. Being involved in several projects, he pursued a career in research by applying to a PhD project specifically involving robotics and biomechanics in humans. This PhD project was a joined effort of among others Maastricht University Medical Centre and University Medical Centre Groningen with experiments from chapter 3 and 4 at the latter location. Parallel to his PhD project, he collaborated with many other researchers and had various educational roles. Consequently, he wishes to combine his passion for engineering, biomechanical research, and education as a (practical) teacher at the NMS department. As such, Hans is currently employed as a university teacher at University Maastricht.

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