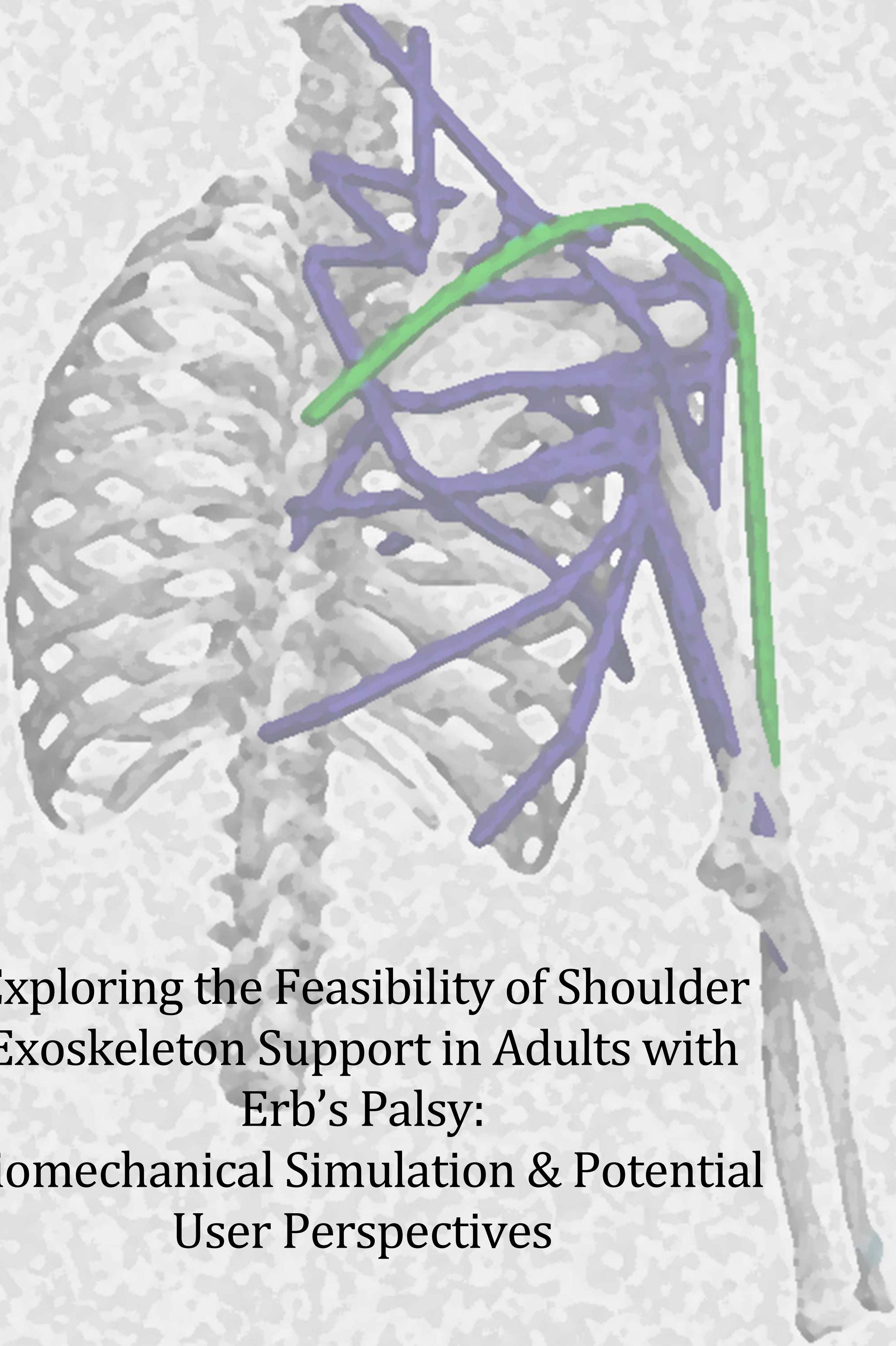

Exploring the Feasibility of Shoulder Exoskeleton Support in Adults with Erb's Palsy: Biomechanical Simulation & Potential User Perspectives

By Caitlyn Kramer

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Abstract

Background: Shoulder muscle weakness (e.g. in Erb's palsy) limits the ability to lift the arm and requires high muscular activation to perform and sustain elevation leading to rapid fatigue. This leads to a restricted reachable workspace resulting in impairments in daily activities . Although passive shoulder exoskeletons based on gravity compensation reduce muscle activity in healthy individuals, the effects on muscle activation, shoulder kinematics and activities of daily life under reduced shoulder strength conditions remains unclear. To our knowledge, no orthosis/exoskeleton is currently available for supporting arm elevation through gravity-compensation principles during activities of daily life.

Methods: This mixed-methods feasibility study evaluated whether passive gravity-compensating shoulder assistance can restore muscle-driven elevation capacity and reduce muscular activation under graded weakness. Additionally, user-informed design requirements were identified using semi-structured interviews exploring daily limitations and needs for support in adults with Erb's palsy. In parallel, musculoskeletal simulations in OpenSim modelled abduction and forward flexion under strength reductions corresponding to MRC levels 2-4. Outcomes included peak and cumulative muscle activation and the resulting maximal muscle-driven active humerothoracic elevation range of motion. Assistance was implemented using gravity scaling and a physically modelled cam-cable actuator.

Results: Interview findings highlighted rapid fatigue, restricted elevation, and reliance on compensatory movement strategies, alongside strong requirements regarding comfort, adjustability and selective use. Simulations showed that weakness reduced maximal muscle-driven range of motion. Passive assistance restored full elevation with 15-45% gravitational compensation, depending on severity of weakness, while reducing peak and cumulative activation without disproportionate compensatory activation.

Conclusion: Passive shoulder elevation support, implemented through both gravity scaling and a modelled cam-cable actuator, under simulated weakness and aligns with user-identified needs, supporting further orthosis development.

Index

1 Introduction

1.1 Shoulder Muscle Weakness and Functional Impact

Shoulder muscle weakness is a reduced force capacity of the shoulder muscles. In many conditions, both conservative and surgical interventions fail to achieve full functional recovery, resulting in persistent impairments in activities of daily living (1-7).

This impairment may arise from peripheral neurological conditions, such as brachial plexus injuries including Erb's palsy and cervical radiculopathy, central neurological conditions such as stroke and cerebral palsy (CP), neuromuscular disorders such as Duchenne or Becker muscular dystrophy and multiple sclerosis (MS), as well as from structural muscle-tendon pathology or traumatic injury (6, 8-13). Depending on the underlying cause, weakness may result from impaired neural activation, partial denervation, or intrinsic muscle dysfunction.

When the shoulder elevators, including the deltoid and cranial rotator cuff muscles, are weakened, force production may become insufficient to generate the torque required for arm elevation and external rotation (14). This may lead to imbalanced muscle function around the glenohumeral joint, reduced dynamic stabilization and altered glenohumeral joint reaction forces, which may contribute to progressive joint deformity and secondary structural alterations in affected shoulders (15-18).

Muscle weakness becomes most apparent during active arm elevation in which even moderate reductions in muscle activation can markedly impair elevation against gravity (19-21). As a result, shoulder abduction and forward flexion may be incomplete or not achievable.

Shoulder weakness restricts active range of motion resulting in disabilities in daily activities such as grooming, dressing, and lifting objects, often necessitating compensatory strategies and leading to fatigue, overuse, and pain (22-27).

1.2 Exoskeletons for Shoulder Support

For individuals with chronic weakness, assistive options are largely limited to slings or externally mounted dynamic arm supports (9), mainly targeting shoulder stability. These approaches can be helpful for better performance of daily tasks, but do not address the underlying deficit in shoulder strength and seem to contribute only limited to arm elevation. As such, there remains a clear unmet need for additional, wearable solutions capable of supporting active shoulder function and strength during daily activities (2).

For functional support of shoulder and arm function, an exoskeleton may be a potential solution by addressing functional limitations associated with shoulder weakness by supporting movement and reducing musculoskeletal load (28). However, such support is primarily meaningful in individuals with residual distal arm, wrist, and hand function, as sufficient motor control at these levels is required for functional task execution once shoulder elevation is facilitated.

An exoskeleton is a wearable mechanical system that operates together with the human body to support or unload movement through the application or redistribution of external forces and thereby joint moments (29). Exoskeletons currently range from active systems working with actuators and sensors to passive systems relying solely on mechanical elements such as springs, cables, or linkages, and may be applied for movement assistance, rehabilitation, or strength support in industry (29, 30).

Most existing shoulder exoskeletons have been developed for industrial use, where they assist healthy individuals during physically demanding or repetitive overhead tasks to reduce fatigue and prevent work-related muscle overload (28, 31). Despite the growing interest in shoulder exoskeletons designed for medical and rehabilitation purposes, most of these systems are still in the experimental phase and have mainly been tested on healthy individuals (29, 30).

Several shoulder exoskeletons have been designed for clinical use, such as the Cleverarm, which provides assistance for various shoulder and elbow movements; however, it remains mainly a research prototype with limited practical assessment (32). Other innovative devices like LIFTYA are under development to enhance shoulder stability and movement in patient populations, but these solutions currently lack validation through peer-reviewed clinical studies (33, 34).

1.3 The Asgari Exoskeleton

In the preceding literature review, particular attention was given to exoskeleton support mechanisms aimed at reducing muscular load, especially those relevant to shoulder weakness. Passive mechanical strategies such as gravity compensation have been shown to reduce shoulder muscle activity during arm elevation (35-37).

A relevant example of such a system is the shoulder exoskeleton developed by Asgari and colleagues, which was designed to partially compensate for the gravitational moment acting on the shoulder joint during arm elevation. This exoskeleton is fully wearable and mechanically passive, and it provides assistance during arm elevation in both abduction and forward flexion, which are movement directions that are essential for many activities of daily living, as earlier addressed.

To date, the Asgari exoskeleton concept has been evaluated in an experimental setting involving only healthy participants, where its use was associated with a reduction in shoulder muscle activity during arm elevation tasks. The concept has not yet been investigated in individuals with neurologically induced shoulder weakness. This causes it to remain unclear to what extent this form of gravity-compensating assistance aligns with the residual functional capacity and movement strategies of this population, indicating a clear gap in literature (38-40).

1.4 Research Gap

Despite the promising biomechanical principles underlying the Asgari shoulder exoskeleton, it remains unclear whether its application can contribute to improvements in daily functional performance and quality of life in individuals with shoulder muscle weakness. While various passive and quasi-passive shoulder exoskeletons have been developed and evaluated in healthy individuals or in occupational settings to reduce shoulder muscle activity and fatigue during overhead work, these devices have not been specifically investigated in populations with clinical shoulder muscle weakness or neurological impairment (36, 41). This causes limited knowledge regarding both the biomechanical effects of exoskeleton-assisted shoulder support in the presence of weakness and the extent to which such assistance aligns with the remaining muscular capacity and movement abilities of affected individuals.

In addition, it is currently unknown whether a wearable shoulder exoskeleton would be acceptable and usable in terms of comfort, wearability, and individual preferences, particularly given the presence of sensory disturbances and the considerable variability in functional limitations within various populations. Moreover, previous research has not combined biomechanical analyses of shoulder exoskeleton support with the perspectives and needs of the target populations itself.

At the Leiden NerveCenter, patients with Erb's palsy are routinely seen during childhood and adolescence, however, follow-up typically does not continue into adulthood. Many of these patients exhibit persistent shoulder weakness. Clinical experience at the Leiden NerveCenter suggests that adults with Erb's palsy continue to experience limitations in daily activities, and that unmet needs may exist with respect to biomechanical assistance.

1.5 Goal of This Research

The aim of the present study is therefore to evaluate, from both a biomechanical and a user-centred perspective, whether the Asgari shoulder exoskeleton can serve as a suitable supportive solution for shoulder weakness.

As the central hypothesis of this study, it is expected that the Asgari shoulder exoskeleton provides biomechanically effective support and addresses user-relevant needs for daily activities in individuals with shoulder muscle weakness, with adults with Erb's palsy serving as a representative target group for user-perspective evaluation.

Based on this aim, the following overarching research question was formulated:

Is the Asgari shoulder exoskeleton suitable for daily use in individuals with shoulder muscle weakness in adults with Erb's palsy?

To address this overarching question, the following questions were defined:

1. *Which daily limitations, compensatory strategies, and support needs do adults with Erb's palsy experience with respect to shoulder elevation, and how do they perceive the potential use of a shoulder exoskeleton in everyday life?*
2. *Can passive shoulder assistance by the Asgari exoskeleton compensate for graded shoulder muscle weakness expressed as MRC scores by restoring functional arm elevation capacity and reducing relative neuromuscular activation demand, without introducing adverse compensatory muscle activation patterns?*

2 Background

2.1 Shoulder Biomechanics Relevant to Arm Elevation

Functional arm elevation is achieved through the coordinated interaction between the glenohumeral and scapulothoracic joints, enabling the upper limb to be positioned against gravity during activities of daily living (ADL). Due to the large arm moment of the arm mass acting at the glenohumeral joint, the shoulder muscles are required to generate substantial force to produce sufficient torque to counteract gravitational loading. Arm elevation during both forward flexion and abduction are shown in figure 2.1 and is primarily generated by the deltoid muscle in combination with the cranial parts of the pectoralis and the rotator cuff muscles supraspinatus, infraspinatus and subscapularis. During abduction, the contribution of the middle deltoid is more prominent, whereas the anterior deltoid plays a greater role during forward flexion (21, 42-44).

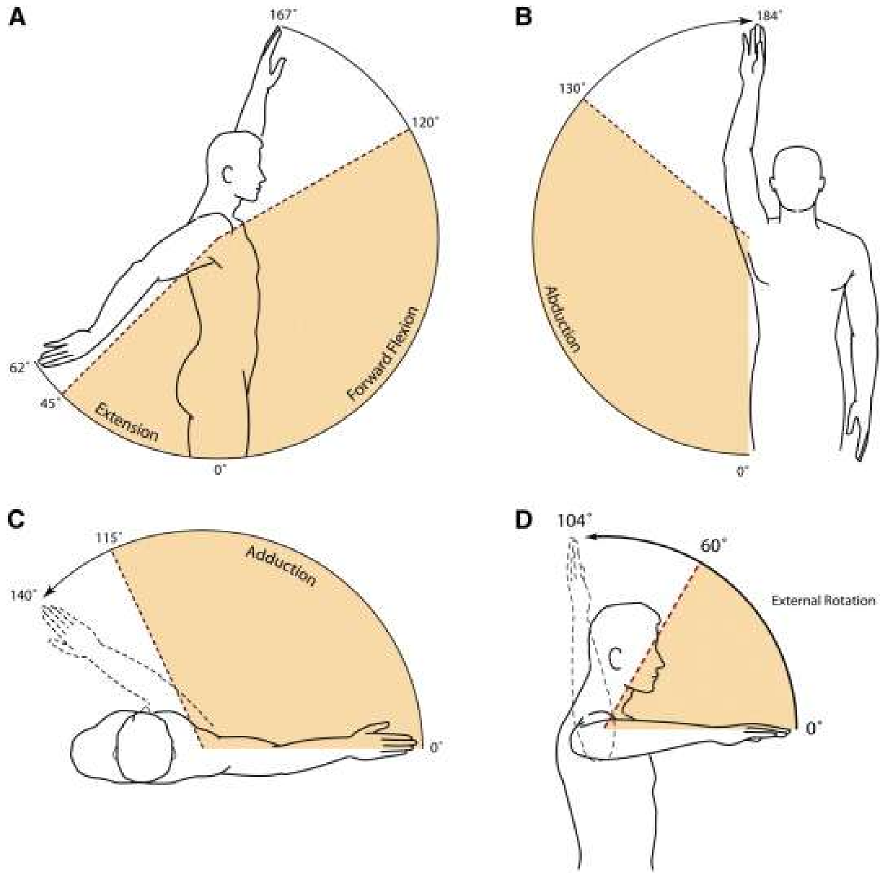

Figure 2.1: The motions of shoulder flexion, abduction, and external rotation (and adduction), with the functional range of motion required for activities of daily living highlighted in orange (45).

As a secondary effect of deltoid activation during arm elevation, upward directed shear forces are produced at the glenohumeral joint. To maintain dynamic joint stability, compensatory activation of the rotator cuff muscles (teres minor, infraspinatus, supraspinatus, subscapularis) is required. Visualisation of shoulder muscles can be found in figure 2.2. The rotator cuff generates compressive forces that centre the humeral head within the glenoid fossa, thereby counteracting the destabilizing shear forces (16, 17, 46-48).

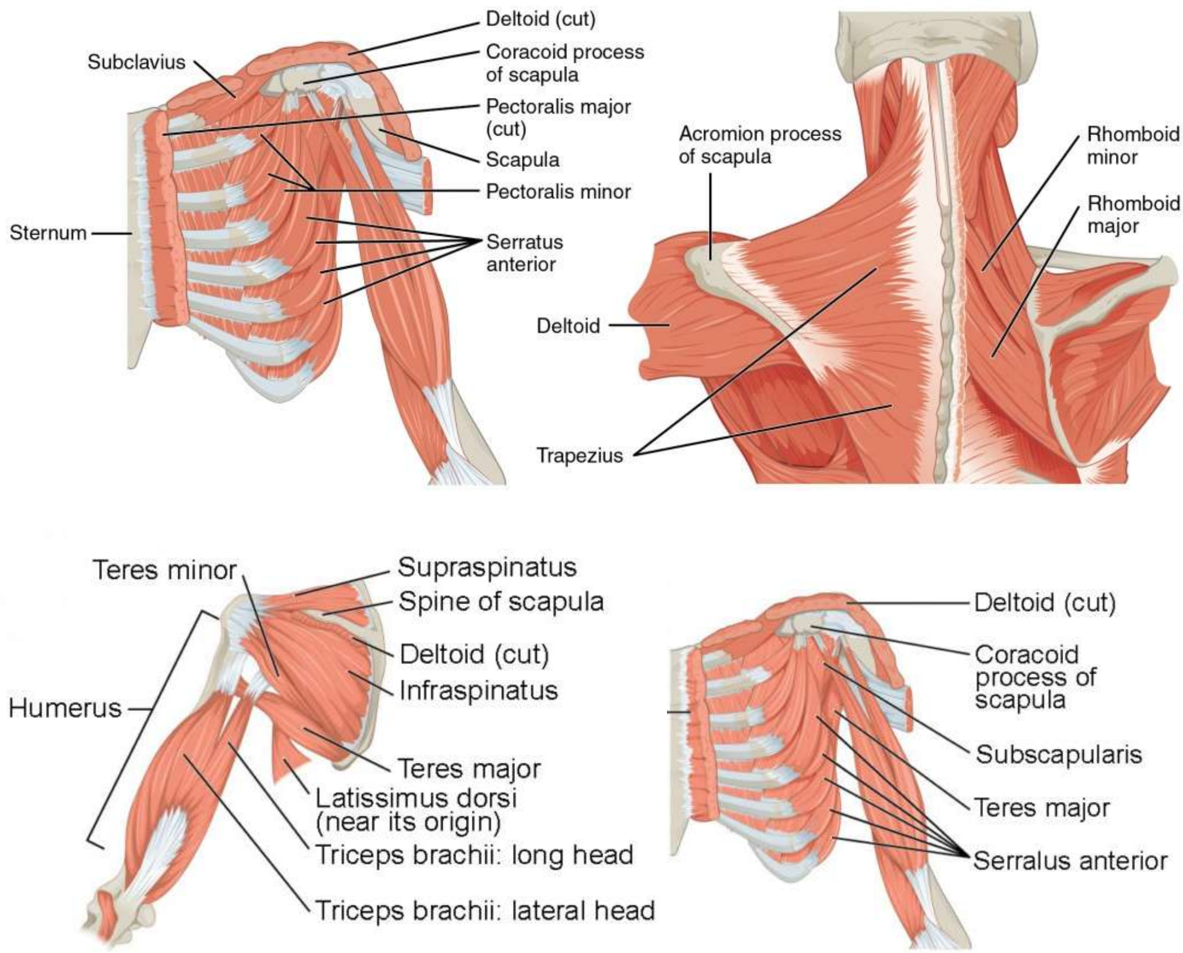

Figure 2.2: Overview of several shoulder muscles shown from multiple perspectives, included to support anatomical orientation and facilitate interpretation of the muscle names referenced in the text. Superficial muscles have been partially removed in some views to visualize deeper structures (49).

With increasing degrees of the arm elevation angle, scapulothoracic motion becomes progressively more important. The scapula upwardly rotates and posteriorly tilts relative to the thorax, allowing continued elevation of the arm. These movements are primarily facilitated by the serratus anterior and trapezius muscles. Through this coordinated action, glenohumeral joint stability and adequate subacromial clearance are preserved. Without sufficient scapular motion, the humeral head may impinge beneath the acromion, potentially leading to subacromial Pain Syndrome (SAPS). Such impingement can restrict active range of motion and result in increased compensatory muscle activation (50-53).

During arm elevation, glenohumeral external rotation occurs concurrently with flexion and abduction and constitutes an essential component of normal shoulder kinematics. Progressive external rotation of the humerus allows the greater tuberosity to clear the undersurface of the acromion, thereby contributing to the maintenance of subacromial space during elevation. This motion is primarily generated by the infraspinatus and teres minor muscles of the rotator cuff and works synergistically with scapular upward rotation to optimize joint stability and minimize the risk of subacromial impingement. Insufficient external rotation during arm elevation may therefore increase mechanical compression within the subacromial region and lead to altered movement strategies and increased muscular compensation (54-56).

2.2 Clinical Characterisation of Shoulder Function

(Residual shoulder) Muscle strength is commonly quantified using the Medical Research Council (MRC) scale, which classifies muscle function according to the ability to generate movement under increasing mechanical demands. These range 0-5: from the absence of observable or palpable movement (MRC 0) to observable or palpable movement without joint movement with gravity eliminated (MRC 1), to movement possible only with gravity eliminated (MRC2), active movement against gravity (MRC3), movement against external resistance (MRC4), and normal upper limb function (MRC5). These categories reflect clinically meaningful functional thresholds rather than proportional or linear differences in force-generating capacity, and are shown in table 1.1 (57).

MRC grade	*Functional description*
0	No visible or palpable muscle contraction
1	Visible or palpable contraction without joint movement
2	Active movement with gravity eliminated
3	Active movement against gravity
4	Active movement against external resistance
5	Normal muscle strength

Table 1.1: Medical Research Council (MRC) Scale for Shoulder Muscle Strength.

To relate muscle strength to functional performance in daily activities, tasks derived from the Mallet classification are frequently employed for brachial plexus injury. Specifically, the hand-to-mouth and hand-to-head tasks evaluate the ability to position the upper limb in functionally relevant postures and inherently capture compensatory strategies involving the scapula, thorax, and elbow. These tasks are visualized in figure 2.3. The score increases with better function, from inability to actively elevate the arm (1), inability to reach the head (2), difficulty during reaching of the head (3), easy reaching of the head (4), or normal function (5). During reaching of the mouth, the score is based on the presence of a visible trumpet sign (2), a partial trumpet sign of >40° (3), or no trumpet sign (>40°) (4), or normal hand-to-mouth. The characteristic elbow abduction associated with the trumpet sign is illustrated in figure 2.3. In the present study, MRC levels were used to define representative degrees of muscle weakness for simulation purposes, while Mallet tasks were used to support the interpretation of functional relevance. It should be noted that Mallet scores may be influenced not only by muscle weakness but also by structural joint limitations, particularly in abduction and external rotation (58).

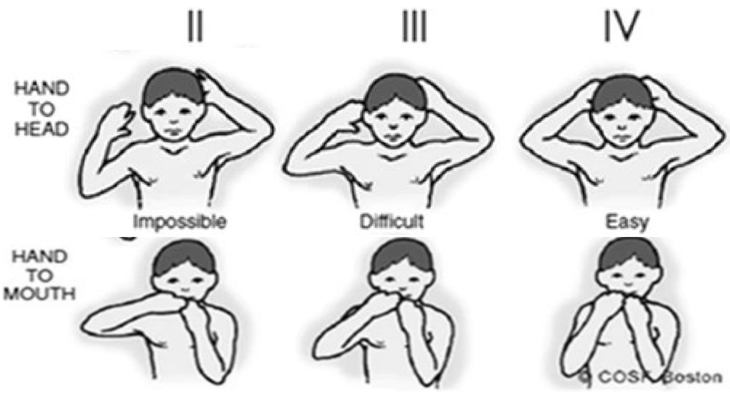

Figure 2.3: Relevant Mallet scores, showing the tasks hand-to-mouth and hand-to-head. The trumpet sign, characterized by compensatory elbow abduction during hand-to-mouth movement, is visible in the lower scoring condition. Adapted from Kruit et al. (2016) (59).

2.3 Musculoskeletal Simulation and OpenSim

Musculoskeletal simulation is a mathematical in silico approach used to analyse how human movement is generated by predicted muscle forces acting on the skeletal system. By combining prescribed joint motion with anatomical and mechanical representations of muscles and joints, internal biomechanical quantities such as joint torque, muscle force and, muscle activations can be estimated.

In musculoskeletal models, the human body is represented as a system of rigid segments connected by joints and actuated by muscles. Joints define the degrees of freedom of the system, while muscles generate forces that act on these joints through their moment arms. This modelling framework provides a mechanical link between observed movement and the forces required to produce it (60).

2.3.1 Joint Kinematics and Inverse Dynamics

Joint kinematics describe movement in terms of joint angles, velocities, and accelerations, without reference to the forces that generate this motion. Kinematic data represent the time-varying configuration of the model and can be obtained from experimental measurements or predefined motion trajectories.

Inverse dynamics uses joint kinematics as input to compute the net joint torques required to reproduce the observed motion. These torques account for gravitational loading, inertial effects, and external forces, and represent the mechanical demand acting at each degree of freedom.

Importantly, inverse dynamics determines only the net torque at each joint, not how that torque is distributed among individual muscles. Estimating muscle-level contributions requires an additional muscle recruitment strategy (61-63).

2.3.2 OpenSim as a Musculoskeletal Simulation Platform

OpenSim is an open-source software platform for musculoskeletal modelling and biomechanical simulation. It provides tools to define musculoskeletal models, prescribe joint kinematics, and compute joint torques and muscle-driven solutions using inverse-dynamics-based methods.

Within OpenSim, movement is described using generalized coordinates, each representing a single degree of freedom (DoF) such as a joint rotation or translation. Muscles are modelled as force-generating actuators spanning one or more joints.

In the commonly used Hill-type muscle model implemented in OpenSim, each muscle-tendon unit consists of:

- A contractile element representing active force generation,
- A series elastic element representing tendon elasticity,
- A parallel elastic element representing passive muscle tissue stiffness.

The contractile element produces active force in response to neural activation. The series elastic element models tendon compliance, transmitting force to the skeleton. The parallel elastic element represents passive resistance arising from connective tissue when the muscle is stretched. As a result, total muscle-tendon force consists of both active and passive components. At the joint level, passive stiffness arises from the combined effects of muscle-tendon elasticity and, when modelled, additional passive joint structures such as capsular and ligamentous constraints (60, 64).

This structure allows the model to account for passive elastic resistance (parallel elasticity) and tendon compliance (series elasticity), meaning that muscle force is influenced not only by activation but also by muscle length and contraction velocity. So OpenSim captures key mechanical characteristics of muscle-tendon behaviour within a dynamically consistent framework (60, 64-66).

2.3.3 Neural and Muscle Activation, Force, and Joint Torque

Muscle force generation originates from neural excitation. In physiological terms, neural input triggers calcium influx in muscle fibres, enabling cross-bridge formation and active force production. In musculoskeletal models, this process is represented in simplified form through the state variable muscle activation, a dimensionless quantity between zero and one that reflects neural drive to the muscle.

Muscle activation determines the fraction of the muscle's maximal force-generating capacity that is recruited at a given time. Active force production depends on activation, muscle fibre length (force-length relationship), contraction velocity (force-velocity relationship), and maximal isometric strength. In addition, passive elastic forces arise when the muscle-tendon unit is stretched.

In simplified form, muscle force can be approximated as:

$$F(t) \approx \alpha(t) \cdot F_{max} \tag{1}$$

where $\alpha(t)$ represents muscle activation (0-1) and F_{max} the maximal isometric force.

Maximal isometric force depends on the physiological cross-sectional area (PCSA) and maximal muscle stress (σ_{max}):

$$F_{max} = PCSA \cdot \sigma_{max} \tag{2}$$

In Hill-type models, force is additionally modulated by muscle fiber length and contraction velocity, and includes passive elastic contributions.

The resulting muscle-tendon force is transmitted to the skeleton via the tendon and produces a joint torque:

$$\tau = r \cdot F \tag{3}$$

Where F is the muscle force and r is the corresponding arm moment.

Muscle activation, muscle force, and joint torque represent different mechanical levels of the system and are therefore related but not interchangeable. High activation does not necessarily imply high force when maximal strength (F_{max}) is reduced, and a given joint torque can be achieved through multiple combinations of muscle forces due to redundancy in the musculoskeletal system (60, 67-69).

2.3.4 Coordinates, Reserve Actuators, and Residuals

Movement in musculoskeletal models is expressed using generalized coordinates that define the configuration of the system at each time point. For each coordinate, the equations of motion describe how joint torques generated by muscles, external forces, and inertial effects contribute to joint acceleration.

When muscle-generated torque is insufficient to meet the required joint moment, reserve actuators may apply direct torques to individual coordinates. These actuators do not represent physiological mechanisms and are therefore penalized in muscle-driven analyses. Small reserve contributions are commonly accepted as numerical compensation, whereas substantial reserve actuator torques indicate that the modelled musculature alone is insufficient to reproduce the prescribed motion. In line with commonly applied OpenSim guidelines, reserve contributions exceeding approximately 10% of the required joint torque were interpreted as indicative of non-physiological assistance.

Residual actuators apply forces and moments to a reference segment of the model to correct for global dynamic inconsistencies, such as modelling errors in segment inertial properties or external forces. Their purpose is to ensure overall dynamic consistency of the system rather than to represent joint-level actuation, and they are conceptually distinct from coordinate-level reserve actuators (68, 70, 71).

2.4 Working Principle of the Rapid Muscle Redundancy solver

In musculoskeletal models, the number of muscles exceeds the number of mechanical degrees of freedom, meaning that multiple muscle activation patterns can generate the same joint torques. This indeterminacy is known as the muscle redundancy problem. To obtain an unique and physiologically logical activation pattern, an optimization-based recruitment strategy is required.

In this study, muscle recruitment was resolved using the Rapid Muscle Redundancy (RMR) solver. RMR is formulated within an inverse-dynamics framework. Joint kinematics (positions, velocities and accelerations) are prescribed, and the corresponding net joint torques required to reproduce the motion are computed. At each discrete time step k, the solver determines a set of muscle activations that reproduces the required accelerations while satisfying physiological and mechanical constraints. An conceptual overview of the RMR pipeline is shown in figure 2.4.

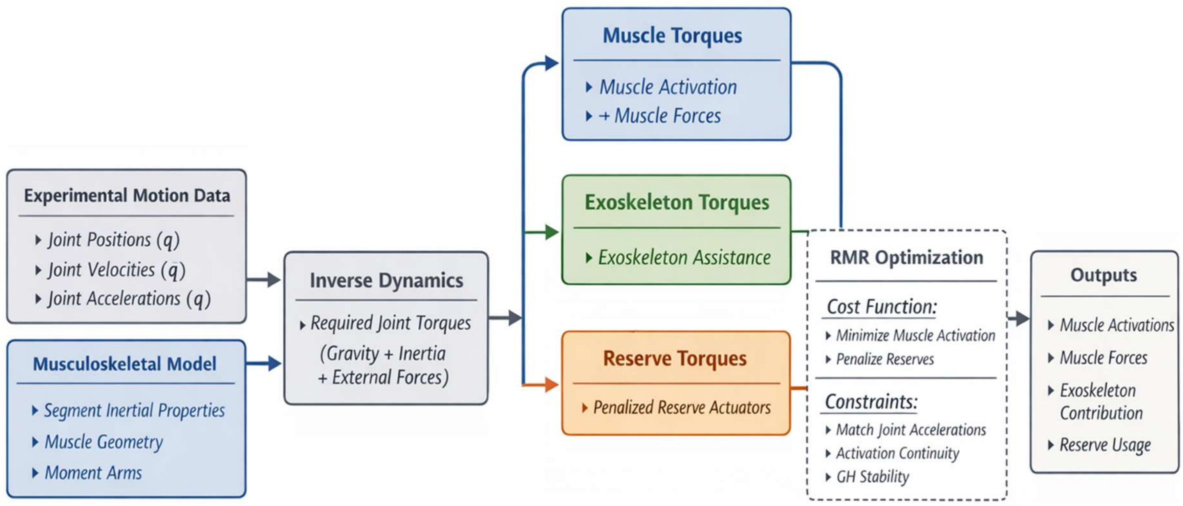

Figure 2.4: Conceptual overview of the inverse-dynamics-based muscle redundancy framework used in this study. Kinematic motion data and a musculoskeletal model are used to compute the required joint torques through inverse dynamics. These torques are subsequently distributed over muscle activations, prescribed exoskeleton assistance when applicable, and penalized reserve actuators by means of a Rapid Muscle Redundancy (RMR) optimization, subject to physiological and stability constraints. This schematic was based on the OpenSim inverse analysis workflow and the Rapid Muscle Redundancy formulation described by Belli et al. (2023) (72, 73).

2.4.1 Optimization Objective

At every time step, RMR solves a constrained minimisation problem of the form:

$$J(a_k, c_k) = \sum_{i=1}^{N_m} w_i\, a_{i,k}^2 + \sum_{j=1}^{N_q} v_j\, c_{j,k}^2 \tag{4}$$

Where:

- J_k Is the total cost of the solution at time step k
- $a_{i,k} \in [0.1]$ is the activation of muscle i at time step k
- N_m = 33 is the total number of muscles included in the model
- $c_{j,k} \in [0.1]$ is the control input of reserve actuator j
- N_q = 9 is the number of unlocked generalized coordinates
- w_i and v_j are weighting coefficients (1 and 50, respectively)

The first summation penalizes muscle activation. Because activations are squared, higher activation levels are disproportionately penalized, encouraging distributed and moderate recruitment patterns. The second summation penalizes the use of reserve actuators. To favor physiologically realistic muscle-driven solutions, the weighting factors v_j were chosen substantially larger than muscle weights (v_j = 50). In the present formulation, equal weighting factors were applied to all muscles (w_i= 1), meaning that the optimization penalizes neural activation equally across muscles regardless of muscle size. Consequently, small and large muscles are treated equivalently in terms of activation cost. However, differences in maximal force capacity (F_{max}) and moment arm ensure that larger muscles contribute more strongly to joint torque for a given activation level. The objective function therefore minimizes neural effort rather than muscle force or stress.

Importantly, this objective function minimizes muscle activation and serves as a model-based proxy for neuromuscular activation demand, not muscle force or metabolic energy.

2.4.2 Dynamic Consistency Constraint

Minimization of the objective function is subject to mechanical consistency. The activations must reproduce the prescribed accelerations:

$$A_{acc,k} \left[\frac{a_k}{c_k}\right] = \ddot{q}_k \tag{5}$$

Where:

- a_k is the vector of all actuator activations
- $A_{acc,k}$ is how each actuator contributes to generalized accelerations
- $\ddot{q}_k$ is the accelerations that must be generated by active actuators

The required accelerations are derived from the full multibody equations of motion of the system. These equations incorporate inertial effects, velocity-dependent terms (e.g., Coriolis and centrifugal effects, which arise from motion-dependent coupling between rotating body segments), gravitational loading, external forces, and passive muscle-tendon contributions arising from length- and velocity-dependent properties. This formulation ensures that the muscle recruitment solution remains dynamically consistent with the imposed motion.

Unlike classical static optimization, which typically matches joint torques, the RMR formulation matches accelerations while explicitly accounting for passive muscle behavior.

2.4.3 Activation Dynamics

Muscle activation cannot change instantaneously. So, upper and lower bounds on activation are imposed at each time step based on activation and deactivation time constants. These bounds restrict how quickly activations can increase or decrease between consecutive frames. As a result, recruitment patterns evolve smoothly over time and remain physiologically plausible.

2.4.4 Glenohumeral Stability Constraint

For the shoulder simulations, glenohumeral stability was enforced through an inequality constraint on the direction of the joint reaction force (JRF). The total joint reaction force at time step.

$$F_k = A_{F,k} \begin{bmatrix} a_k \\ \hline c_k \end{bmatrix} + F_{0,k} \tag{6}$$

Where:

- F_k is the glenohumeral JRF vector
- $A_{F,k}$ is a matrix describing how unit activation contributes to the joint reaction force
- a_k and c_k are the vectors of muscle and reserve activations at time step k
- $F_{0,k}$ is the passive contribution to the joint reaction force resulting from gravity, external loads, and passive muscle forces

This formulation shows that the direction and magnitude of the JRF depend directly on actuator activations.

To ensure that the reaction force remains oriented toward the glenoid surface, the angle between the JRF and a predefined central axis of the glenoid was constrained according to:

$$\left(\frac{\theta_k(a_k, c_k)}{\theta_{max}}\right)^2 - 1 \leq 0 \tag{7}$$

where $\theta_k(a_k, c_k)$ is the angle of the JRF relative to the glenoid center line, and θ_{max} defines the maximum allowable deviation, as can be seen in figure 2.5. This inequality ensures that the JRF vector remains within a predefined stability cone. Because the JRF depends on muscle activations, the constraint may induce additional activation of stabilizing musculature when required.

2.4.5 Incorporation of External Assistance

When assistive torques generated by the modelled exoskeleton are applied, these are included in the joint moment balance prior to optimization and are treated as externally applied joint torques within the equations of motion. Consequently, the net torque that must be generated by the musculature is reduced. Muscle activation patterns are then recalculated under the same objective function and constraints, allowing redistribution of neuromuscular activation demand in response to assistance. (60, 73-76).

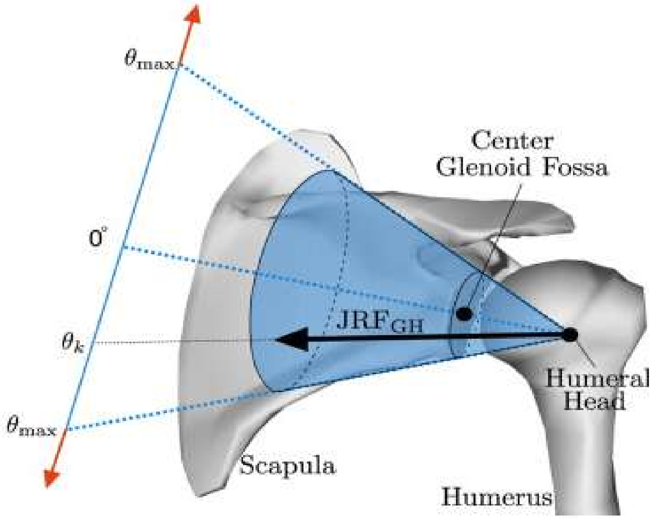

Figure 2.5: Schematic representation of the glenohumeral joint reaction force (JRF) constraint used in the RMR solver. The direction of the JRF is constrained such that its angle θ_k with respect to the centre line does not exceed a maximum value θ_{max}, chosen to keep the force oriented towards the glenoid surface. Adapted from Belli et al. (2023) (73).

2.5 The Asgari Passive Shoulder Exoskeleton

The shoulder exoskeleton developed by Asgari and colleagues is a wearable, mechanically passive system designed to assist arm elevation by partially compensating for the gravitational load acting on the shoulder joint. The device was developed as a prototype intended to support individuals with shoulder disability by partially compensating for the gravitational load acting on the shoulder joint. The device operates without motors or electronic control and generates assistance through purely mechanical means.

The exoskeleton is intended to function in parallel with the human musculoskeletal system. Rather than imposing predefined joint trajectories or constraining movement, it applies an external assistive moment that reduces the mechanical demand placed on the shoulder during arm elevation (38-40).

2.5.1 Mechanical Principle

During arm elevation, gravity generates a moment about the glenohumeral joint that increases nonlinearly with elevation angle. The exoskeleton counteracts this effect by producing an external moment in the direction of shoulder elevation. Assistance is designed to scale with arm elevation, providing limited support at low angles and increasing support at higher elevation angles.

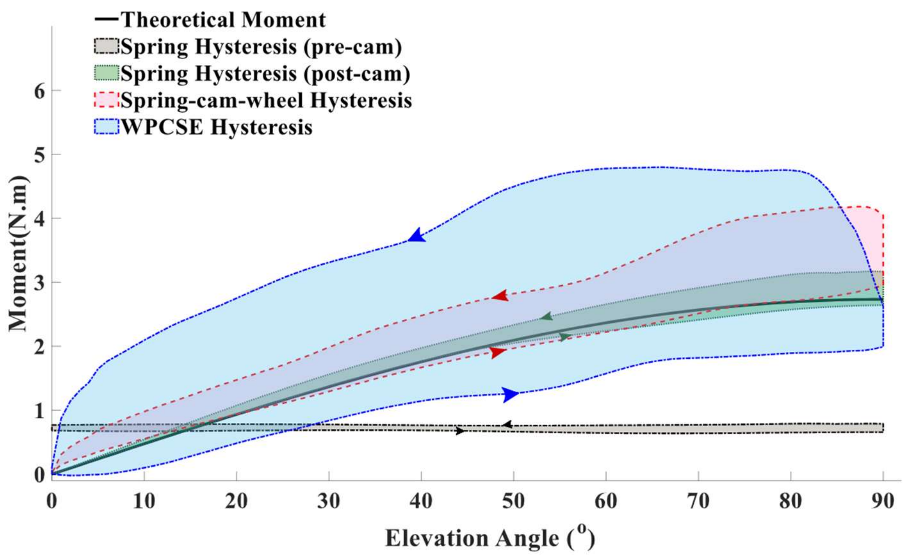

Figure 2.6: Theoretical gravitational shoulder moment and assistive moment generated by the passive spring-cam mechanism as a function of humerothoracic elevation angle. The cam-modulated system approximates the nonlinear gravitational profile and exhibits hysteresis between loading and unloading phases. From Asgari et al. (39, 40)

The assistive moment is generated using a passive spring-cam mechanism, as can be seen in figure 2.6. In this system, a spring provides a constant mechanical input, while a cam profile modulates the effective moment arm as a function of shoulder elevation. This configuration allows the exoskeleton to approximate the nonlinear profile of the gravitational shoulder moment using purely mechanical means.

The system has primarily been configured to compensate for a partial proportion of the gravitational shoulder moment, typically in the range of approximately 20-30%, rather than fully balancing the arm against gravity (38-40).

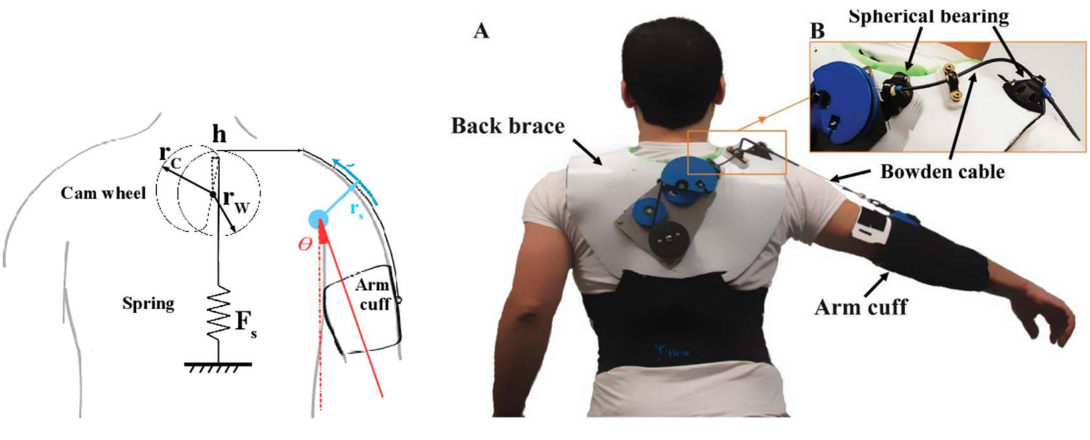

Figure 2.7: Posterior views on the diagram (left), and the experimental shoulder exoskeleton configuration (right); illustrating the mechanism, the cable routing and geometric relationships used in the model. In the figure, r_c denotes the effective cam radius (variable moment arm), r_w the radius of the constant wheel, h the geometric offset between the shoulder joint center and the cam-wheel module, and F_s the spring force transmitted through the cable. On the right, r represents the moment arm of the Bowden cable about the glenohumeral joint and θ the shoulder elevation angle. From Asgari et al. (2023) (39, 40).

2.5.2 Force Transmission and Assisted Movements

Assistive forces generated by the spring-cam mechanism are transmitted to the arm through a cable-driven system. The cable passes over the superior aspect of the shoulder and is attached distally to an arm cuff, as can be seen in figure 2.7. When tensioned, the cable applies an upward force to the arm, producing an external shoulder elevation moment about the glenohumeral joint.

At the superior aspect of the shoulder, the cable is guided through a self-aligning spherical bearing. This bearing allows the cable to rotate freely and align with the direction of arm movement, reducing friction and preventing the transmission of undesired transverse forces to the shoulder. As a result, force application remains directed along the cable regardless of the plane of arm elevation.

The cable runs within a flexible outer sheath (Bowden cable housing) that supports tensile force transmission while allowing relative motion between the cable and the body. The outer sheath enables the routing of assistive forces from the back-mounted mechanical components to the arm without rigid mechanical linkage across the shoulder joint.

This cable-based transmission allows the exoskeleton to apply assistance without constraining shoulder motion to predefined axes. Shoulder elevation can therefore occur across different planes.

The exoskeleton assists arm elevation primarily during abduction and forward flexion. In experimental evaluations reported to date, these movements have been tested within an humerothoracic elevation range of approximately 0-90° (38-40).

2.5.3 Biomechanical Context and Prior Evaluation

By partially compensating for the gravitational shoulder moment, the exoskeleton reduces the net torque that must be generated by the shoulder musculature during arm elevation. The device does not initiate movement independently and does not replace active muscle function.

Existing evaluations of the Asgari shoulder exoskeleton have been conducted in controlled laboratory settings using healthy adult participants. These studies focused on characterising the mechanical behaviour of the system and its general biomechanical interaction with the shoulder during arm elevation within the tested range of motion. Specifically, these evaluations quantified the exoskeleton-generated shoulder elevation moment relative to the theoretical gravitational moment, and assessed its effects on shoulder muscle activation and kinematics (38-40).

3 Methods

3.1 Study Design

To be able to answer the main research question, the study was divided into two methodological different domains:

1. user needs and design requirements for shoulder Asgari exoskeleton support, and
2. the biomechanical feasibility and potential effects of Asgari exoskeleton assistance in the target population.

Accordingly, this thesis was designed as an exploratory mixed-methods feasibility study, combining qualitative and simulation-based findings.

The first question was investigated through semi-structured interviews with adults with shoulder muscle weakness due to Erb's palsy, aiming to identify experienced limitations in daily life activities, used compensatory strategies, and requirements for acceptable and meaningful exoskeleton support. These findings were used to describe functional context and usage conditions relevant for future device design and evaluation, and to create a wishing and requirements inventory for a future exoskeleton design.

The second question was investigated using musculoskeletal simulations in OpenSim to explore the biomechanical contribution of the Asgari shoulder exoskeleton concept during multiple arm elevation motions. Because patient-specific biomechanical datasets were not available and recruiting and testing participants was not possible within this project time, a model-based approach was adopted to estimate potential effects of exoskeleton assistance for shoulder muscle weakness.

3.2 Qualitative Interview Study on Daily Activity Limitations and Support Needs

3.2.1 Study Design and Methodological Approach

The qualitative study employed a semi-structured interview design to explore daily activity limitations, used compensatory strategies, and perceived needs for support in adults with Erb's palsy. A qualitative approach was chosen to capture experiential perspectives and contextual factors.

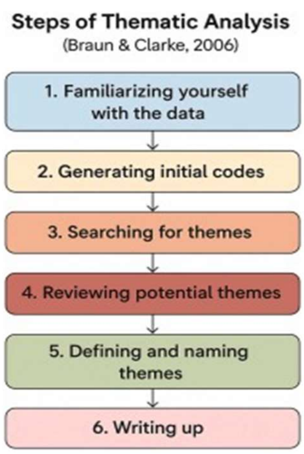

Figure 3.1: Schematic overview of the thematic analysis workflow, depicting the iterative phases of familiarization, coding, theme development, review, and reporting. From Braun and Clarke (77).

The data were analysed using reflexive thematic analysis, following the framework of Braun and Clarke as can be seen in figure 3.1 (77). Analysis followed a predominantly inductive approach, while being informed by concepts derived from the study aims. This approach was selected because it allows for systematic identification of shared patterns of meaning across participants, while acknowledging the active interpretative role of the researcher. Reporting was checked for the guidelines of the Consolidated Criteria for Reporting Qualitative Research (COREQ) (78).

3.2.2 Participants and Recruitment

Adults with Erb's palsy affecting one or both upper extremities were eligible for inclusion. Participants were required to have residual active shoulder function, defined as the ability to actively (partly) perform shoulder abduction and/or forward flexion.

Participants were recruited via two routes:

1. an open online recruitment message distributed through the national Erb's palsy patient association, and
2. retrospective identification of patients visiting the outpatient clinic of the Nerve Center at Leiden University Medical Center (LUMC), with the support of with an involved rehabilitation physician.

Interested individuals received written study information and provided informed consent prior to participation. Interviews were conducted online via Microsoft Teams.

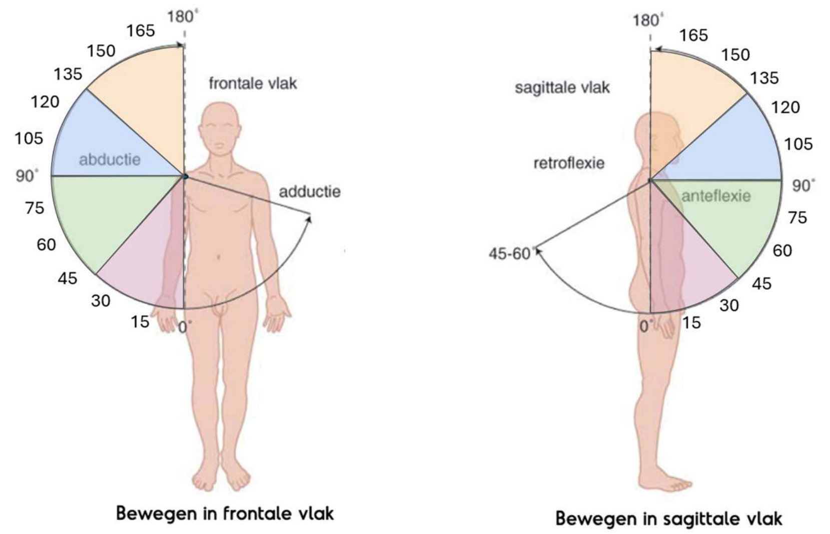

Figure 3.2: Division of shoulder flexion and abduction angles (0°-180°) into four color-coded quadrants used to classify participants based on maximum active range of motion (79).

3.2.3 Data Collection

Each interview consisted of two components: identifying participant characterization, and a semi-structured qualitative interview (appendix A).

Participant characterization included demographic information, affected side(s), known neurological level of injury, prior use of aiding devices, and surgical history. Functional shoulder performance was assessed during the interview using quartiles for determining the active range of motion for shoulder abduction and forward flexion (see figure 3.2), and two items from the Mallet classification relevant for daily activities (hand-to-mouth and hand-to-head tasks). More information about the Mallet classification and MRC scale can be found in Background 2.2. These measures were used descriptively to contextualize qualitative findings and were not included in the thematic analysis.

During the semi-structured interviews, participants were asked multiple questions about experienced limitations in activities of daily living, used compensatory strategies, fatigue, frustration and pain, and priorities for functional improvement. Participants were subsequently introduced to the concept of the Asgari shoulder exoskeleton support using verbal explanations and visual materials of the Asgari shoulder exoskeleton design. More information about the exoskeleton can be found in Background 2.5. Participants were asked to discuss perceived usefulness, potential application contexts, and requirements related to comfort, usability, and support characteristics.

The full interview questionnaire, explanatory text, and images are provided in Appendix A. Interviews were recorded and auto-transcribed via Microsoft Teams function. The transcripts were subsequently reviewed and manually checked against the recordings, and corrected where necessary to ensure accuracy. The mean interview duration was 34 minutes (range: 23-47 minutes).

3.2.4 Qualitative Data Analysis

Qualitative data were analysed using reflexive thematic analysis with the use of ATLAS.ti software using a LUMC-license The methodological steps followed can be found in figure 3.3.

First, all transcripts were globally coded, assigning descriptive labels to segments of text capturing relevant data. These initial codes were then reviewed and merged into a reduced set of codes based on conceptual similarity and analytical relevance.

In a second step, codes were grouped into higher-order categories. These categories were then further integrated into a limited amount of overarching themes that captured shared patterns across participants.

For identification of patterns between the themes, a co-occurrence analysis was performed within ATLAS.ti to examine frequently co-occurring codes. Patterns of co-occurrence were used as an aid to explore how themes were related to each other.

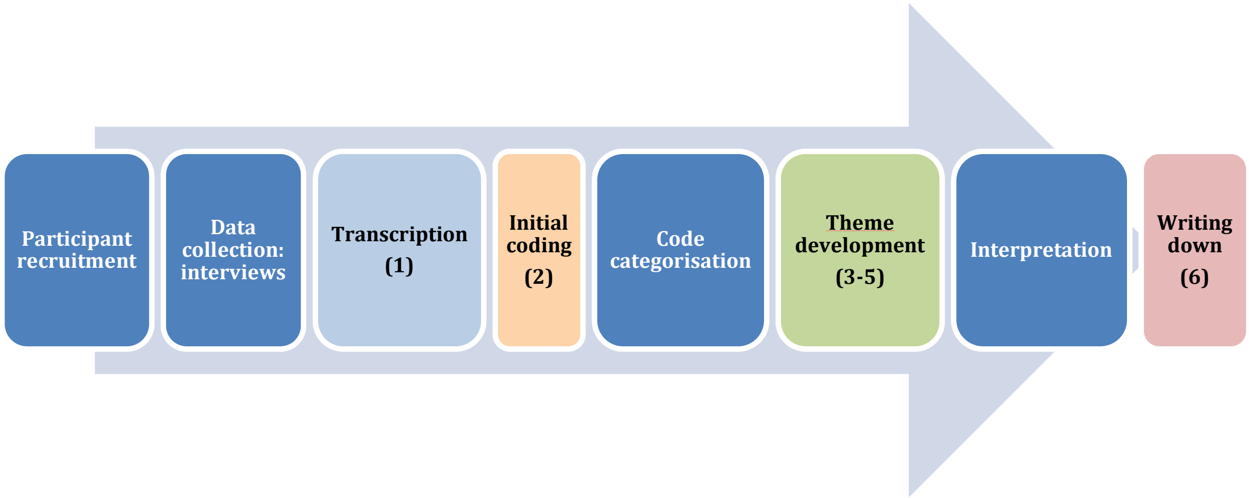

Figure 3.3: Overview of the qualitative research process, from participant recruitment to theme development and interpretation. The color-coded phases correspond to the stages of reflexive thematic analysis as described by Braun and Clarke (77).

3.2.5 Methodological Rigor and Reflexivity

All interviews and qualitative analyses were conducted by a single researcher (CK). Reflexivity was maintained throughout the process through iterative reading of the data. In line with reflexive thematic analysis, inter-coder reliability was not pursued, as themes were developed through interpretative analysis. Analytic rigor was supported by the use of a transparent and systematic progression from codes to categories and followed overarching themes.

3.2.6 Ethical Considerations and Data Protection

This study was conducted as a feasibility study and in consultation with the Rehabilitation department of the LUMC not subject to the Dutch Medical Research Involving Human Subjects Act (WMO), nor formal ethical approval by a (Medical) Ethics Review Committee was asked. No interventions were performed and participation involved minimal risk.

All participants provided informed consent prior to participation and recording by written mail and/or complying with the interview after notifying this as a requirement for participation. Interviews were audio-recorded and automatically transcribed using Microsoft Teams transcription functionality, after which transcripts were reviewed and adjusted for accuracy using the recordings for eligibility. Transcripts were anonymized prior to analysis. The key linking participant identities to anonymized codes was stored separately form the research data in a secure, access-restricted location to ensure confidentiality.

Quoted material is presented using anonymized participant codes to protect confidentiality, characterized by participant number, affected side and neurological level of injury, for example: (P3, left, C5-6).

3.3 Biomechanical Simulation Study of Asgari Shoulder Exoskeleton Assistance

The biomechanical component of this study was designed to evaluate whether passive gravity-compensating shoulder assistance can address two primary biomechanical unmet needs in individuals with shoulder muscle weakness:

1. Limited active arm elevation capacity against gravity, and
2. Increased neuromuscular activation demand of the remaining shoulder musculature during elevation.

In the present study, neuromuscular activation demand is operationalized as the relative level of muscle activation required to generate the necessary joint moments. Within Hill-type musculoskeletal models, muscle activation represents normalized neural excitation (ranging from 0 to 1) driving force production (60, 64, 67). It should be noted that muscle activation in the model does not directly represent metabolic cost or fatigue, but provides a relative measure of neural drive within the simulated framework.

This simulation study operationalizes the central hypothesis that passive assistance provided by the Asgari shoulder exoskeleton can biomechanically compensate for reduced shoulder muscle strength while maintaining physiologically logical muscle activation patterns.

Muscle weakness was modelled as a reduction in maximal muscle strength (F_{max}) of selected shoulder elevators affected by Erb's palsy. This reduction represents the underlying impairment, while the functional consequences were evaluated in terms of:

- Muscle-driven elevation feasibility in terms of active Range of Motion (aRoM), and
- Neuromuscular activation demand, quantified using muscle activation metrics.

The simulations therefore aimed to determine whether passive assistance can:

- Restore muscle-driven elevation capacity under graded weakness, and
- Reduce peak (A_{max}) and cumulative muscle activation (A_{cum}), as a proxy for neuromuscular activation demand, without inducing more than a twofold increase in activation of other muscles or reliance on non-physiological actuators.

The overall simulation workflow is summarized in figure 3.4.

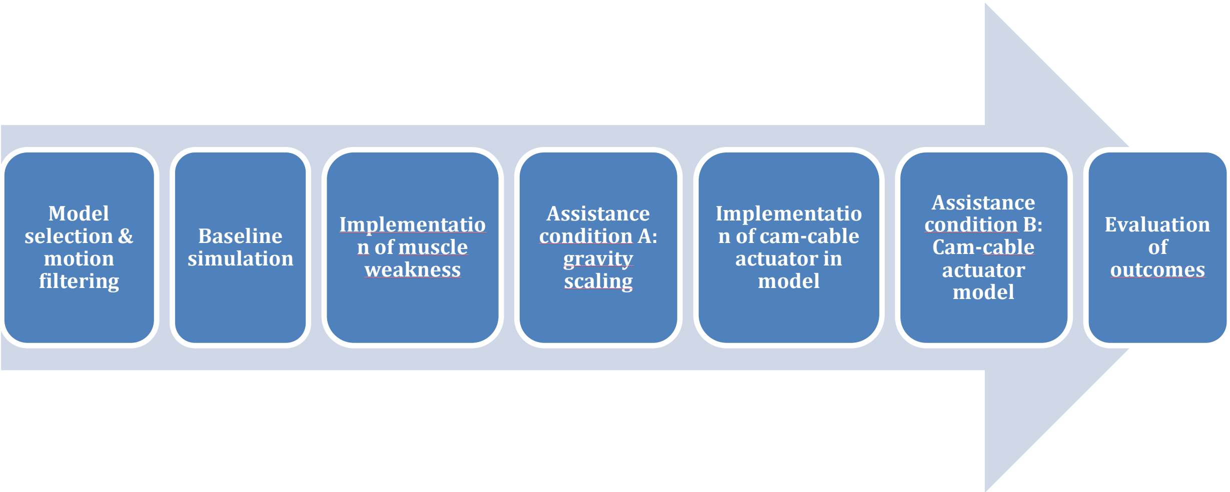

Figure 3.4: Overview of the biomechanical simulation workflow.

3.3.1 Musculoskeletal Model and Simulation Environment

Simulations were performed using a validated three-dimensional upper extremity musculoskeletal model implemented in OpenSim (80). The model represents the shoulder and arm using anatomically defined joints and musculotendon parts. The musculoskeletal model and associated configuration files were obtained from previously published OpenSim models, and via the Computational Biomechanics Lab of the Delft University of Technology and is illustrated in figure 3.5 (81).

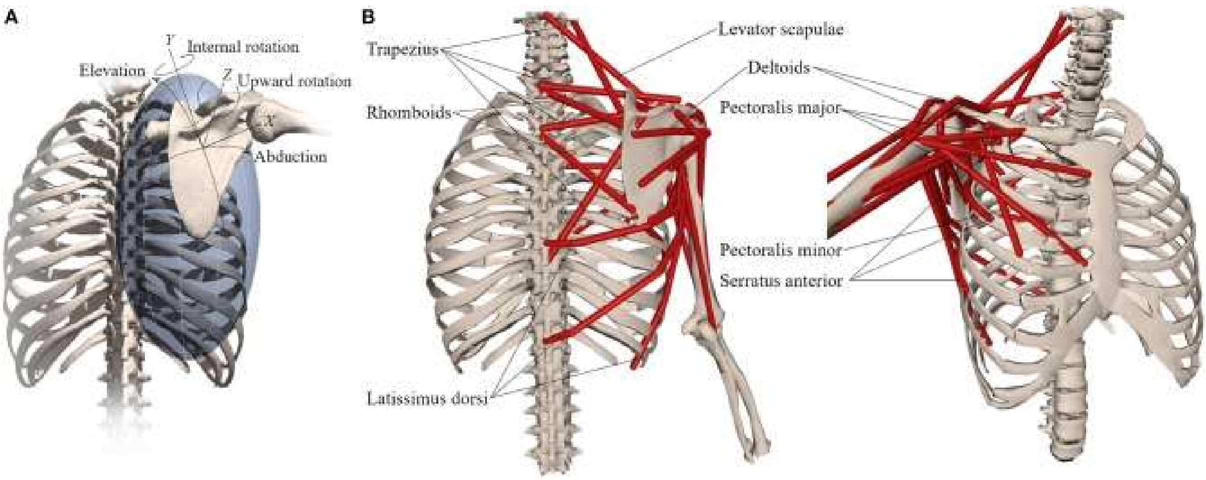

Figure 3.5: Thoracoscapular musculoskeletal shoulder model used in this study, originally developed by Seth et al. (2019), illustrating the scapulothoracic and glenohumeral degrees of freedom (A) and the included musculotendon actuators with selected shoulder and scapular muscles labelled (B)(81).

The following model and simulation files were used:

- The OpenSim musculoskeletal model file, including the corresponding geometry files required for visualization and muscle path definitions.
- Prescribed motion files for shoulder abduction (Abduction2) and forward flexion (Flexion1), containing the experimentally derived joint kinematics.
- The associated marker location files corresponding to each prescribed motion.
- Python-based Rapid Muscle Redundancy (RMR) solver scripts used to compute muscle activation patterns under prescribed kinematics.

All simulations were executed using these predefined and previously validated model files. No modifications were made to joint geometry or muscle-tendon parameters.

3.3.1.1 Joint Kinematics and Degrees of Freedom

The underlying musculoskeletal model comprised 17 generalized coordinates in total. Of these, six ground-thorax coordinates were retained as residual degrees of freedom for numerical stabilization, two distal coordinates were constrained, and nine shoulder-related generalized coordinates were actively prescribed in the simulations (Table 3.1).

The nine shoulder-related degrees of freedom required for coordinated arm elevation were prescribed using experimentally derived kinematic trajectories. These degrees of freedom were implemented as generalized coordinates in the OpenSim model. These included clavicular, scapular, and glenohumeral degrees of freedom necessary to reproduce physiological elevation kinematics.

Coordinate	*Coordinate*	*Role in simulation*
Sternoclavicular	Clav_prot	Active elevation kinematics
	Clav_elev	Active elevation kinematics
Scapulothoracic	Scapula_abduction	Active elevation kinematics
	Scapula_elevation	Active elevation kinematics
	Scapula_upward_rot	Active elevation kinematics
	Scapula_winging	Active elevation kinematics
Glenohumeral	Plane_elv	Active elevation kinematics
	Shoulder_elv	Primary elevation angle
	Axial_rot	Axial coordination
Elbow	Elbow_flexion	Constrained
Radioulnar	Pro_sup	Constrained
Ground-thorax	Ground_thorax DOFs (6)	Numerical stabilization

Table 3.1: Overview of generalized coordinates included in the simulation and their functional role. From Seth et al. (2019) (81).

Elbow flexion-extension and forearm pronation-supination were constrained by fixing their joint angles at constant values and excluding them from optimization Constraining these coordinates did not remove the corresponding joints from the model. The joints remained structurally present but were not allowed to vary during simulation. This prevented artificial compensatory strategies involving distal joint motion while preserving the biomechanical integrity of the shoulder-arm chain.

Ground-thorax coordinates were retained as residual degrees of freedom to ensure dynamic consistency and numerical robustness but were not prescribed as active movement variables.

Constraining non-essential degrees of freedom reduces the dimensionality of the optimization problem and improves numerical stability by limiting redundancy in the inverse-dynamics and muscle recruitment solution. This reduces the risk of ill-conditioned optimization problems while maintaining the essential kinematic structure of shoulder elevation.

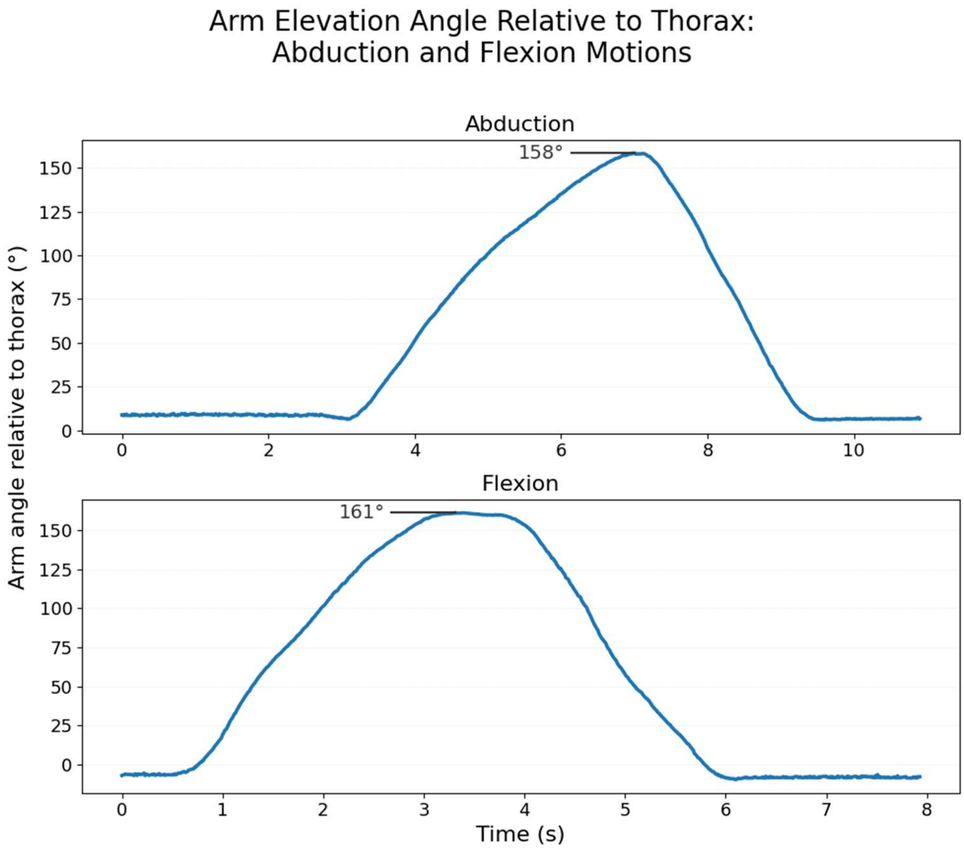

Figure 3.6 : Prescribed humerothoracic elevation angle relative to the thorax during abduction and flexion. Peak elevation angles of 158° (abduction) and 161° (forward flexion) were used as input kinematics for all simulation conditions (81).

Shoulder abduction and forward flexion were simulated as representative elevation movements relevant for activities of daily living. Identical kinematics were prescribed across all conditions to allow direct biomechanical comparison. The kinematic trajectories of thorax, clavicle, scapula and humerus were obtained by Seth et al. (2019) from experimental motion capture using reflective skin-mounted markers placed on anatomical landmarks of the upper extremity (81). The humerothoracic angle of both movements over time is shown in figure 3.6.

This modelling approach is based on the assumption that, when sufficient mechanical support is provided, individuals with shoulder muscle weakness are biomechanically capable of approximating non-compensatory, healthy elevation kinematics. Under this assumption, the simulations evaluate whether the available musculature, potentially supported by passive assistance, is mechanically sufficient to reproduce healthy elevation patterns against gravity.

All movements were simulated under quasi-static using the RMR solver.

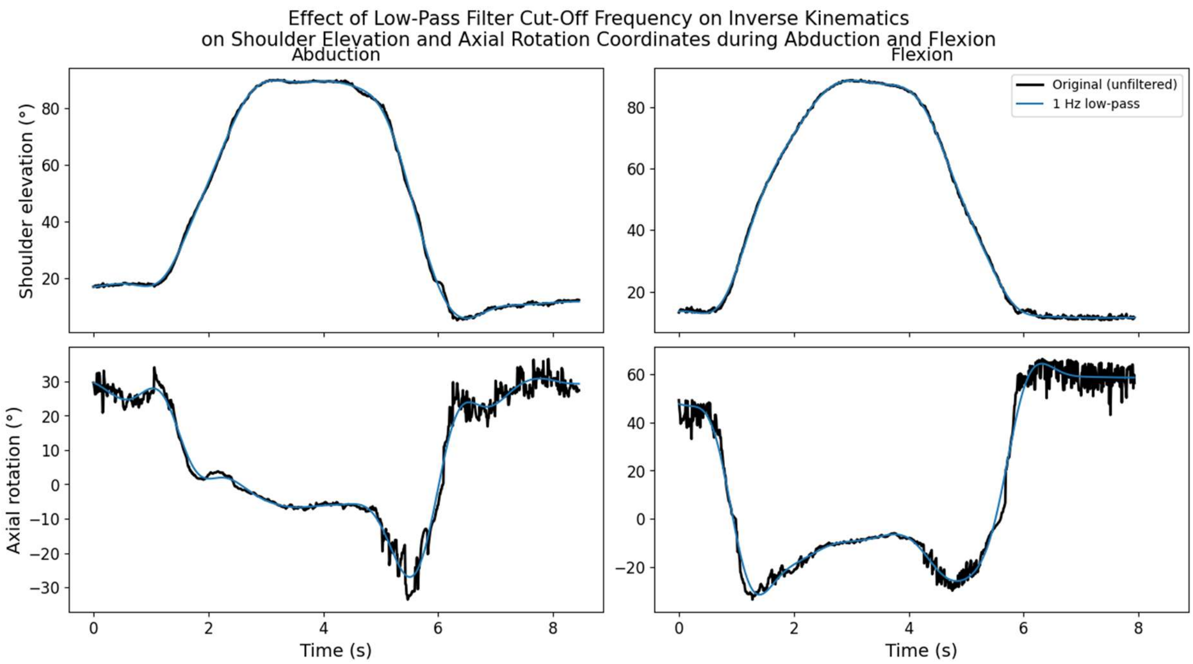

Figure 3.7: Graphs of effect of the 1Hz cut-off frequency on the abduction and flexion motion, for the coordinates shoulder elevation and axial rotation (81).

3.3.1.2 Motion Data Processing

Marker-based measurements are sensitive to soft tissue artefacts and tracking noise, which may introduce high-frequency fluctuations in the reconstructed joint angles (82). To prevent non-physiological accelerations in the inverse-dynamics analysis, the kinematic data were low-pass filtered prior to simulation. The motions were filtered using a cut-off frequency 1 Hz during motion preprocessing. The effect of the 1Hz cut-off frequency onto the motion can be seen in figure 3.7. The filtered motions were subsequently used as prescribed input for all simulations at 50 samples per second.

Note: identical kinematics were applied across all conditions, differences in muscle activation patterns and muscle-driven feasibility between scenarios can only be attributed to changes in muscle strength or externally applied assistance rather than to differences in movement trajectories.

3.3.1.3 Muscle Redundancy Solver

Simulations were performed using the RMR solver in python scripts with prescribed joint kinematics, more detailed information about the RMR solver can be found in Background 2.4. The simulation ran with Python was conducted in VScode software. At each time step, muscle activations were solved in accordance with acceleration matching and activation continuity constraints.

Exoskeleton assistance was implemented as an externally applied, angle-dependent tensile force. This assistive force contributed to the shoulder elevation moment and reduced the torque that had to be generated by the shoulder musculature during the simulated movements.

Reserve actuators were included at all unlocked degrees of freedom to ensure numerical feasibility. Their activation magnitude was recorded and used to assess the ability of the muscle set to reproduce the prescribed motion.

3.3.1.4 Solver Settings

Unless otherwise specified, all RMR solver settings were adopted from the default configuration. Only parameters directly relevant to convergence and constraint enforcement were adjusted for the present simulations. Activation rate limits were applied through the solver's built-in activation constraint formulation.

The solver tolerance parameter *xDelta* does not modify activation dynamics or the simulation time step. Instead, *xDelta* defines the allowable tolerance on the linear acceleration matching constraint within the Rapid Muscle Redundancy (RMR) optimization. Specifically, prescribed joint accelerations are enforced within a bounded tolerance interval, such that deviations between muscle-generated and prescribed accelerations are permitted within ±*xDelta*.

Smaller xDelta values therefore enforce stricter matching of prescribed accelerations, whereas larger values relax this constraint and allow greater deviation. Increasing xDelta reduces the strictness of the acceleration constraint and can thereby reduce the required contribution of reserve actuators, potentially improving convergence in challenging conditions.

Based on this behaviour, *xDelta* values were selected separately for abduction and forward flexion simulations. For abduction, *xDelta* was set to 0.005 with an upper bound of 0.02. For forward flexion, a primary *xDelta* of 0.015 was used with the same upper bound as above. The upper bound was only used when convergence could not be achieved using the primary *xDelta*, which occurred predominantly in forward flexion simulations and in the healthy reference condition for abduction. Automatic increases of *xDelta* were disabled for weakened abduction conditions.

3.3.2 Representation of Shoulder Muscle Weakness

Shoulder muscle weakness was represented by uniformly reducing the F_{max} of selected shoulder muscles primarily innervated by C5-C6 and contributing to elevation within the musculoskeletal model (83-86). The weakened muscles comprised the anterior, middle, and posterior portions of the deltoid, the supraspinatus, the infraspinatus, the clavicular portion of pectoralis major, and the long head of the biceps brachii. The original maximum muscle strength per modelled muscle part can be found in table 3.2.

Muscle	*Muscle part*	*Model name*	*Original F_{max} (N)*
Deltoid	Pars anterior	DeltoideusClavicle_A	707.7
	Pars medialis	DeltoideusScapula_M	2597.8
	Pars posterior	DeltoideusScapula_P	1324.4
Infraspinatus	Pars inferior	Infraspinatus_I	1037
	Pars superior	Infraspinatus_S	967.4
Supraspinatus	Pars posterior	Supraspinatus_P	326.2
	Pars anterior	Supraspinatus_A	543.2
Pectoralis Major	Pars clavicularis	PectoralisMajorClavicle_S	408.8
Biceps brachii	Caput longum	BIC_long	485.8

Table 3.2: Original maximum muscle force values (F_{max}) for the anterior, middle, and posterior deltoid, supraspinatus, infraspinatus, clavicular portion of pectoralis major, and long head of the biceps brachii, reported per modeled muscle subdivision in the OpenSim shoulder model (81). The model-specific muscle names are provided as implemented in OpenSim, reflecting individual muscle subdivisions and, where applicable, their anatomical origin.

The reduced F_{max} levels were calibrated to represent clinically interpretable MRC grades (2-4), based on the ability to complete standardized elevation tasks under defined loading conditions (see 2.3 for more information about the MRC scale). MRC definition per level are given in table 3.3.

	Clinical meaning	*Feasible*	*Not feasible*
MRC2	Full aRoM with gravity eliminated	Gravity eliminated	Gravity
MRC3	Full aRoM against gravity only	Gravity only	1 kg
MRC4	Full aRoM against moderate resistance	Gravity + 1kg	3 kg

Table 3.3: Clinical meaning and mechanical calibration of simulated MRC levels.

For loading conditions requiring external resistance (no gravity, gravity, +1 kg, or +3 kg), the additional mass was implemented by increasing the mass of the hand segment in the OpenSim model. This approach allowed the external load to be represented as distal segment mass, thereby affecting the gravitational shoulder moment without altering joint kinematics or inertial structure elsewhere in the model.

Only F_{max} of the potentially affected muscles was altered (table 3.2). Muscle activation constraints, tendon properties, and joint kinematics were kept identical across conditions.

The F_{max} percentages were determined in a preliminary calibration phase prior to the assistive simulations. For each movement direction, F_{max} was iteratively reduced until the model approached the transition between muscle-driven feasibility and non-feasibility according to the predefined reserve actuator criteria. The selected F_{max} value represented the lowest strength level at which the prescribed motion could still be completed without exceeding the non-feasibility thresholds. The resulting values were subsequently fixed and applied consistently across all assistance conditions. The exact muscle strength reduction percentages corresponding to each MRC grade and movement condition are reported in table 3.4.

	MRC2	*MRC3*	*MRC4*
Abduction	10%	13%	18%
Flexion	2%	5%	7%

Table 3.4: Percentage of original maximal muscle force (F_{max}) retained for each simulated MRC grade and movement condition.

For each prescribed movement, simulations were performed under three primary conditions:

1. a healthy reference condition with full maximal muscle strength (100% F_{max}) and no assistance,
2. weakened conditions representing graded F_{max} reductions (MRC24) without assistance, and
3. weakened conditions combined with externally applied shoulder elevation assistance, divided in:
    a. Partial gravity compensation, and
    b. Modelled cam-based exoskeleton actuator based on Asgari.

3.3.3 Modelling of Assistive Strategies

3.3.3.1 Gravity-Based Assistance Model

First, gravity-based assistance was implemented as a simplified modelling approach to isolate the biomechanical effects of partial gravitational compensation, since the Asgari exoskeleton also based their function on gravity compensation technique. Assistance was modelled by uniformly scaling the gravitational acceleration vector applied to the musculoskeletal model, thereby reducing gravitational loading while preserving inertial properties and joint kinematics.

An initial gravity reduction of 25% was applied as stated ideal gravity compensation in healthy individuals by Asgari et al. (2023), and which was also approximately achieved during their experiments (40). When necessary, gravity compensation was adjusted in increments or decrement of 5% to determine the minimal level of compensation required to restore functional shoulder elevation in terms of full range of motion for each muscle weakness condition (40).

Gravity scaling was applied globally onto the model, its influence on non-target segments (e.g. thorax) was considered acceptable given their limited gravitational displacement during the simulated movements.

3.3.3.2 Asgari Exoskeleton Assistance

The Asgari exoskeleton assistance model was developed to represent the mechanical behaviour of the previously published passive cam-based shoulder exoskeleton proposed by Asgari et al. (2023), with its mechanical design and corresponding function described in detail in Background Section 2.5.

Exoskeleton assistance was modelled using an actuator representation to mimic the biomechanical effect of tensile support applied by the shoulder exoskeleton. Assistance was implemented using a 'PathActuator', an OpenSim actuator that applies tensile force along a three-dimensional path between attachment points and can be routed over wrapping geometries. The PathActuator represented the resultant tensile force transmitted through the cam-spring and Bowden cable mechanism and was implemented as nan externally prescribed force, not included in the muscle recruitment optimization criterion.

The added value of incorporating a physical actuator into the model compared to simplistic gravity compensation is that the actuator applies a tensile force along a defined three-dimensional path originating at the thorax and inserting at the humerus, thereby generating a shoulder elevation moment about the glenohumeral, acromioclavicular, and sternoclavicular joint. This allows the simulation to capture the mechanical interaction between the exoskeleton and the musculoskeletal system, including the influence of cable routing and pulling direction.

Actuator Routing and Geometry

The representation of exoskeleton assistance was implemented using OpenSim Creator software to include the exoskeleton cable routing and its interaction with the musculoskeletal model. Exoskeleton assistance was represented using a PathActuator, which provides an appropriate abstraction of the cam-spring and Bowden cable mechanism by applying a external tensile force along an anatomically correct and exoskeleton defined path.

The actuator path was defined to closely follow the intended Bowden cable route. The proximal attachment point was defined in the thorax reference frame at the location corresponding to the back-mounted cam-spring on the back-plate. The path was then routed to a superior guiding point representing the self-aligning spherical bearing of the exoskeleton; because the exact location was not reported, this point was estimated from the provided design images and placed superiorly on the acromion, defined in the scapular reference frame. The distal attachment point was placed at the lateral epicondyle of the humerus to represent the arm sleeve attachment near the lateral elbow epicondyle, defined in the humeral reference frame. This setup allows the PathActuator to adapt dynamically to scapulothoracic and glenohumeral motion, yielding a physiologically correct routing across both abduction and forward flexion.

Because a shortest-path formulation can produce unrealistic shortcuts through anatomical structures, ellipsoidal wrapping objects were used to constrain the PathActuator and achieve a realistic cable trajectory as would be in a physical setting. A large ellipsoid in the thorax reference frame guided the PathActuator over the back-plate region between the proximal attachment and the superior via point. Distal to the via point, a second ellipsoidal wrap object defined in the scapular reference frame guided the PathActuator over the deltoid, approximating the smooth curvature imposed by the Bowden cable outer sheath and preventing intersection with bony structures.

A spherical wrapping object centred at the glenohumeral joint was initially considered as second wrap object to approximate a constant moment arm. However, due to the shortest-path formulation of the PathActuator, a purely spherical geometry allowed the cable to take an unrealistic posterior route, effectively “cutting through” the posterior deltoid region. To prevent this artefact, a slightly enlarged and posteriorly tilted ellipsoid was defined. By increasing the posterior radius relative to the anterior and lateral dimensions, the ellipsoid discouraged non-physiological posterior wrapping while preserving a physiologically plausible anterior-lateral cable path. The final path of the PathActuator and the use of the wrapping objects can be seen in figure 3.8.

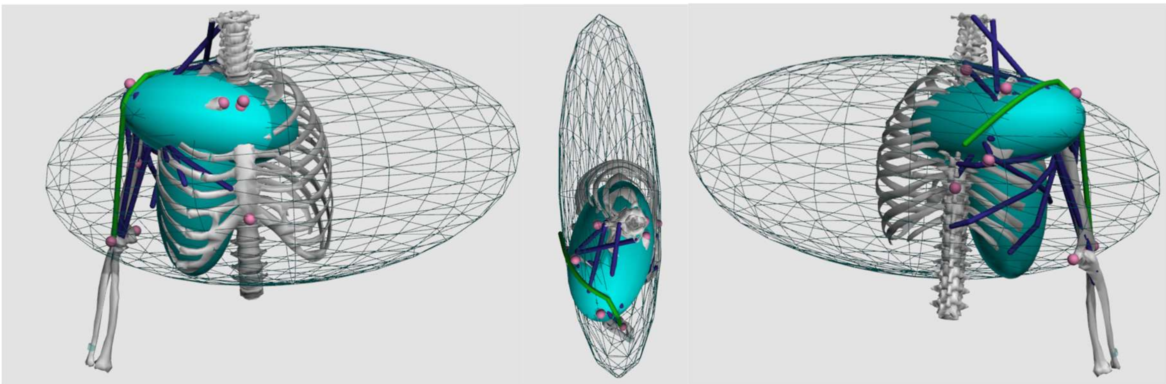

Figure 3.8: Anterior, superior, and posterior views of the shoulder model showing the ellipsoidal wrapping objects used to define PathActuator path. A big ellipsoid is used for back-wrapping (black frame wire), while a distal ellipsoid near the shoulder joint (blue) guides the path actuator (green) during abduction and shoulder flexion.

Over the region of contact with the wrapping ellipsoid during abduction and forward flexion, the ellipsoid closely approximated this spherical surface achieved with minor alternations compared to initial forming, allowing the PathActuator moment arm round the glenohumeral joint to be treated as approximately constant over the simulated range of motion. This enabled the use of a single moment-arm value for frontal- and sagittal-plane shoulder elevation. Based on this routing and wrapping configuration, the actuator moment arm (r) was fixed at 0.048 m for all simulations.

Computation of Humerothoracic Elevation Angle

Because no single model coordinate directly represents overall humerothoracic elevation (see table 3.1), the humerothoracic elevation angle ($\alpha(t)$) was computed from model kinematics at each simulation time step. Using OpenSim output data, an arm orientation vector was defined between shoulder and elbow markers, while a trunk orientation vector represented the thorax. Both vectors were projected onto the plane of motion (frontal plane for abduction, sagittal plane for forward flexion). This time-varying angle was used to evaluate the gravitational shoulder moment and, via the predefined assistance factor, the corresponding assistive shoulder torque in order to calculate the required tensile force of the PathActuator. The resulting time-dependent tensile force was applied as a prescribed input to the PathActuator, ensuring that exoskeleton assistance scaled consistently with shoulder elevation throughout the movement.

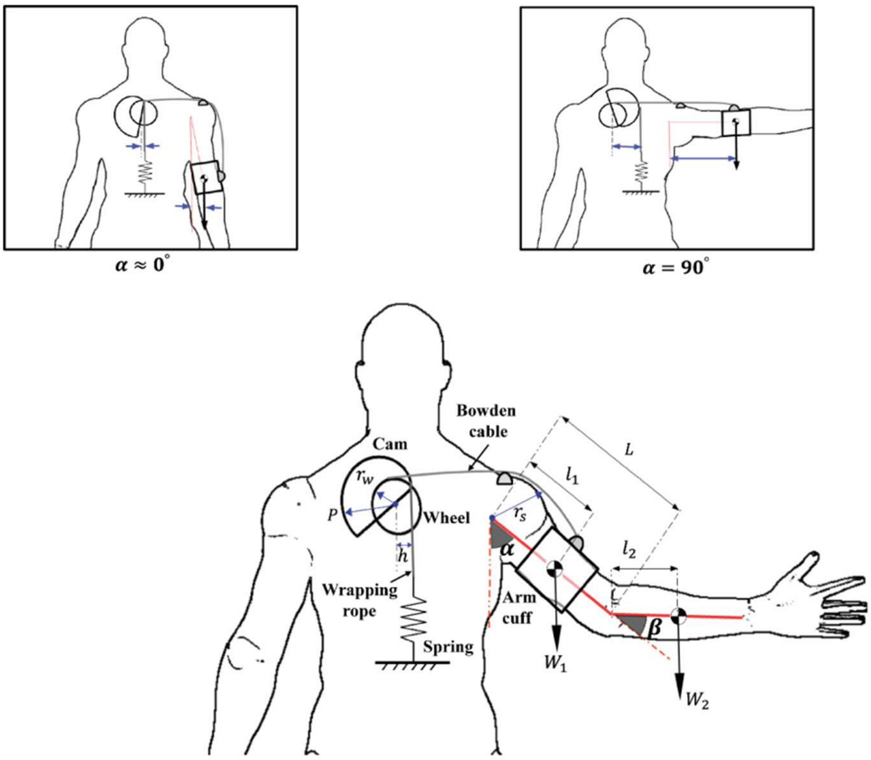

Figure 3.9: Schematic overview of the first version of the Asgari shoulder exoskeleton geometry used to define the assistive moment as a function of shoulder (α) and elbow (β) configuration. r_w denotes the cam-wheel cadius, r_s the effective moment arm of the cable around the glenohumeral joint, and P the rope-cam contact point. L represents the humeral length, with l_1 and l_2 indicating the proximal and distal segment distances used in the gravitational moment formulation. W_1 and W_2 denote the gravitational forces acting on the humerus and forearm-hand segment, respectively. From Asgari et al 2023. (40).

Gravitational Shoulder Moment and Assistive Torque

The formulation of the gravitational shoulder elevation moment (τ_{grav}) was based on a previously published article of Asgari et al. (2023), where the same planar arm model and moment decomposition approach was followed, of which an schematic overview can be seen in figure 3.9 (40). The gravitational moment was expressed as a function of humerothoracic elevation angle and elbow flexion angle (β). The gravitational shoulder elevation moment was derived using information from the used model of the humerus and a combined forearm-hand segment. The gravitational moment (τ_{grav}) acting about the glenohumeral joint was expressed as:

$$\tau_{grav}(\alpha, \beta) = W_1 l_1 sin(\alpha) + W_2\big(Lsin(\alpha) + l_2 sin(\alpha + \beta)\big) \tag{8}$$

Here are W_1 and W_2 the gravitational forces acting on the humerus and the combined forearm-hand segment, respectively; l_1 is the distance from the glenohumeral joint centre to the humeral centre of mass; l_2 is the distance from the elbow joint centre to the forearm-hand centre of mass; L is the humeral length; α is the humerothoracic elevation angle; and β is the elbow flexion angle.

Because elbow flexion was assumed constant throughout the simulated movements, β was set to 0, but also was already stated as locked. Under this assumption, the expression simplifies to:

$$\tau_{grav}(\alpha) = \left(W_1 l_1 + W_2(L + l_2)\right) sin(\alpha) \tag{9}$$

Substitution of segment masses, lengths, and centre-of-mass locations obtained from the musculoskeletal model yielded:

$$\tau_{grav}(\alpha) = 8.2 sin(\alpha) Nm \tag{10}$$

Exoskeleton assistance moment (τ_{assist}) was defined as a fraction k of the gravitational shoulder moment:

$$\tau_{assist}(t) = k \cdot \tau_{grav}\left(\alpha(t)\right) \tag{11}$$

The corresponding PathActuator tensile force (F_{act}) was calculated as:

$$F_{act}(t) = \frac{\tau_{assist}(t)}{r} \tag{12}$$

where r = 0.048 m was the earlier defined fixed effective PathActuator moment arm.

Simulations were initiated at k = 0.25, after which the assistance factor was adjusted in increments of 5% to determine the minimal level of exoskeleton assistance required to complete the range of motion under each muscle weakness condition.

The gravitational shoulder moment was calculated based on the humerothoracic elevation angle relative to the thorax (α). This formulation assumes that the thorax is aligned with the global vertical axis. Analysis of the prescribed motion data showed that thorax inclination deviated less than approximately 4° from the global vertical during both abduction and forward flexion. Given this limited deviation, the resulting effect on the gravitational moment was considered negligible. This simplification was therefore deemed acceptable within the scope of the present feasibility study.

3.3.4 Simulation Outcome Measures

To compare the different simulation conditions (healthy reference, graded weakness without assistance, graded weakness with gravity compensation, and graded weakness with exoskeleton assistance), three primary outcome measures were defined:

1. Maximal active range of motion
2. Peak muscle activation
3. Cumulative muscle activation

The healthy simulation without weakness and without exoskeleton assistance served as the baseline condition for comparison.

3.3.4.1 Maximal Active Range of Motion

Functional elevation capacity was quantified using the concept of maximal active range of motion (aRoM). This parameter represents the highest elevation angle that can be completed without excessive reliance on reserves.

Because identical kinematics were prescribed across all simulations, loss of muscle-driven feasibility was detected using reserve actuator behaviour. For each unlocked degree of freedom (defined as coordinate), the maximal absolute reserve actuator moment observed in the healthy reference simulation was used as baseline. This approach was chosen to prevent trivial absolute increases in very small reserve moments from being misclassified as functional failure, as relative percentage changes may appear large despite negligible absolute differences. Only non-axial coordinates were considered decisive for detection of functional failure, as the primary objective of the exoskeleton is to support shoulder elevation rather than axial rotation.

Loss of muscle-only feasibility was defined when both of the following criteria were met:

- The reserve actuator moment exceeded ten times the healthy-reference baseline for that coordinate, and
- The reserve actuator contributed at least 10% of the required inverse dynamics moment for a minimum duration of 0.1 seconds.

The 10% threshold was selected based on the OpenSim guideline indicating that reserve actuator contributions should remain below approximately 10% of the total joint moment to ensure physiologically realistic muscle-driven simulations (69). The humerothoracic elevation angle at the first occurrence of these criteria was defined as the maximal active range of motion. Completion of the full prescribed motion without meeting these criteria was interpreted as full muscle-driven elevation capacity.

For assisted conditions, this criterion was additionally used to determine the minimal level of assistance required to restore full muscle-driven completion of the movement. Assistance magnitude was incrementally increased or decreased until the motion could be completed without exceeding the predefined reserve actuator thresholds.

It should be noted that all simulations were executed with prescribed kinematics. Consequently, even in conditions where muscle capacity was insufficient, the full motion could still be completed through increased reliance on reserve actuators. In such cases, muscle activations may reach maximal values while reserves provide the additional required moment, which is not directly visible in muscle activation plots. The aRoM metric therefore represents the maximal elevation angle achievable without excessive reserve contribution rather than purely the maximal simulated angle.

3.3.4.2 Neuromuscular Activation Demand

Neuromuscular activation demand during elevation was quantified using muscle activation metrics. Because identical kinematics were prescribed across conditions, joint torques were equivalent between scenarios. Therefore, changes in muscle activation directly reflect differences in how the required joint moments were distributed over the musculature under varying strength and assistance conditions.

Peak Muscle Activation

First, peak muscle activation (A_{max}) of weakened shoulder elevators was used to represent relative neuromuscular activation demand. High peak activation values approaching the upper bound (1.0) indicate operation near maximal capacity and reflect overdemand relative to available strength.

Cumulative Muscle Activation

To capture sustained neuromuscular demand throughout the movement, the area under the activation-angle curve (AUC) was calculated as the cumulative muscle activation (A_{cum}) for selected muscles over the full duration of the prescribed motion.

A_{cum} represents the integrated, normalized muscle activation over the elevation trajectory. Because muscle activation in OpenSim reflects the neural drive required relative to maximum force-generating capacity, A_{cum} provides an index of cumulative relative neuromuscular activation demand.

Importantly, A_{cum} does not directly quantify mechanical work, metabolic energy expenditure, ATP consumption, or physiological muscle fatigue, as no metabolic or fatigue model was implemented. Instead, cumulative activation was used as a biomechanical proxy for sustained relative activation demand. Experimental studies have demonstrated that higher relative muscle activation levels are associated with greater peripheral fatigue development during sustained or high-intensity contractions (e.g., Ducrocq & Blain, 2022), supporting the relevance of integrated activation metrics as indicators of fatigue susceptibility (87).

Changes in A_{cum} between weakened and assisted conditions were therefore interpreted as relative changes in sustained neuromuscular activation demand of the weakened shoulder elevators, rather than as direct measures of physiological fatigue.

3.3.4.3 Activation Redistribution and Stability Considerations

To assess whether assistance introduced unintended biomechanical trade-offs, activation patterns of non-weakened and stabilizing muscles were analysed. Particular attention was given to rotator cuff muscles and other synergists contributing to glenohumeral stability.

The glenohumeral joint stability constraint embedded in the Rapid Muscle Redundancy solver was maintained across all simulations. Stability was therefore enforced in all conditions. Analyses focused on whether assisted conditions required disproportionate increases in stabilizing muscle activation in order to satisfy this constraint.

Assistance was considered biomechanically favourable when it:

- Restored muscle-driven elevation capacity to the full range of the motion,
- Reduced peak and cumulative muscle activation, as a proxy for neuromuscular activation demand of weakened elevators, compared to the corresponding weakened baseline condition, and
- Did not induce more than a twofold increase in activation of other muscles or increased reliance on reserve actuators.

3.3.4.4 Scenario Comparison Framework

For each movement direction and each simulated muscle strength level, four scenarios were evaluated, as previously mentioned in section 3.3:

1. Healthy condition: full F_{max}, without assistance
2. Weakened condition: reduced F_{max} of selected shoulder muscles according to representative MRC levels, without assistance
3. Assisted condition: reduced F_{max} combined with externally applied shoulder elevation assistance, implemented using two assistance profiles:
    a. Gravity-based assistance: partial gravity compensation
    b. Physically based exoskeleton assistance: application of a 'PathActuator' generating tensile force to reproduce the cam-based shoulder exoskeleton

Comparisons were performed:

- Within each MRC level to evaluate the effect of assistance relative to the weakened condition, and
- Relative to the healthy reference to interpret the extent to which assistance restored biomechanical behaviour toward non-weakened patterns.

This structured comparison enabled consistent evaluation of both gravity-based and physically modelled exoskeleton assistance across graded levels of shoulder muscle weakness, and an overview per condition can be found in Table 3.5.

Condition	*MRC level*	*Strength*	*Assistance*	*Outcomes*
1	Healthy	100% F_{max}	None	Max. aRoM A_{cum} A_{max}
2	MRC2	Reduced F_{max}	None	Max. aRoM A_{cum} A_{max}
3a	MRC2	Reduced F_{max}	Gravity compensation	Max. aRoM A_{cum} A_{max}
3b	MRC2	Reduced F_{max}	Exoskeleton assistance	Max. aRoM A_{cum} A_{max}

Table 3.5: Within-level comparison structure used for all simulated MRC levels, MRC2 is shown as illustrative example.

4 Results

4.1 Interview Findings on Daily Activity Limitations and Support Needs

4.1.1 Participants

A total of seven participants were included in this study. All participants were female, and the median age was 27 years (range 23-49 years).

Two participants were recruited via the outpatient clinic of the Leiden NerveCenter of the LUMC, from a total of four individuals who were approached. Five participants were recruited through an open recruitment message via social media in agreement with the team communication advice of the LUMC. Of the seven individuals who initially responded to this call, one did not respond further, and one registered after the inclusion period had closed.

Clinical characteristics of the participants are summarized in table 4.1. Participants differed with regard to side of brachial plexus involvement (left, right, or bilateral), neurological level of the lesion, Mallet score and history of surgical interventions. For one participant, the neurological level of the brachial plexus lesion was unknown.

Characteristics	*Total (N=7)*
Sex (male/female)	0/7
Age (years), median [min-max]	27 [23-49]
Side of brachial plexus lesion (left/right/bilateral)	2/2/3
Neurological level of plexus lesion	
C5-6	3
C5-7	2
C5-8	1
Unknown	1
History of surgery (yes/no)	6/1
Nerve transfer	3
Derotational humeral osteotomy	2
Tendon or muscle transfer	2
Dynamic assistive devices	2
Dynamic arm stabilizer	2
Supportive devices	4
Sling	2
Brace (elbow/shoulder)	3/1
Clinical characterisation	
Mallet score	
hand-to-mouth, median [min-max]	3.5 [1-4]
hand-to-head, median [min-max]	3 [2-4]
Active RoM in quartiles	
abduction, median [min-max]	2.5 [1-4]
forward flexion, median [min-max]	2.5 [1-4]

Table 4.1: Demographic, clinical, and functional characteristics of participants with Erbs' palsy included in the qualitative interview study.

4.1.2 Thematic Analysis of Interview Findings

The final themes and categories identified through the analysis are presented in the word web shown in figure 4.1, and provides an overview of the main themes and their associated categories, together with the most prominent relations between themes observed in the data. To provide a concise overview of the principal findings per theme, table 4.2 at the end of this section summarizes the core elements emerging from the thematic analysis.

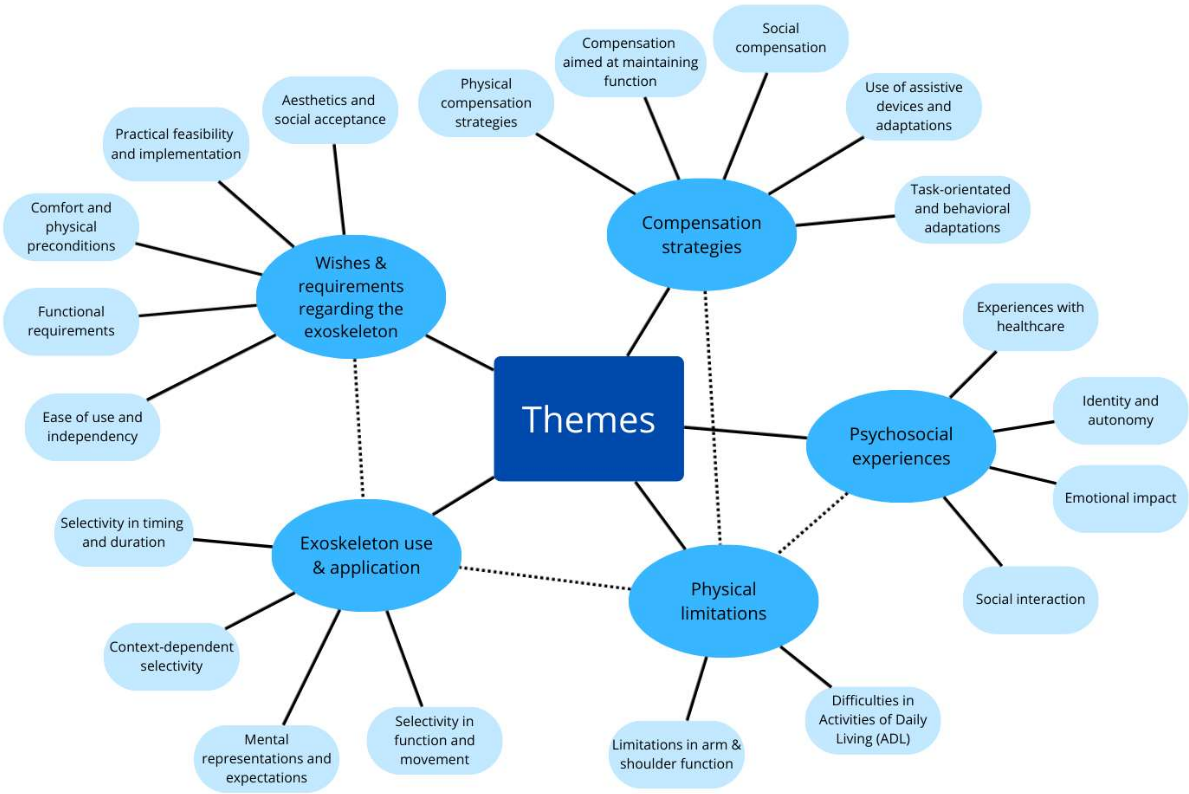

Figure 4.1: Visualization of the thematic analysis structure. Themes are shown in blue shades. Categories linked to each theme are displayed in the outer layer. Solid black lines indicate hierarchical relations, whereas dashed lines highlight recurring relationships between themes.

Physical limitations had a pattern with compensation strategies, psychosocial experiences, and exoskeleton use and application. In addition, a pattern was observed between exoskeleton use and application and wishes and requirements regarding the exoskeleton. The connections shown reflect recurring patterns in the data and do not imply causal relationships between themes.

4.1.2.1 Theme Physical Limitations in Daily Functioning

Across all interviews the participants described physical limitations related to arm and shoulder function as a central and persistent aspect of daily life. These limitations most frequently involved rapid physical fatigue, strength deficits during functional tasks, and restricted active arm elevation, which fell under limitations in arm and shoulder function. Rather than occurring independently, these limitations commonly co-occurred and jointly constrained participation in daily activities, work, and sport or hobby related activities.

Rapid physical fatigue of the arm and shoulder was among the most frequently reported limitations and was described across a broad range of prolonged ADL functions, e.g. eating, household tasks, hobbies, and sports.

> *Since using a dynamic arm stabilizer: "I notice that I now eat very slowly, because at some point I started eating faster and faster, almost leaning halfway onto the table just to finish as quickly as possible, because it started to hurt." (P3, bilateral, C5-7)*

> *"And I also notice that when I have to hang up lots of small items, like socks or something like that. Nine out of ten times I just dry them, but when I do hang them up, at some point I*

end up doing everything only with my right arm, because the left one is simply fatigued. So that's really something where the pain, well, I wouldn't really call it pain, but it's just unpleasant. It's an uncomfortable feeling." (P4, left, C5-6)

Strength deficits during functional tasks commonly popped up together with limitations in active arm elevation, particularly during activities requiring lifting at or above shoulder height.

"I lack the strength to lift a filled oven dish, but the height is also a problem. It's at shoulder height, and that's no longer my working area. My working area is around my waist and lower. Anything above my waist already becomes very demanding." (P1, bilateral, C5-7)

"Or that you just have a little more strength to lift something. I notice that I'm gradually doing less and less. I really enjoy baking and cooking, but I do it less and less because it takes too much effort. Yeah, I do notice that, and I do find that a shame." (P3, bilateral, C5-7)

In participants with bilateral affected arms, these limitations were described as asymmetrical, with each arm contributing different functional capacities.

"My left arm is stronger, that's the one I can do the most with. And my right arm is more flexible, I can lift it higher. So in the end, it balances out quite nicely." (P3, bilateral, C5-7)

4.1.2.2 Theme Compensation Strategies in Response to Physical Limitations

Participants described a wide range of compensation strategies commonly used in situations with physical limitations, particularly rapid fatigue and restricted arm elevation. These strategies included behavioural adaptations, physical compensation through altered movement patterns, and reliance on the less- or non-affected arm.

Behavioural adaptations such as planning, pacing, or avoiding activities were frequently reported and described as necessary to manage energy and prevent overuse.

"So I have to make a choice: do I shower independently, or do I get help with showering and still have enough energy to work? And that's really difficult. For example, if I exercise in the morning, I have to wait until the evening to shower, because my partner isn't home until then." (P1, bilateral, C5-7)

"Some activities, I just don't go to anymore, because I think: yeah, I won't be able to keep that up anyway. And at some point, that's just not fun anymore." (P3, bilateral, C5-7)

Physical compensation strategies involving altered kinematics, such as the use of momentum, gravity, or trunk movement, were also commonly described.

"Just like opening a door. I spent almost a year in rehabilitation at Blixembosch. There I learned to work with momentum. So if I want to open a door, I already have to think: the door is 1.5 meters away, I need to start the swing now, and then I can open the door by basically using gravity." (P1, bilateral, C5-7)

"I can do many things, but if I saw myself, I'd probably think: wow, that's poor form; using speed and tricks and all that. For example, grabbing things from kitchen cabinets, anything where you need to use your arms above shoulder height. I noticed I had a strong tendency to overextend my back to compensate. So sometimes you can do the movement, but you know: this is kind of cheating." (P7, bilateral, C5-6)

Less frequently, but consistently across interviews, participants reported overuse of the less- or non-affected arm.

> *"So often I end up carrying very heavy pans with a lot of food using just one hand." (P5, right, C5-8)*

4.1.2.3 Theme Psychosocial Experiences Accompanying Physical Limitations

Psychosocial experiences were closely intertwined with physical limitations and compensation strategies. Emotional burden, frustration, and challenges related to coping with pain and bodily sensations commonly co-occurred with rapid fatigue and activity restriction.

Frustration related to reduced endurance and thereby restricted participation was frequently expressed.

> *"That's what I find so frustrating, that I'm being held back by this stupid arm." (P3, bilateral, C5-7)*

In some cases, emotional impact extended to major life considerations, such as parenting and independence. These reflections occurred less frequently but were particularly striking.

> *"I do want children, but that's basically not possible. Because how am I supposed to do that? I can't lift a child all the time… Feeding a child will also be difficult, because I'd constantly have to lift my arm forward like this, which is very demanding." (P1, bilateral, C5-7)*

Earlier disappointment with healthcare also contributed to emotional burden, commonly described in terms of disappointment and uncertainty.

> *"Once you turn 18, it's basically: good luck. Figure it out yourself." (P1, bilateral, C5-7)*

> *"Somehow my medical file from my childhood and baby years is just gone… I would have liked to know that [the diagnosis] myself, but unfortunately nobody can tell me." (P2, right, level unknown)*

4.1.2.4 Expectations and Selective Use of A Shoulder Exoskeleton

Expectations regarding the use of a shoulder exoskeleton were closely linked to experienced limitations and emotional burden. Participants described selective use across contexts, functions, and durations, rather than continuous wear.

Work-related contexts were commonly mentioned as potential use situations.

> *"Especially at work, because I'm there nine hours a day. That's where I need it most, and I think it would also be nice to take it off afterwards." (P7, bilateral, C5-6)*

Hobby and sport activities were also frequently identified.

> *"I think it could support me at the music association, during rehearsals and maybe also concerts. That long-term load might become a bit less, making it easier to keep going." (P6, left, C5-6)*

Participants described concrete expected effects, often accompanied by uncertainty.

> *"Maybe it could also be very helpful for cycling, to unload the arm. But I have to be honest, I still know too little about it to really say anything sensible about that." (P2, right, level unknown)*

4.1.2.5 Wishes and Requirements Regarding Exoskeleton Support

Wishes and requirements regarding a shoulder exoskeleton were not expressed as isolated design preferences, but commonly co-occurred with physical discomfort, frustration, and broader future-oriented expectations.

Concerns regarding physical comfort were frequently raised, particularly in relation to pressure on the back, straps, and weight.

> *"What really appeals to me is that it looks much less robotic than I had expected. It's less bulky, but I do wonder to what extent you would feel discomfort from the part on your back when sitting somewhere. That's really the only thing: the part on the back. Can you still sit properly with it without developing back pain again?" (P1, bilateral, C5-7)*

> *"And when I saw the back with that spring and everything, that doesn't look very comfortable. So it could be more compact, if possible, it would be preferable for it to be more compact. And not too heavy either, because it shouldn't feel like a burden to wear something." (P4, left, C5-6)*

> *"It depends on what kind of straps they are. Often they're made with Velcro or something like that, but that can really cut into your skin… when you take it off, you can feel like: oh, that was actually quite tight." (P5, right, C5-8)*

Beyond physical comfort, participants expressed broader desires related to autonomy, independence, self-worth, and mental rest.

> *"It's partly about self-worth, but also about having control again: being able to decide when you do something." (P1, bilateral, C5-7)*

Several participants described hopes that exoskeleton support could reduce frustration, create mental rest, and enable progress.

> *"If something like this could help with daily tasks… then it could be a really useful aid to create a bit of rest in your arm during certain tasks and prevent you from getting frustrated so quickly." (P4, left, C5-6)*

4.1.3 Overarching Wishes and Requirements Inventory

This section presents an overarching inventory of wishes and requirements regarding shoulder exoskeleton support, derived from participants' explicit statements within the theme Wishes and requirements regarding exoskeleton support, as not all the wishes and requirements always co-occurred. The aim is to synthesize concrete user requirements that an exoskeleton would need to meet in order to be acceptable and wearable for this target group. To enhance clarity and conciseness, illustrative quotations are selectively presented to represent shared specific requirements rather than exhaustive repetition.

4.1.3.1 Comfort and Physical Preconditions

As earlier described, physical comfort emerged as a fundamental prerequisite for exoskeleton use. Participants consistently indicated that discomfort would strongly limit acceptance, regardless of potential functional benefits. Key concerns related to pressure on the back, weight, bulkiness, strap-induced compression, and the risk of pressure points, particularly in areas with reduced or absent sensation.

> *"It's less bulky than I expected, but I do wonder how much discomfort you would feel from the part on your back when sitting. That's really the only thing: the part on the back. Can you still sit properly with it without developing back pain again?" (P1, bilateral, C5-7)*

> *"On my right arm, where I would use it, I don't have any sensation at all. Then you might run into pressure point problems again." (P1, bilateral, C5-7)*

Participants emphasized that straps should be adjustable and should not constrict the arm or add to existing physical strain.

> *"As long as the straps are adjustable and don't constrict your arm, I wouldn't see that as a problem. And it shouldn't be too heavy, it shouldn't feel like a burden to wear something." (P4, left, C5-6)*

Several participants expressed a preference for designs that are flexible and body-conforming rather than rigid.

> *"Ideally, it would fit like a sports bra: something that moves with you but still provides support." (P7, bilateral, C5-6)*

4.1.3.2 Comfort Related to Heat, Skin, and Maintenance

In addition to mechanical comfort, participants highlighted requirements related to heat regulation, skin irritation, and hygiene. These aspects were considered important for prolonged or repeated use and for maintaining usability in daily life.

> *"The part that's directly on your body should be easy to clean, whether that's in the washing machine or with a cloth. That's actually the most important thing." (P2, right, level unknown)*

> *"I didn't really wear the physical brace during summer, not necessarily because of visibility, but because it's uncomfortable with heat. I get warm quickly." (P7, bilateral, C5-6)*

Participants indicated that washability and ease of maintenance were key requirements to ensure continued use.

> *"It's important that it's washable. That's probably the most important thing: being able to clean it." (P5, right, C5-8)*

4.1.3.3 Functional Support Requirements

Participants also expressed clear wishes regarding the type of functional support an exoskeleton should provide. Desired support primarily focused on unloading muscles, assisting arm elevation, and supporting activities involving reaching, lifting, eating, and sports.

> *"That upward movement is always a problem. You make that movement many times a day." (P3, bilateral, C5-7)*

> *"If I had to choose one action, it would be being able to reach something without discomfort, especially above shoulder height." (P4, left, C5-6)*

Support was also expected to compensate for strength loss during functional tasks and sports.

> *"I notice my strength drops quickly when something is heavier. Support would be helpful when carrying a pan with food or during cooking." (P5, right, C5-8)*

4.1.3.4 Adaptability and Tuning of Support

A key requirement expressed across all participants was the ability to adjust the level of support. Participants emphasized that support needs vary across activities and over time.

> *"Some activities require more support than others, and as you move more and train your muscles, you might eventually need less support. So being able to adjust it would be ideal." (P1, bilateral, C5-7)*

“You should be able to set it yourself: for some activities you want more support, for others less.” (P5, right, C5-8)

4.1.3.5 Ease of Use and Independence

Ease of use was described as a crucial factor of whether participants would actually use an exoskeleton. Participants emphasized that donning and doffing should be simple, quick, and possible without assistance.

Participants with bilateral affected arms expressed differing preferences regarding one-sided versus two-sided support, while consistently emphasizing simplicity and feasibility.

“Maybe only for the right side. On the left I’m very used to it, and it works reasonably well. Right is just very difficult. I think a double setup would become complicated.” (P1, bilateral, C5-7)

In contrast, another participant with bilateral affected arms expressed a preference for bilateral support, while acknowledging potential design challenges.

“Ideally, it would be nice if it could be balanced, so both sides. I don’t know if that’s possible to design, but if it is, then both; otherwise, the right side.” (P7, bilateral, C5-6)

Ease of independent use was consistently emphasized as essential.

“If it’s a whole operation to wear it, so if I need my partner to help me put it on, that would really hold me back from using it.” (P2, right, level unknown)

“Mainly that it’s easy to put on and take off, and independently. One thing I hated most growing up was having to ask for help.” (P5, right, C5-8)

4.1.3.6 Conditions for Duration of Use and Deployability

Participants did not express a fixed preference for continuous or short-term use. Instead, acceptable duration of use was described as conditional and dependent on comfort, weight, ease of use, and context.

“That depends on how heavy it is and how long per day you would wear it.” (P1, bilateral, C5-7)

“If it’s comfortable enough and it doesn’t cause any harm, then I would wear it all day.” (P2, right, level unknown)

Others emphasized that duration of use should remain adaptable to individual needs and daily activities.

“For daily use, I wouldn’t need it all day, because I’m not reaching all the time.” (P4, left, C5-6)

To facilitate a structured overview of the core findings across themes, Table 4.2 summarizes the principal elements emerging from the qualitative interview study.

Theme	*Core findings*	*Typical expression or patterns*
Physical Limitations in Daily Functioning	Rapid physical fatigue Strength deficits during functional tasks Restricted active arm elevation Limited functional working area (above waist/shoulder height) Discomfort during sustained use	Fatigue across ADL, work, hobbies Tasks above shoulder height most demanding Bilateral asymmetry in function
Compensation Strategies in Response to Physical Limitations	Behavioural adaptation (planning, pacing, avoidance) Use of momentum Thorax overextension Reliance on less-affected arm Task modification	"Working around" limitations Energy-consuming strategies Perceived as unnatural or inefficient
Psychosocial Experiences Accompanying Physical Limitations	Accompanying Physical Limitations Frustration Emotional burden Reduced endurance causing reduced participation Impact on autonomy and independence Healthcare-related disappointment	Feeling held back Activity withdrawal Desire for control and normality
Expectations and Selective Use of a Shoulder Exoskeleton	Selective use (work, hobbies, sustained tasks) Not intended for continuous wear Expected fatigue reduction Anticipated prolonged activity endurance Uncertainty about real-life usability	Context-dependent use Targeted support during high-demand tasks
Wishes and Requirements Regarding Exoskeleton Support	Comfort (back component, straps, pressure points) Lightweight Compact / non-bulky Natural-looking movement Adjustability of support Enhancement of autonomy	Avoid discomfort and stigma Maintain natural kinematics Support without restricting

Table 4.2: Overview of the main findings per theme identified through reflexive thematic analysis of the semi-structured interviews.

4.2 Biomechanical Simulation Results of Shoulder Exoskeleton Assistance

This section presents the results of the biomechanical simulation study evaluating the effects

This section presents the results of the biomechanical simulation study evaluating the biomechanical feasibility of passive shoulder exoskeleton assistance during arm elevation under graded muscle weakness. Simulations were performed for shoulder abduction and forward flexion under strength reductions corresponding to clinically interpretable MRC grades (MRC2-4), alongside a healthy reference condition.

For MRC3 and MRC4 conditions, additional external loads (1 kg and 3 kg, respectively) were applied to reflect clinically representative functional demands, as described in the Methods section.

Results are structured according to the predefined experimental conditions:

1) healthy reference,

2) weakened without assistance,

3a) weakened with gravity compensation, and

3b) weakened with modelled exoskeleton assistance.

For each movement direction, outcomes are evaluated using three primary biomechanical metrics:

- maximal muscle-driven active range of motion,
- peak muscle activation (A_{max}), and
- cumulative muscle activation expressed as area under the activation-elevation curve (A_{cum}).

Results are first presented for functional performance (aRoM), followed by analyses of muscle activation demand and potential compensatory activation patterns. Detailed muscle activation profiles are primarily illustrated for the MRC2 condition, as higher MRC levels demonstrated similar qualitative trends with differences mainly in magnitude. Complete activation profiles for MRC3 and MRC4 are provided in Appendix B.

Where applicable, tables and comparative figures summarize differences between experimental conditions to facilitate direct interpretation of the biomechanical effects of assistance.

4.2.1 Shoulder Abduction

The effects of shoulder exoskeleton assistance during arm elevation were first evaluated for shoulder abduction.

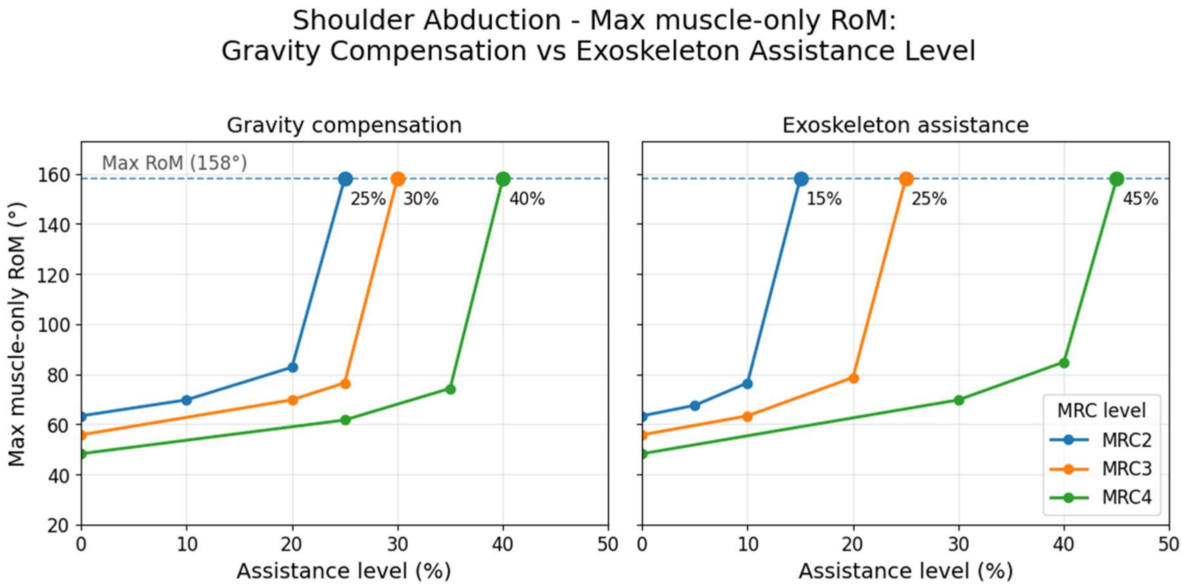

Figure 4.2. aRoM during shoulder abduction across MRC levels and assistance levels.

4.2.1.1 Functional Performance: Range of Motion

In the healthy reference condition, full shoulder abduction was achieved without reliance on reserve actuators under MRC conditions.

All weakened conditions (MRC 2-4) demonstrated a reduced active range of motion, with decreasing maximal elevation as muscle weakness and external load increased (table 4.3). In the MRC2 condition, abduction was only feasible without gravity, while under normal gravity loss of muscle-only feasibility occurred at a humerothoracic elevation of 63.3°. For MRC3 and MRC4, reserve actuator detection occurred at progressively lower elevation angles when external loads were applied, indicating increasing functional limitation.

In addition to the restoration of full abduction, increasing assistance resulted in a gradual increase in maximal aRoM across all MRC levels (Figure 4.2). The required assistance level to achieve full active range of motion differed between gravity compensation and exoskeleton assistance.

MRC level	*Condition*	*Assistance level*	*Max aRoM*
MRC2	No assistance	-	63.3°
	Gravity compensation	25%	Full aRoM
	Exoskeleton assistance	15%	Full aRoM
MRC3	No assistance (1kg load)	-	55.8°
	Gravity compensation	30%	Full aRoM
	Exoskeleton assistance	25%	Full aRoM
MRC4	No assistance (3kg load)	-	48.2°
	Gravity compensation	40%	Full aRoM
	Exoskeleton assistance	45%	Full aRoM

Table 4.3: Assistance level required to achieve full aRoM during abduction. Full aRoM indicates completion of the prescribed abduction trajectory.

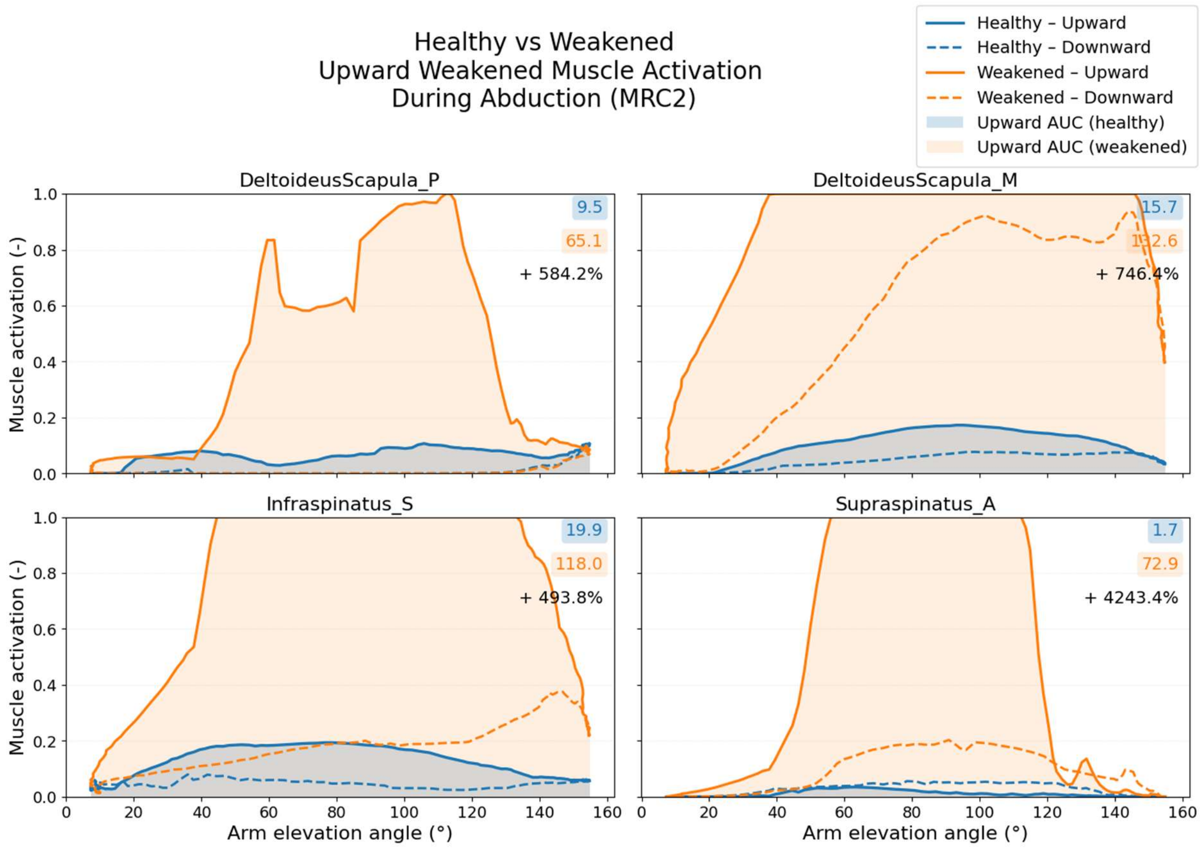

Figure 4.3: Activation-angle loops of weakened shoulder elevators during abduction in the healthy and weakened MRC2 conditions. Shaded areas denote the upward-phase A_{cum}. Relative to healthy, upward A_{cum} increased by +584.2%, +746.4%, +493.8%, and +4243.4% for the four muscles shown, respectively.

4.2.1.2 The Effect of Weakness

In the MRC2 condition, all four weakened abductors showed markedly increased activation compared to healthy. The activation patterns of the superior and inferior portions of both the supraspinatus and infraspinatus were highly comparable across conditions; therefore, one representative portion per muscle was reported for clarity, as the second portion demonstrated similar behaviour and did not alter the overall interpretation.

The posterior deltoid increased from an upward A_{cum} increase of 584.2%, and showed a non-monotonic profile with a visible dip in activation around mid-elevation before rising again. The middle deltoid increased with 746.4%, reaching maximal activation early in the upward phase and remaining elevated across most of the range. The infraspinatus increased with 493.8%, with activation rising steeply after 40° and remaining high until late elevation. The supraspinatus showed the largest relative increase of 4243.4%, shifting from minimal activation in the healthy condition to near-maximal activation from approximately 60° to 120°.

Across all four muscles, the weakened condition was characterised by expanded loops and extended high-activation plateaus compared to healthy.

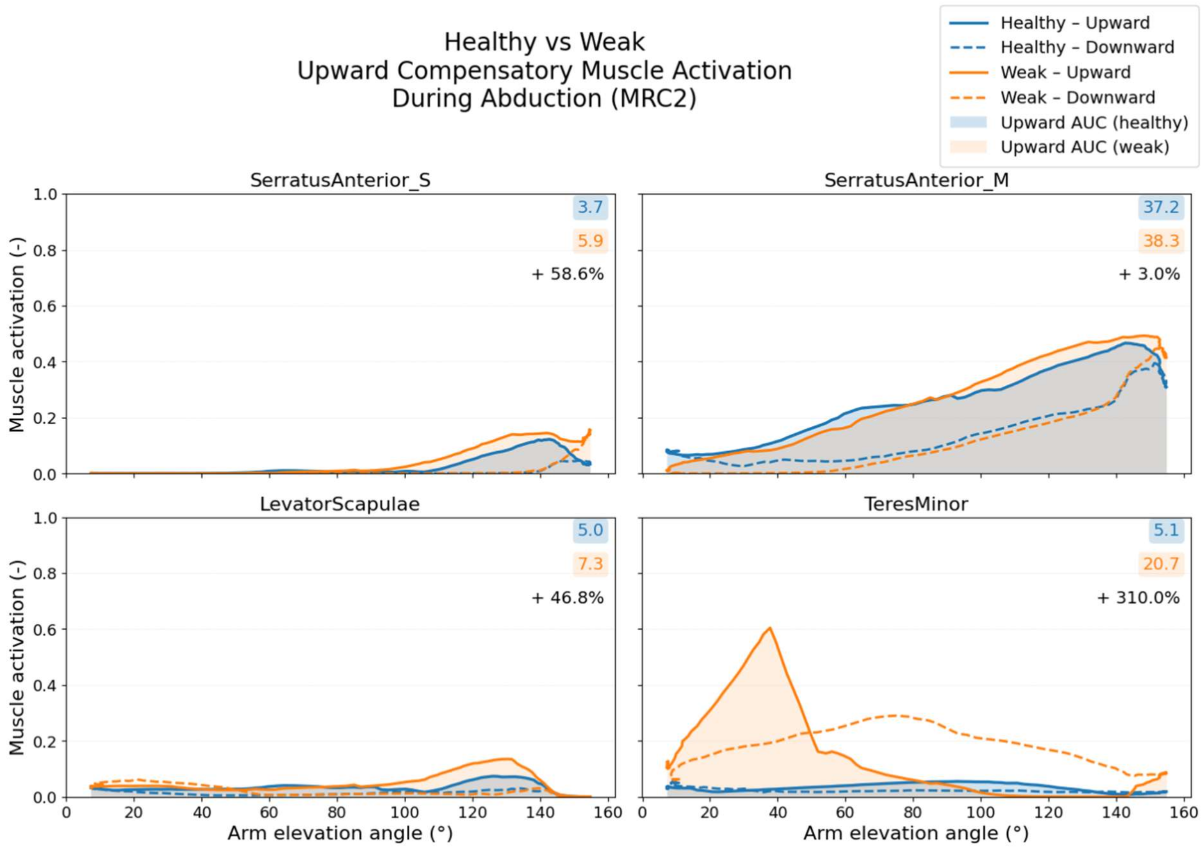

Figure 4.4: Activation-angle loops of compensatory shoulder muscles during abduction in the healthy and weakened MRC2 conditions. Shaded areas denote the upward-phase A_{cum}. Relative to healthy, upward A_{cum} increased by +58.6%, +3.0%, +46.8%, and +310.1% for the four muscles shown, respectively.

In the MRC2 condition, all four compensatory muscles showed increased activation compared to healthy.

The (scapulothoracic) serratus anterior (superior part) increased by 58.6%, with slightly higher activation primarily at higher elevation angles. The (scapulothoracic) serratus anterior (middle part) showed a small increase of 3.0%, with largely similar activation patterns between conditions. The (scapulothoracic) levator scapulae increased by 46.8%, showing higher activation mainly in the second half of elevation. The (scapulohumeral) teres minor showed the largest relative increase of 310.1%, characterized by a pronounced early peak around 30-50° that was not present in the healthy condition.

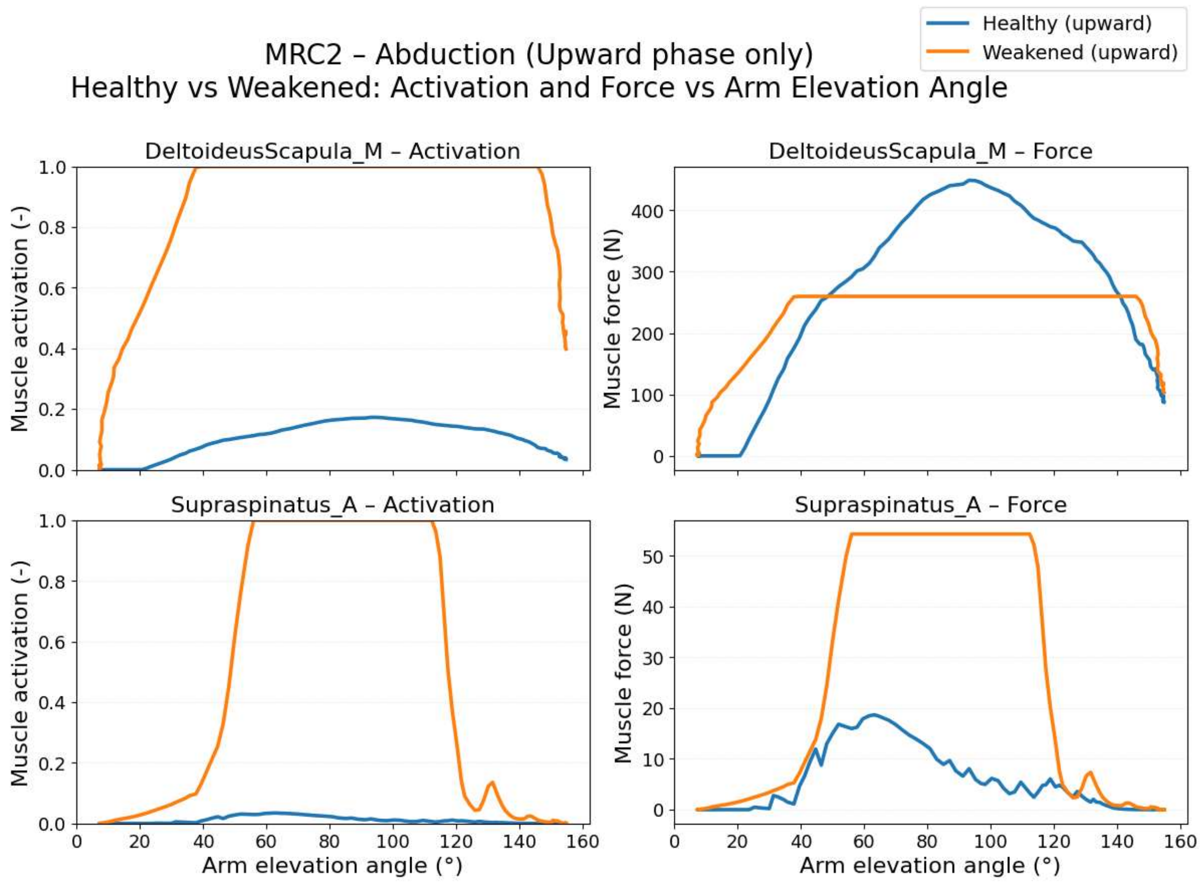

Figure 4.5: Muscle activation and muscle force as a function of arm elevation angle during the upward phase of shoulder abduction (MRC2). Data are shown for middle deltoid (top) and anterior part of the supraspinatus (bottom), comparing healthy and weakened conditions.

4.2.1.3 Activation versus Force Comparison

In the weakened condition, both muscles show near-maximal activation over a large part of the elevation range, whereas activation remains clearly lower in the healthy condition. Despite this high activation, the corresponding muscle forces in MRC2 are substantially lower than in the healthy reference. This can be seen in figure 4.5.

Thus, under reduced maximal muscle strength, increased activation does not result in comparable force production, indicating a force-limited rather than activation-limited condition.

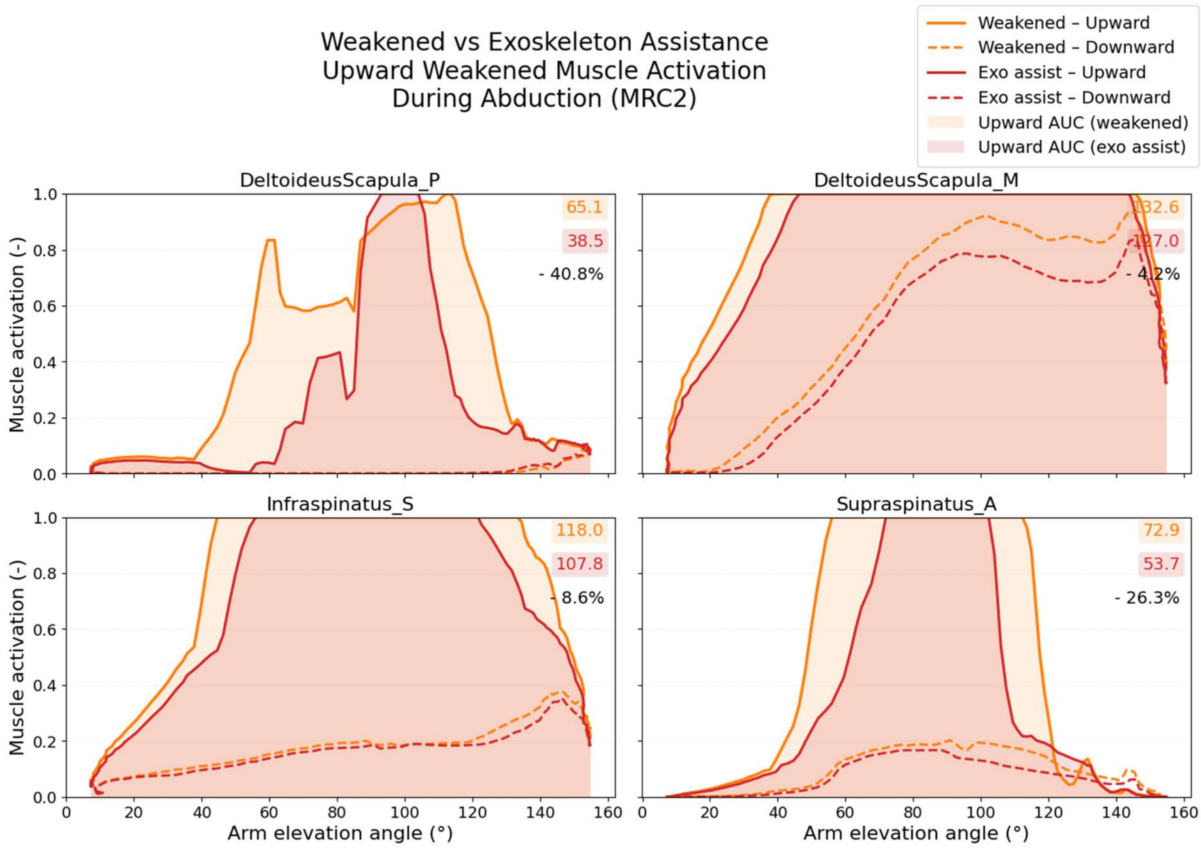

Figure 4.6: Activation-angle loops of weakened shoulder elevators during abduction in the unassisted MRC2 condition and with exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the unassisted weakened condition, upward A_{cum} decreased by -40.8%, -4.2%, -8.6%, and -26.3% for the four muscles shown, respectively.

4.2.1.4 The Effect of Exoskeleton Assistance

With exoskeleton assistance, activation decreased in all four shown weakened elevators relative to the unassisted MRC2 condition.

The posterior deltoid decreased by 40.8%, with a shorter near-maximal activation plateau and a remaining irregular, step-like activation profile during mid-elevation. The middle deltoid showed a small reduction of 4.2%, with a slightly delayed steep rise in activation before reaching a comparable maximal plateau. The infraspinatus decreased by 8.6%, while still reaching near-maximal activation; the high-activation plateau was moderately shortened. The supraspinatus decreased by 26.3%, with a clearly shortened near-maximal activation plateau compared to the unassisted condition.

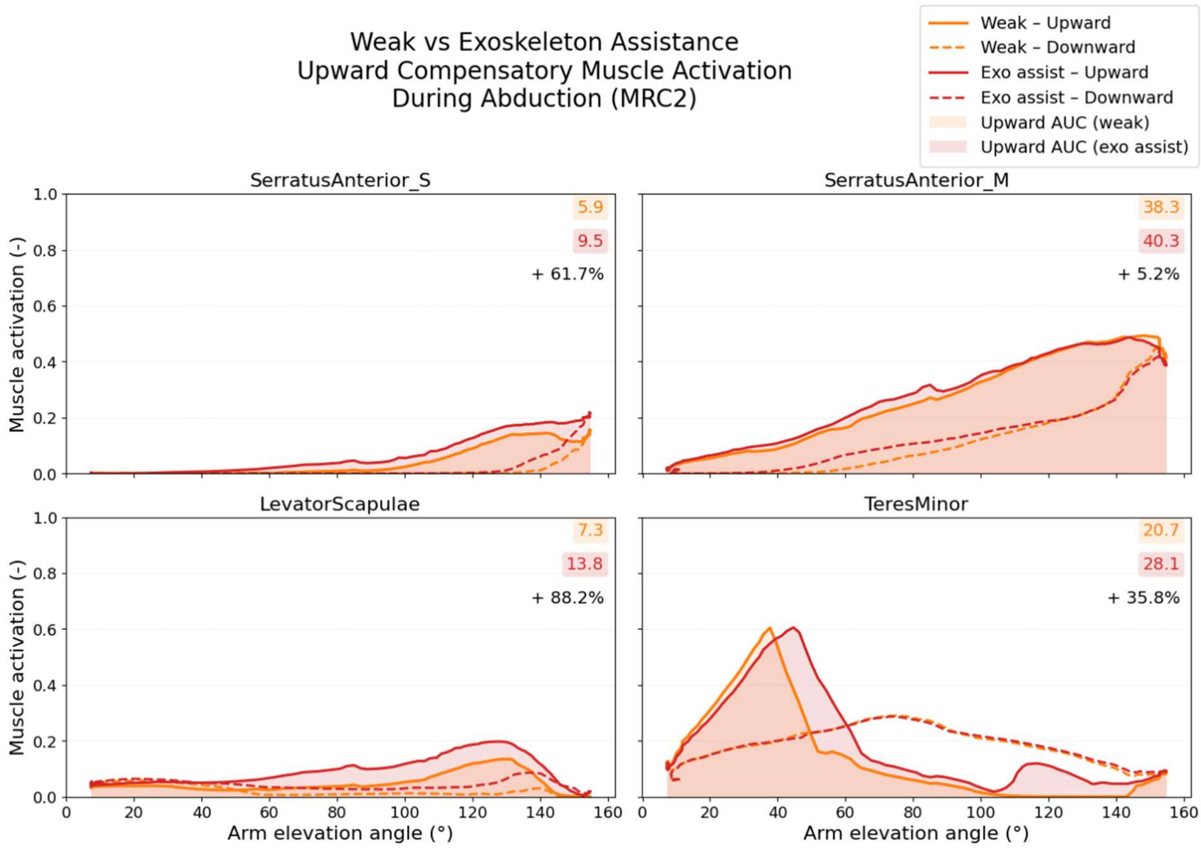

Figure 4.7: Activation-angle loops of compensatory shoulder muscles during abduction in the unassisted MRC2 condition and with exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the unassisted weakened condition, upward A_{cum} increased by +61.7%, +5.2%, +88.2%, and +35.8% for the four muscles shown, respectively.

With exoskeleton assistance, activation increased in all four shown compensatory muscles relative to the unassisted MRC2 condition.

The serratus anterior (superior part) increased by 61.7%, with consistently higher activation across mid- to high-elevation angles. The serratus anterior (middle part) showed a modest increase of 5.2%, with a largely similar activation profile but slightly higher amplitude throughout the upward phase. The levator scapulae increased by 88.2%, with visibly elevated activation from mid-elevation onward. The teres minor increased by 35.8%, showing a slightly broader early activation peak around 30-50°, followed by persistently higher activation levels and additional peak in high range compared to the unassisted condition.

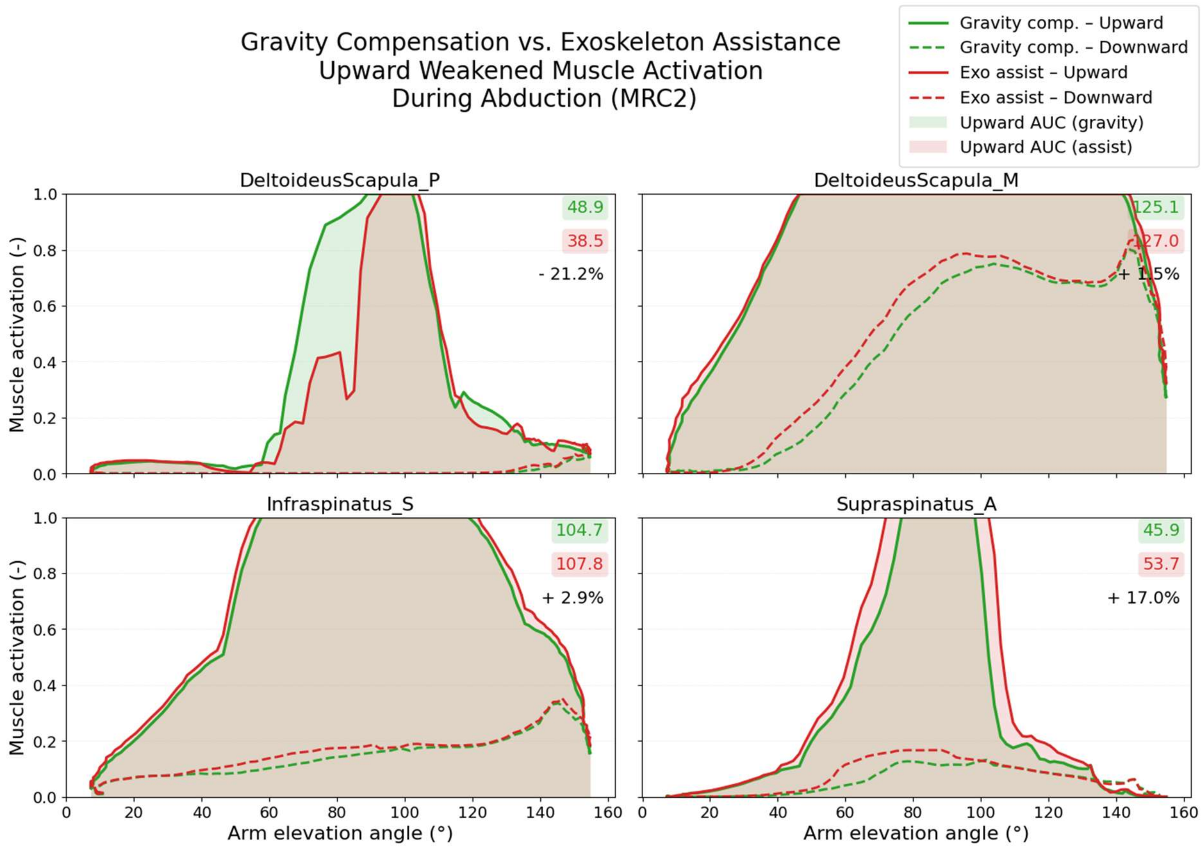

Figure 4.8: Activation-angle loops of weakened shoulder elevators during abduction in the MRC2 condition under partial gravity compensation and exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the partial gravity compensating condition, upward A_{cum} changed by -21.2%, +1.5%, +2.9%, and +17.0% for the four muscles shown, respectively.

4.2.1.5 Deviation from Ideal Gravitational Compensation

Compared to gravity compensation, exoskeleton assistance altered activation differently across the weakened elevators.

The posterior deltoid decreased by 21.2%, with a shorter high-activation plateau and a more irregular activation profile during mid-elevation under exoskeleton assistance. The middle deltoid showed a small increase of 1.5%, and the infraspinatus increased by 2.9%, both maintaining a comparable near-maximal activation plateau with minimal differences in duration. The supraspinatus increased by 17.0%, with a visibly longer and slightly higher near-maximal activation plateau under exoskeleton assistance compared to gravity compensation.

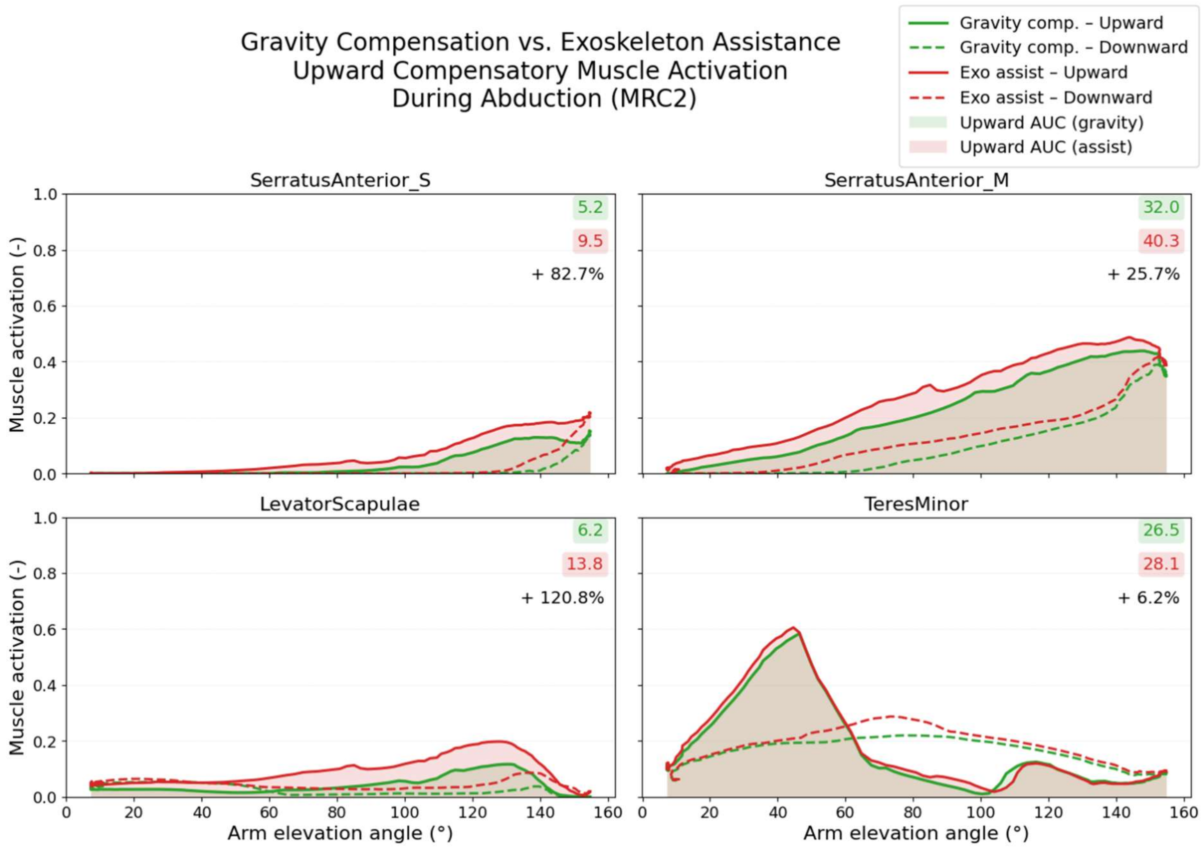

Figure 4.9: Activation-angle loops of compensatory shoulder muscles during abduction in the MRC2 condition under partial gravity compensation and exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the partial gravity compensating condition, upward A_{cum} increased by +82.7%, +25.7%, +120.8%, and +6.2% for the four muscles shown, respectively.

Compared to gravity compensation, exoskeleton assistance increased activation in all four shown compensatory muscles.

The serratus anterior (superior part) increased by 82.7%, with consistently higher activation across mid- to high-elevation angles. The serratus anterior (middle part) increased by 25.7%, showing a similar activation pattern but higher amplitude throughout the upward phase. The levator scapulae increased by 120.8%, with visibly elevated activation from mid-elevation onward. The teres minor increased by 6.2%, with a comparable early peak but slightly higher activation maintained into late elevation

4.2.2 Shoulder Forward Flexion

Secondly, the effects of shoulder exoskeleton assistance during arm elevation were evaluated for shoulder forward flexion.

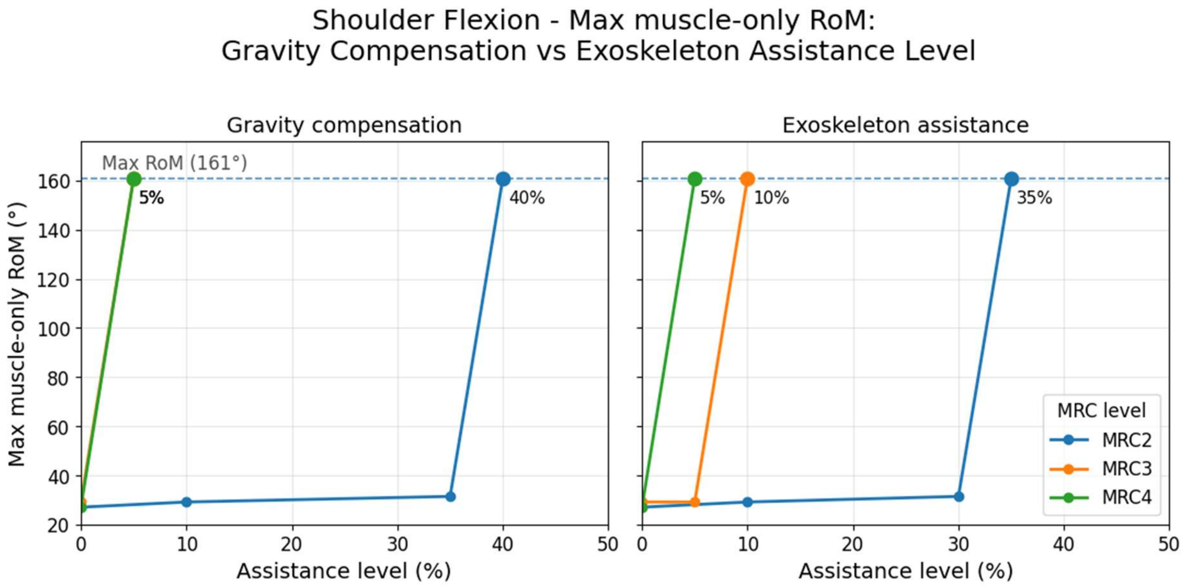

Figure 4.10: Maximal active range of motion during forward flexion across MRC levels and assistance conditions. The figure shows the maximal humerothoracic elevation angle achieved without biomechanically relevant reserve actuator contribution for healthy, weakened, gravity-compensated, and exoskeleton-assisted conditions.

4.2.2.1 Functional Performance: Active Range of Motion

In the healthy reference condition, full shoulder forward flexion was achieved without help of reserve actuators under all tested loading conditions.

In contrast, all weakened conditions demonstrated a reduced active range of motion, with decreasing maximal elevation as muscle weakness and external load increased (Table 4.4). In the MRC2 condition, forward flexion was only feasible under gravity disabled conditions. Under normal gravity, loss of muscle-only feasibility occurred early in the movement, with reserve actuator detection at humerothoracic elevations of approximately 29° and 27°, respectively.

For MRC3 and MRC4, higher maximal elevations were achieved without assistance. However, when external hand loads were applied, reserve actuator detection occurred at progressively lower elevation angles.

In addition to the restoration of full forward flexion, increasing assistance resulted in a gradual increase in maximal active range of motion across all MRC levels (Figure 4.10). The assistance level required to achieve full range of motion differed between gravity compensation and exoskeleton assistance.

MRC level	*Condition*	*Assistance level*	*Max aRoM*
MRC2	No assistance	-	27.0
	Gravity compensation	40	Full aRoM
	Exoskeleton assistance	35	Full aRoM
MRC3	No assistance (1kg load)	-	29.1
	Gravity compensation	5	Full aRoM
	Exoskeleton assistance	10	Full aRoM
MRC4	No assistance (3kg load)	-	27.01
	Gravity compensation	5	Full aRoM
	Exoskeleton assistance	5	Full aRoM

Table 4.4: Assistance level required to achieve full aRoM during forward flexion. Full aRoM indicates completion of the prescribed abduction trajectory without biomechanically relevant reserve actuator contribution.

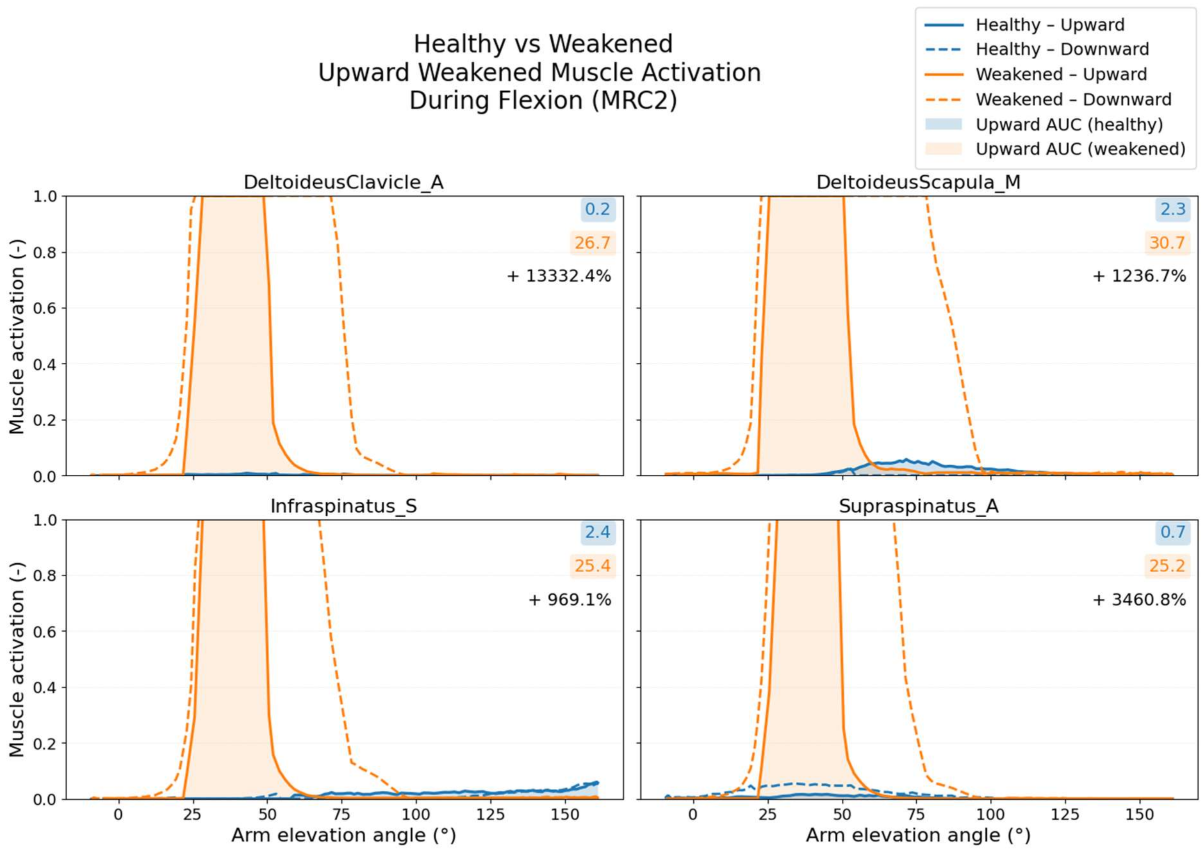

Figure 4.11: Activation-angle loops of weakened shoulder elevators during forward flexion in the healthy and weakened MRC2 conditions. Shaded areas denote the upward-phase A_{cum}. Relative to healthy, upward A_{cum} increased by +5133332.4%, +1236.7%, +969.1%, and +3460.8% for the four muscles shown, respectively.

4.2.2.2 The Effect of Weakness

In the MRC2 condition, all four shown weakened flexors also showed substantially increased activation compared to healthy for the forward flexion motion.

The anterior deltoid increased by 13,332.4%, shifting from negligible activation in the healthy condition to a sharp, early peak around 30-40°, followed by an abrupt decline after mid-elevation. The middle deltoid increased by 1,236.7%, reaching near-maximal activation early in the upward phase, with a short high-activation interval before a steep drop. The infraspinatus increased by 969.1%, showing a similar early peak with a brief near-maximal plateau and rapid decline thereafter. The supraspinatus increased by 3,460.8%, with a pronounced peak between approximately 30° and 60°, followed by a sharp reduction toward higher elevation angles.

Across all four shown muscles, activation was characterized by short, high-amplitude peaks during both the upward and downward phases, with limited gradual build-up or tapering of activity compared to the healthy condition.

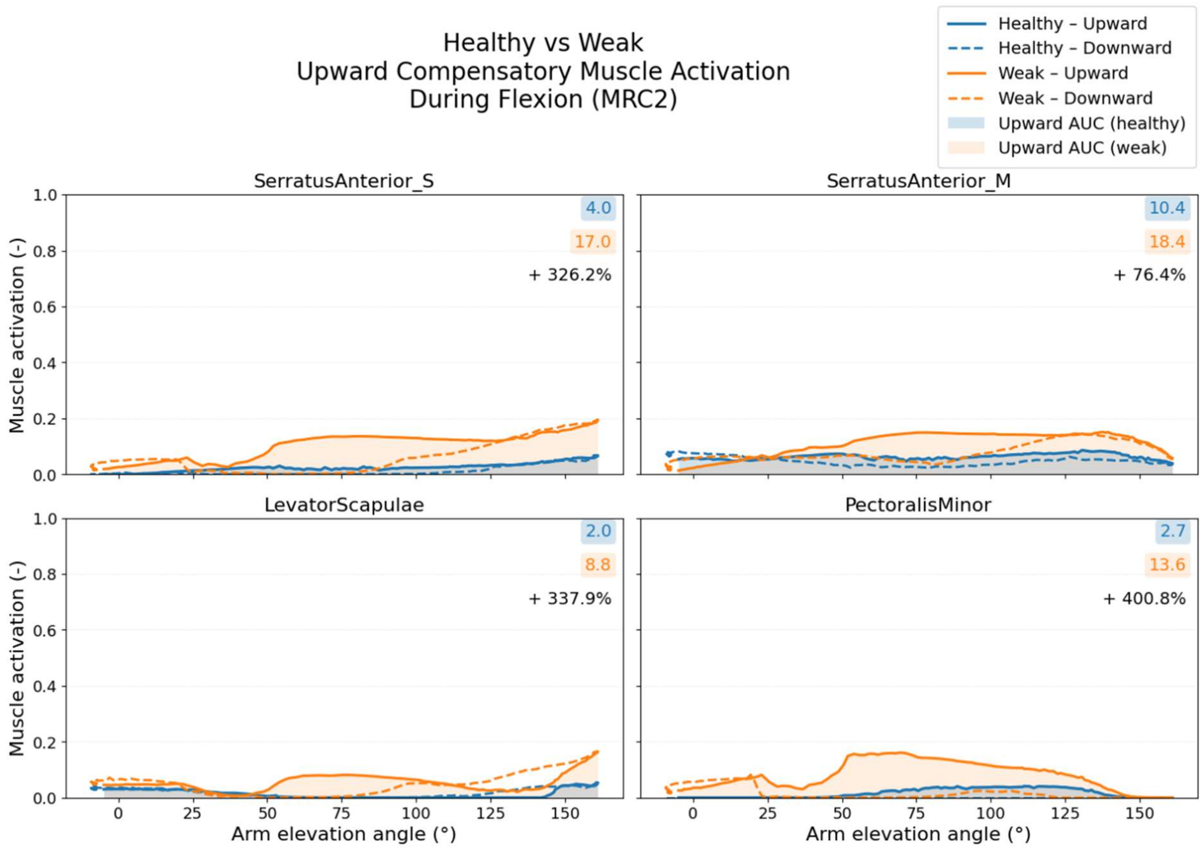

Figure 4.12: Activation-angle loops of compensatory shoulder muscles during forward flexion in the healthy and weakened MRC2 conditions. Shaded areas denote the upward-phase A_{cum}. Relative to healthy, upward A_{cum} increased by +326.2%, +76.4%, +337.9%, and +400.8% for the four muscles shown, respectively.

In the MRC2 condition, activation increased in all four shown compensatory muscles compared to healthy.

The serratus anterior (superior part) increased by 326.2%, with higher activation primarily during mid- to late elevation. The serratus anterior (middle part) increased by 76.4%, showing a modest elevation in activation while maintaining a similar overall profile. The levator scapulae increased by 337.9%, with clearly higher activation toward higher elevation angles compared to healthy. The pectoralis minor showed the largest relative increase (+400.8%), with elevated activation across mid-elevation and a distinct peak preceded by a dip around the early mid-range.

In contrast to the sharp, short activation peaks observed in the weakened prime elevators, compensatory muscles displayed lower-amplitude but more sustained activation across the elevation range.

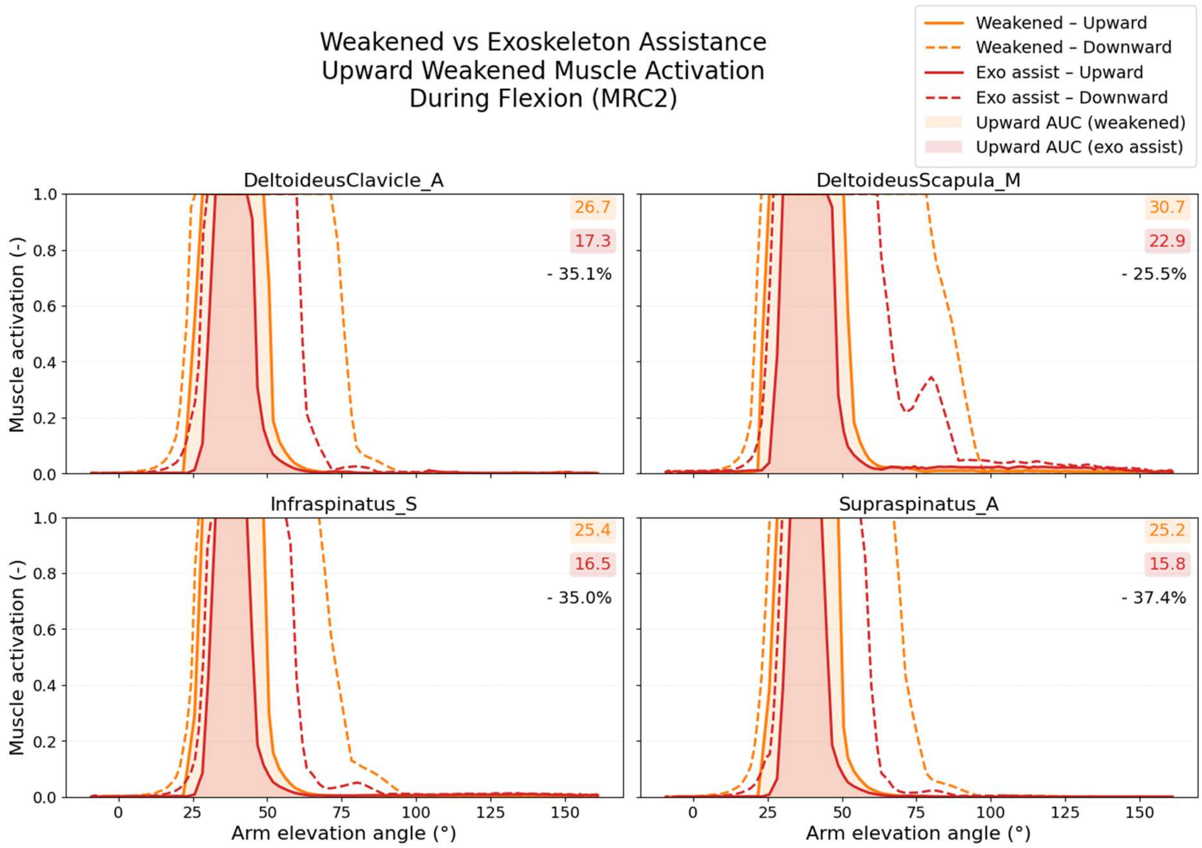

Figure 4.13: Activation-angle loops of weakened shoulder elevators during forward flexion in the unassisted MRC2 condition and with exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the unassisted weakened condition, upward A_{cum} decreased by -35.1%, -25.5%, -35.0%, and -37.4% for the four muscles shown, respectively.

4.2.2.3 The Effect of Exoskeleton Assistance

With exoskeleton assistance, activation decreased in all four weakened flexors relative to the unassisted MRC2 condition (-35.1%, -25.5%, -35.0%, and -37.4%, respectively).

Across all four muscles, the sharp, high-amplitude activation peaks observed in the weakened condition were reduced in magnitude and shortened in duration.

Across all four muscles, the sharp activation peaks characteristic of the weakened condition were uniformly reduced in magnitude. The overall activation profiles remained similar in shape under exoskeleton assistance.

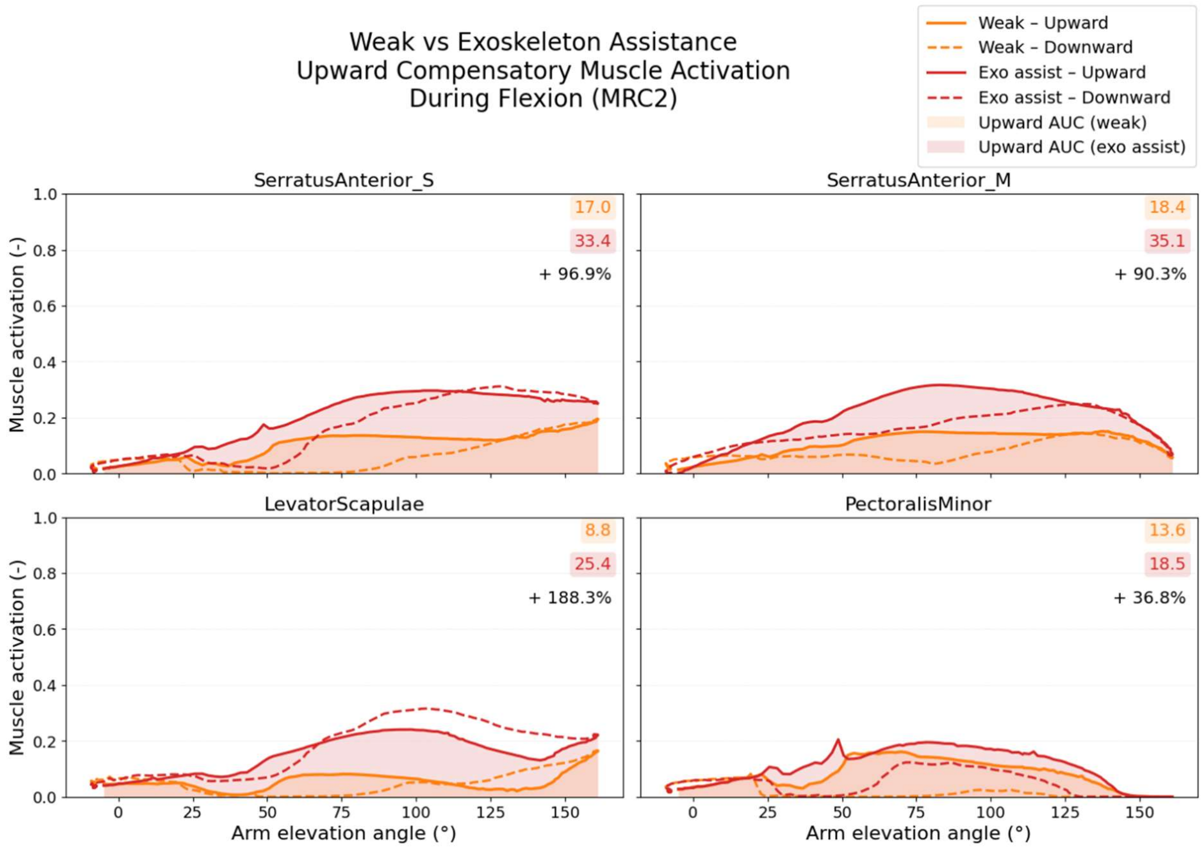

Figure 4.14: Activation-angle loops of compensatory shoulder muscles during forward flexion in the unassisted MRC2 condition and with exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the unassisted weakened condition, upward A_{cum} increased by +96.9%, +90.3%, +188.3%, and +36.8% for the four muscles shown, respectively.

With exoskeleton assistance, activation increased in all four shown compensatory muscles relative to the unassisted MRC2 condition (+96.9%, +90.3%, +188.3%, and +36.8%, respectively).

The serratus anterior (superior part) and serratus anterior (middle part) showed consistently higher activation across mid- to high-elevation angles under exoskeleton assistance. The levator scapulae demonstrated the largest relative increase (+188.3%), with visibly elevated activation throughout the mid-range of elevation. The pectoralis minor increased by 36.8%, with moderately higher activation across mid-elevation while maintaining a similar overall activation profile.

Across all four muscles, activation magnitude was higher under exoskeleton assistance, while the general activation shape remained comparable to the weakened condition.

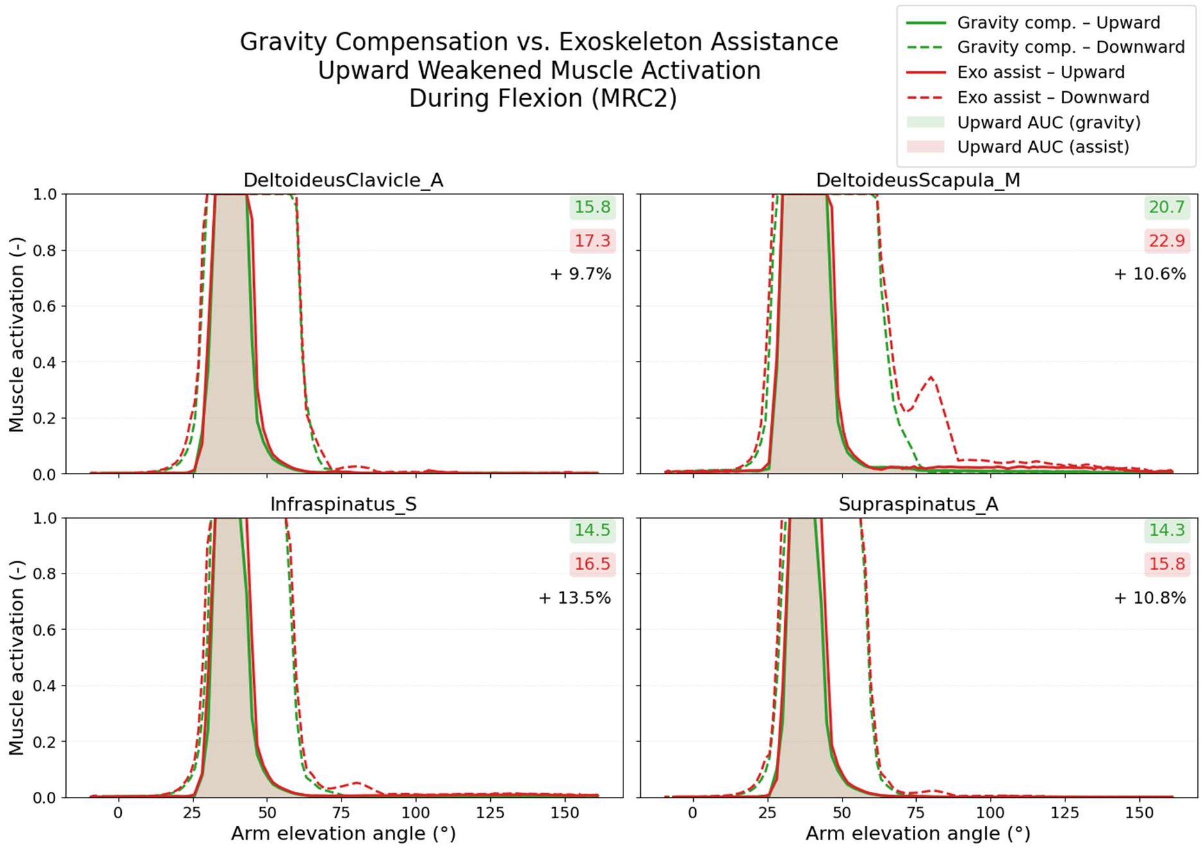

Figure 4.15: Activation-angle loops of weakened shoulder elevators during forward flexion in the MRC2 condition under partial gravity compensation and exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the partial gravity compensating condition, upward A_{cum} increased by +9.7%, +10.6%, +13.5%, and +10.8% for the four muscles shown, respectively.

4.2.2.4 Deviation from Ideal Gravitational Compensation

Compared to partial gravity compensation, exoskeleton assistance resulted in slightly higher activation in all four shown weakened elevators (+9.7%, +10.6%, +13.5%, and +10.8%, respectively).

Across all four muscles, activation profiles remained highly similar between conditions, with narrow, high-amplitude peaks at early elevation. Under exoskeleton assistance, these peaks were slightly higher and marginally prolonged. In addition, the middle deltoid showed an additional lower activation peak during the downward phase under exoskeleton assistance that was not present under gravity compensation.

The overall activation pattern remained consistent between assistance strategies, with only modest increases in activation magnitude under exoskeleton assistance.

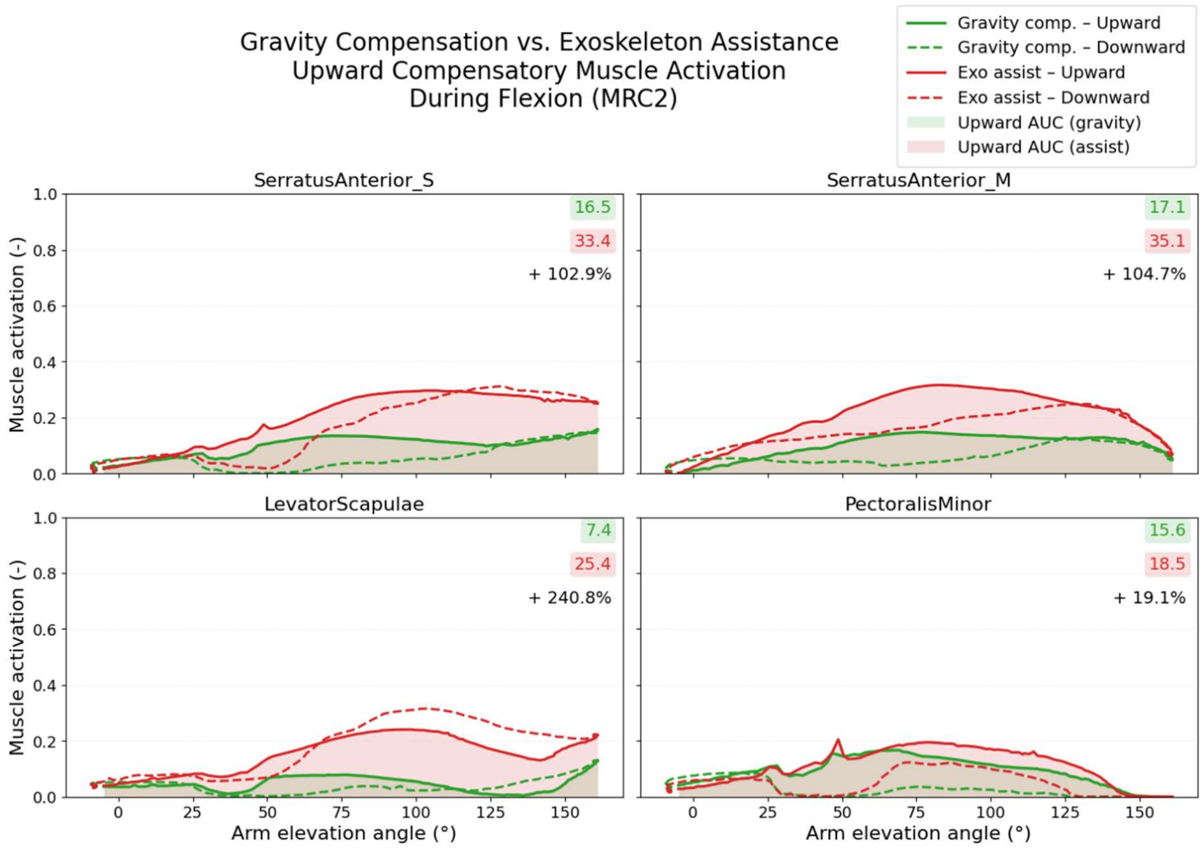

Figure 4.16: Activation-angle loops of compensatory shoulder muscles during forward flexion in the MRC2 condition under partial gravity compensation and exoskeleton assistance. Shaded areas denote the upward-phase A_{cum}. Relative to the partial gravity compensating condition, upward A_{cum} increased by +102.9%, +104.7%, +240.8%, and +19.1% for the four muscles shown, respectively.

Compared to partial gravity compensation, exoskeleton assistance resulted in higher activation in all four compensatory muscles (+102.9%, +104.7%, +240.8%, and +19.1%, respectively).

The serratus anterior (superior and middle parts) showed consistently higher activation across most of the elevation range under exoskeleton assistance. The levator scapulae demonstrated the largest relative increase (+240.8%), with more elevated activation from mid-elevation onward. The pectoralis minor increased by 19.1%, with moderately higher activation across mid-range while maintaining a similar overall activation profile.

Across all four shown muscles, activation magnitude under exoskeleton assistance exceeded that observed under gravity compensation throughout the upward phase.

4.2.3 Comparison of Shoulder Abduction and Flexion

Comparison of the assistance levels required to restore full active range of motion revealed differences between shoulder abduction and forward flexion (Table 4.5).

For shoulder abduction, the required level of compensation increased progressively with increasing MRC level for both gravity compensation and exoskeleton assistance. In contrast, shoulder forward flexion required relatively high assistance at MRC2, while lower and more comparable assistance levels were sufficient for MRC3 and MRC4. Consequently, abduction and forward flexion exhibited opposing compensation trends across MRC levels.

Movement	Assistance type	MRC2	MRC3	MRC4
Abduction	Gravity compensation	25%	30%	40%
	Exoskeleton assistance	15%	25%	45%
Forward flexion	Gravity compensation	40%	5%	5%
	Exoskeleton assistance	35%	10%	5%

Table 4.5: Comparison of assistance levels required to restore full active range of motion for shoulder abduction and forward flexion. Values represent the minimal gravity compensation or exoskeleton assistance required to complete the full prescribed movement for each MRC level.

These differences were also reflected in the activation behaviour of the weakened prime elevators.

During shoulder abduction, weakened elevators exhibited prolonged near-maximal activation plateaus, including muscle-specific features such as the mid-elevation dip in the posterior deltoid. Gravity compensation shortened these plateaus, whereas exoskeleton assistance altered activation more heterogeneously across muscles, including increased supraspinatus activation relative to gravity compensation.

In contrast, during forward flexion, weakened flexors showed short, sharp near-maximal activation peaks during both upward and downward phases rather than sustained plateaus. Both gravity compensation and exoskeleton assistance produced largely uniform reductions in activation magnitude and peak duration across all four muscles, with minimal muscle-specific deviations.

Compensatory muscle responses also differed between movements. During abduction, exoskeleton assistance increased activation selectively in specific stabilizing muscles (e.g., teres minor, levator scapulae, and portions of the serratus anterior), particularly at higher elevation angles. During forward flexion, increases under exoskeleton assistance were more consistent across compensatory muscles, including the serratus anterior, levator scapulae, and pectoralis minor.

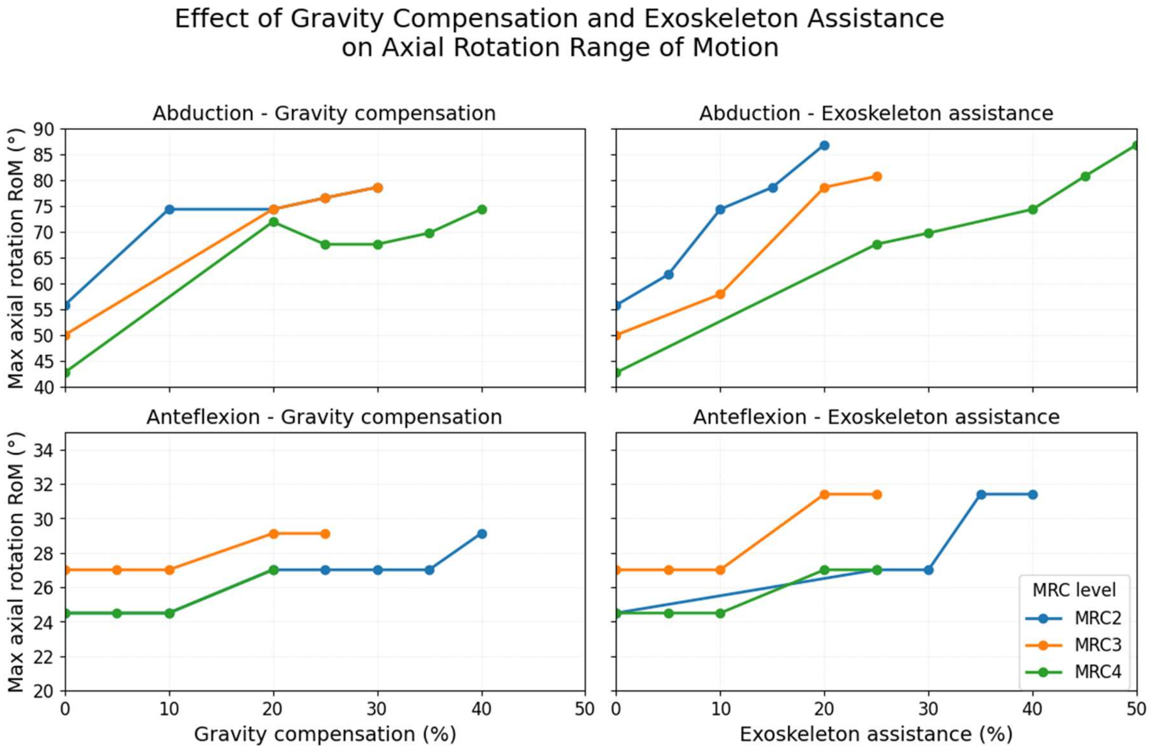

Figure 4.17: Maximal detected axial rotation range of motion as a function of assistance level during shoulder abduction and forward flexion. Axial rotation detection consistently preceded detection in the primary elevation coordinates.

4.2.4 Axial Rotation

Axial rotation was evaluated as a additional degree of freedom during both shoulder abduction and forward flexion. Although axial rotation was not predefined as a primary outcome measure, consistent and systematic patterns were observed across simulations. Given the potential relevance of axial rotation to overall shoulder function and compensatory mechanisms, these findings are reported here for completeness. Across all simulations, reserve actuator detection consistently occurred first in the axial rotation coordinate, preceding detection in the shoulder elevation and plane-of-elevation coordinates.

For shoulder abduction, increasing levels of gravity compensation and exoskeleton assistance were associated with a clear and progressive increase in the maximal detected axial rotation range of motion across all MRC levels (see figure 4.17). In externally loaded and unassisted conditions, axial rotation detection occurred at relatively low angles, while higher assistance levels yielded stepwise increases in maximal axial rotation. This pattern was consistently observed for MRC2, MRC3, and MRC4 and occurred regardless of detection in the primary elevation coordinates.

During forward flexion, maximal axial rotation ranges were smaller than during abduction. Nevertheless, increasing gravity compensation and exoskeleton assistance were associated with gradual increases in maximal axial rotation across all MRC levels. These increases were more limited in magnitude but followed a consistent upward trend with increasing assistance.

5 Discussion

5.1 Answering the Central Research Question

The central research question of this study was whether the Asgari shoulder exoskeleton is suitable for daily use in individuals with shoulder muscle weakness. The underlying hypothesis stated that passive gravity-compensating assistance would be biomechanically effective in restoring elevation capacity under graded weakness while reducing neuromuscular activation demand, and that such assistance would align with user-relevant needs in adults with Erb's palsy.

When integrating the biomechanical simulations with the qualitative findings, the results largely support this hypothesis. In this context, suitability is interpreted within the boundaries of biomechanical simulation and qualitative inquiry, rather than as demonstrated clinical effectiveness in real-world use.

From a biomechanical perspective, two primary unmet needs were operationalized in advance: first, limited active range of motion against gravity, and second, excessive relative neuromuscular activation demand of the remaining shoulder musculature causing fatigue. These unmet needs were not defined abstractly but translated into measurable parameters. Elevation capacity was quantified as maximal active range of motion, with loss of feasibility detected using predefined reserve actuator thresholds reflecting non-physiological torque supplementation. Neuromuscular activation overload and sustained demand was quantified using peak muscle activation and cumulative activation (area under the activation-angle curve).

Under graded reductions in maximal muscle strength corresponding to clinically interpretable MRC levels, weakened models frequently failed to complete the prescribed elevation within physiological limits. This confirms that, even when kinematics are preserved, reduced force-generating capacity alone is sufficient to create a functional mechanical bottleneck during elevation against gravity. In that sense, the simulations mechanistically reproduced the first unmet need: insufficient active elevation capacity.

Passive assistance restored full muscle-driven elevation once a minimal compensation threshold was reached. Assistance remained partial rather than fully gravity-balancing, indicating that the device functions as a strength amplifier rather than a motion-imposing mechanism. Moreover, assistance consistently reduced the duration of peak activation levels, although maximal activation (1.0) was still reached in the more severe weakness conditions, and reduced cumulative activation demand of the weakened elevators. These reductions directly address the second unmet need, namely operation near maximal capacity and sustained neuromuscular overload.

The predefined biomechanical success criteria were largely met: full restoration of muscle-driven elevation, reduction in the duration of peak and cumulative activation relative to the weakened condition, and absence of disproportionate (> twofold) increases in other muscle activations. However, maximal activation levels (1.0) were still reached, indicating that neuromuscular demand was reduced but not fully normalized. Although redistribution of activation toward stabilizing musculature was observed (particularly during abduction) these changes remained within physiological limits and reflected altered joint moment balance rather than pathological compensation.

When these findings are interpreted alongside the qualitative data, a clear convergence emerges. Participants consistently described elevation against gravity as a central functional bottleneck, accompanied by rapid fatigue, reliance on compensatory thorax strategies, and the need to spread effort throughout the day. The biomechanical parameters used in this study provide a mechanistic translation of these experiences. Peak activation reflects operating close to maximal available strength, while cumulative activation reflects sustained neuromuscular demand that plausibly contributes to fatigue-like sensations during repetitive or prolonged tasks (87).

Taken together, the results indicate that passive gravity-compensating shoulder assistance is biomechanically feasible under graded weakness and directly targets the central mechanical bottleneck identified by users. Within the defined criteria of this study, the exoskeleton concept can therefore be considered suitable in principle for daily contexts dominated by gravitational loading of the shoulder. While mechanical feasibility does not equate to improved task performance, restoration of elevation capacity represents a necessary precondition for functional participation in overhead and shoulder-level activities.

5.2 Question 1: Potential User Perspectives and Design Implications

The first question focused on experienced daily limitations, compensatory strategies, and perceptions regarding exoskeleton use in adults with Erb's palsy.

Beyond confirming elevation against gravity as a core limitation, the qualitative findings uniquely emphasize that suitability is multidimensional. Participants did not merely express a desire for increased strength, rather, they described a complex interplay between physical fatigue, compensatory strategies, psychosocial burden, and energy management across the day. Shoulder-related limitations were described as persistent and structurally integrated into daily routines, with compensatory trunk movements, momentum strategies, and reliance on the contralateral arm forming established adaptation patterns.

Importantly, participants emphasized selective and context-dependent use. Assistance was envisioned for work-related tasks, hobbies, or prolonged elevation demands, rather than continuous wear. This insight is highly relevant when interpreted alongside the biomechanical findings. Since assistance primarily reduces activation demand during elevation against gravity, its functional benefit is inherently task-specific. Continuous support during all movements is neither mechanically necessary nor user-desired.

Comfort, weight, adjustability, and discretion were consistently identified as essential design requirements. These requirements reflect an additional unmet need that is not captured by biomechanical parameters alone: the device must integrate into daily life without introducing new physical or psychosocial burdens. In this sense, biomechanical feasibility constitutes a necessary but insufficient condition for real-world suitability.

The convergence between both domains lies in the identification of sustained elevation against gravity as the primary bottleneck. The divergence lies in the broader criteria of acceptance, which extend beyond force and activation metrics to include comfort, autonomy, and discretion.

5.3 Question 2: Feasibility of Biomechanical Compensation

The second question addressed whether passive assistance can compensate graded MRC-defined weakness by restoring functional elevation capacity and reducing relative neuromuscular activation demand without introducing adverse compensatory activation patterns.

The magnitude of assistance required differed across MRC levels. While more severe weakness required higher compensation thresholds to restore muscle-driven elevation, moderate weakness primarily manifested as elevated activation demand rather than complete loss of feasibility. This suggests that passive assistance may serve both as a restorative strategy for active range of motion and a load-reducing strategy depending on strength level.

The simulated MRC levels were calibrated to clinically meaningful thresholds, reflecting the ability to complete elevation under gravity-eliminated, gravity-only, or externally loaded conditions. Only maximal muscle force capacity was altered, while coordination, tendon properties, and kinematics were preserved. This ensured that observed differences directly reflected changes in strength and assistance rather than altered motor control strategies.

The present simulations evaluate the mechanical sufficiency of gravity compensation under the assumption that reduced torque capacity is the primary limiting factor. Neural control strategies and potential alterations in motor coordination were not explicitly modelled. The model therefore evaluates mechanical sufficiency under the assumption that strength is the primary limiting factor, whereas real-world motor adaptations may introduce additional complexity (53, 88).

Under these controlled conditions, weakness led to distinct neuromuscular activation overload. During abduction, weakened elevators exhibited prolonged near-maximal activation plateaus, indicating sustained operation at the upper bound of available capacity. During forward flexion, activation patterns were characterized by sharp peaks during specific phases of the movement rather than extended plateaus. These movement-specific overload patterns demonstrate that the mechanical consequences of weakness are not uniform across elevation directions.

Passive assistance reduced neuromuscular activation overload in both movements, albeit with different redistribution patterns. In abduction, assistance shortened high-activation plateaus and altered the balance between prime movers and stabilizers. In forward flexion, reductions were more uniform across muscles. Crucially, no condition exceeded the predefined threshold for disproportionate activation of non-weakened muscles. Glenohumeral stability constraints remained satisfied without excessive stabilizer overactivation, indicating that assistance did not induce adverse biomechanical trade-offs within the modelled framework.

Although both gravity scaling and exoskeleton-based assistance restored elevation capacity, their redistribution patterns differed, indicating that mechanical implementation influences neuromuscular recruitment beyond simple gravitational unloading. These differences underscore the importance of optimizing attachment geometry and force transmission to refine assistive effects in future designs.

Furthermore, assistance influenced axial rotation feasibility, suggesting that gravitational compensation may indirectly affect functional workspace beyond the primary elevation degree of freedom. Although axial rotation was not a predefined primary outcome, its systematic behaviour across conditions indicates that restoring elevation torque capacity may also reduce secondary kinematic constraints.

Thus, when evaluated against the explicit methodological criteria defined a priori (showing restoration of muscle-driven elevation, reduction in peak and cumulative activation, and absence of pathological redistribution) passive assistance fulfilled the biomechanical requirements across graded weakness levels.

5.4 Limitations and Future Research

This study has several methodological limitations that also define directions for future research. First, simulations were performed using prescribed healthy humerothoracic kinematics, assuming that sufficient gravitational support would enable individuals with shoulder muscle weakness to approximate non-compensatory movement patterns. In practice, long-standing neuromuscular adaptations, altered scapulothoracic coordination, structural joint changes, and habitual compensatory strategies may limit restoration of physiological kinematics, even with assistance. Future work should therefore incorporate patient-specific motion data to evaluate whether exoskeleton support promotes more physiological movement patterns or primarily augments existing compensations.

Muscle weakness was represented as a controlled reduction in maximal muscle force of selected shoulder elevators, without modifying activation dynamics, tendon properties, coordination strategies, or muscle morphology. Although this approach enabled systematic comparison across MRC-calibrated conditions, it does not fully capture the complexity of clinical neuromuscular impairment. Future studies should explore more comprehensive weakness models, including altered activation patterns, and validate model parameters against experimental strength and EMG measurements.

The exoskeleton was implemented as a force-generating PathActuator without inclusion of device mass or inertial properties. As a result, additional compressive forces, altered scapular loading, and the gravitational effect of the device itself were not considered. Incorporating segmental mass and inertia into future simulations would allow more realistic assessment of net biomechanical benefit. In addition, actuator routing and attachment locations were approximated rather than optimised. Because attachment geometry directly affects moment arms and force transmission, systematic optimisation of attachment sites and cam profiles is warranted to improve mechanical efficiency, user comfort, and neuromuscular load distribution.

Movement analysis was restricted to shoulder abduction and forward flexion under controlled conditions, with constrained elbow motion and limited task variability. Clinically relevant multi-joint activities, such as hand-to-mouth movements, were not explicitly simulated. Future work should therefore investigate coupled shoulder-elbow dynamics and task-specific functional movements to better reflect real-world activity demands. Furthermore, assistance levels were defined based on restoration of muscle-driven feasibility rather than optimisation of muscle engagement. Additional analyses are needed to determine support ranges that reduce overload without promoting excessive unloading or potential disuse.

Although axial rotation was analysed as a secondary outcome and showed systematic changes with assistance, it was not a predefined primary objective. Future research should further investigate the coupled relationship between elevation and axial rotation to determine how gravity-compensating support influences overall functional workspace and rotational capacity in patient-specific movement patterns.

Finally, all findings are based on musculoskeletal simulations and have not yet been experimentally validated in the target population. While the modelling framework enforces joint stability and penalises non-physiological actuator use, simulation outcomes remain model-based predictions. Controlled laboratory testing with prototype devices in adults with Erb's palsy is required to assess real muscle activation patterns, comfort, usability, task-specific application, and long-term acceptability.

In addition, the interview sample included a relatively high proportion of individuals with bilateral involvement. Because bilateral impairment may influence compensatory strategies and perceived support needs differently from unilateral weakness, the reported user perspectives may not fully represent the broader Erb's palsy population. Future studies should aim for more balanced sampling with respect to laterality to improve generalisability. Despite these limitations, the present work provides a systematic evaluation of predefined biomechanical feasibility criteria and represents a necessary step toward translational and clinical implementation.

5.5 Conclusions

This mixed method study examined whether passive gravity-compensating shoulder assistance is suitable for adults with Erb's palsy by integrating qualitative insights into daily activity limitations with biomechanical simulations under graded muscle weakness.

Elevation against gravity emerged as the central functional bottleneck in daily life. Participants described restricted working range and fatigue during arm elevation tasks, while biomechanical simulations demonstrated that reduced muscle strength limits muscle-driven elevation capacity and increases relative neuromuscular activation demand. Partial gravity compensation restored elevation feasibility in more severe weakness and reduced activation demand across strength levels without inducing disproportionate compensatory activation.

For individuals operating near the threshold of functional elevation against gravity, assistance may restore elevation capacity and expand functional workspace, whereas in more moderate weakness it primarily reduces relative neuromuscular activation overload. These findings indicate that passive shoulder exoskeleton support is biomechanically feasible and conceptually aligned with user-identified needs.

In conclusion, targeted and adjustable gravity compensation addressing elevation against gravity represents a promising direction for future translational and clinical development in adults with shoulder muscle weakness, provided that its implementation is not guided solely by the aim to increase strength or active range of motion, but is explicitly informed by patient perspectives on fatigue, autonomy, usability, and meaningful participation in daily life.

6 AI Disclosure

Generative artificial intelligence tools were employed in a limited capacity to assist with language refinement, writing advice, and the preparation of schematic figures. AI was not used in the conception of the research questions, experimental design, data analysis, interpretation, or the construction of scientific arguments. All scientific content and conclusions were independently developed by the author.

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Appendix A.1 Questionnaire

A. Algemene achtergrond / Letsel

1. Wat is uw leeftijd?

2. Welke arm is of welke armen zijn aangedaan door de Erbse parese?

3. Bent u hierdoor gedwongen anderhandigs geworden, denkt u?

4. Weet u welke zenuwwortels bij u zijn aangedaan, welke zijn dit?

5. Heeft u ooit een of meerdere operaties gehad, welke?

B. Restfunctie

6. Kunt u uw hand naar uw mond brengen? (Uitbeeldend)

7. Kunt u uw hand op uw hoofd leggen? (Uitbeeldend)

8. Hoe ver kunt u uw arm zijwaarts optillen? (Uitbeeldend)

9. Hoe ver kunt u uw arm voorwaarts optillen? (Uitbeeldend)

10. Hoe ver kunt u uw arm zijwaarts optillen als u op uw rug ligt?

11. Hoe ver kunt u uw arm voorwaarts optillen als u op uw zij ligt?

C. ADL en compensatie

12. Welke dagelijkse handelingen vindt u lastig door uw schouderfunctie, waarom?

13. En op wat voor manier heeft daar last van?

dit kan zowel kracht te kort zijn, of pijn, of gebrek aan goede motoriek

14. Hoe lost u dat nu meestal op?

15. Als u één handeling zou mogen verbeteren, welke zou dat zijn, en hoe ziet u dat voor zich?

D. Ervaring, verwachting en uitleg exoskelet

16. Heeft u eerder armhulpmiddelen/otheses/braces gebruikt? Wat werkte wel of niet?

17. Voor ik het exoskelet aan u wil uitleggen, ben ik benieuwd wat u zelf als exoskelet voor u ziet, wat voor u meerwaarde zou hebben?

Uitleg exoskelet

18. Is deze uitleg duidelijk voor u?

E. Eerste reactie & geschiktheid

19. Wat is uw eerste indruk? Denkt u dat dit u zou kunnen helpen?

20. Voor welke bewegingen of activiteiten zou dit nuttig zijn?

21. Zijn er activiteiten waarbij u het juist niet zou gebruiken?

22. Hoeveel ondersteuning denkt u zelf nodig te hebben?

F. Comfort & draagbaarheid

23. Hoe verwacht u dat het voelt om een hulpmiddel met bandjes om uw schouder, bovenarm en rug te dragen?

24. Heeft u normaal snel last van drukpunten, schuren of bandjes, of juist gevoelloze plekken?

25. Welke dingen zou u prettig vinden aan zo'n hulpmiddel?

26. Wat zou u absoluut niet prettig vinden aan zo'n hulpmiddel?

27. Hoe belangrijk vindt u het dat uw beweging er natuurlijk uitziet wanneer u dit zou gebruiken? En of het ontwerp onopvallend is?

G. Gebruikscenarios & verbeterpunten

28. Wanneer zou u dit hulpmiddel wel dragen en waarom?

29. Wanneer juist niet en waarom niet?

30. Wat zou er moeten worden aangepast of verbeterd zodat het voor u geschikt zou zijn voor dagelijks gebruik?

31. Wat zijn andere eisen om dit te dragen?

32. Stel ik geef u nu een kant en klare versie van dit exoskelet mee, zou u deze dan gaan dragen in het dagelijks leven? Denkt u dat de voordelen groter zijn dan de nadelen? Waarom wel of niet?

Appendix A.2 Given description of the Asgari shoulder exoskeleton during the interviews and accompanying images.

The text below is an example of an explanation given during an interview, based on a transcript. The explanations were not exactly the same in all interviews.

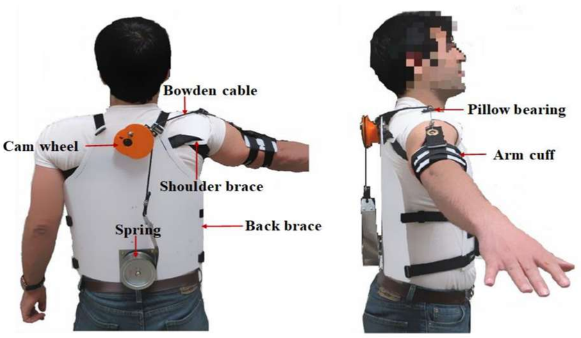
"Op de dia is het exoskelet afgebeeld. Linksboven zijn twee mannen te zien die het oude exoskelet dragen; dit model is inmiddels verouderd. Rechtsonder staat een nieuwere versie van het exoskelet. Deze uitvoering is compacter en vermoedelijk comfortabeler in gebruik.

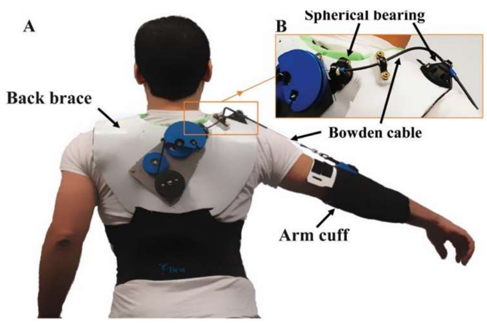
Het betreft een passief exoskelet, wat betekent dat er geen motoren in zijn verwerkt. De ondersteuning wordt volledig geleverd door een veermechanisme. De veer is bevestigd ter hoogte van de rug, zoals linksboven op de afbeelding te zien is. Vanuit deze veer loopt een kabel naar een camwiel. Dit camwiel bepaalt, afhankelijk van de stand van de arm, hoeveel kracht van de veer wordt overgebracht. Vanuit het camwiel loopt vervolgens een tweede kabel naar de bevestiging op de arm.

Het exoskelet levert continu een ondersteuning van ongeveer 20 tot 30% van de kracht die nodig is om de arm te heffen. Het systeem balanceert de arm dus niet volledig in de lucht, maar neemt wel een deel van de benodigde spierkracht weg. Hierdoor worden de spieren ontlast tijdens het uitvoeren van bewegingen.

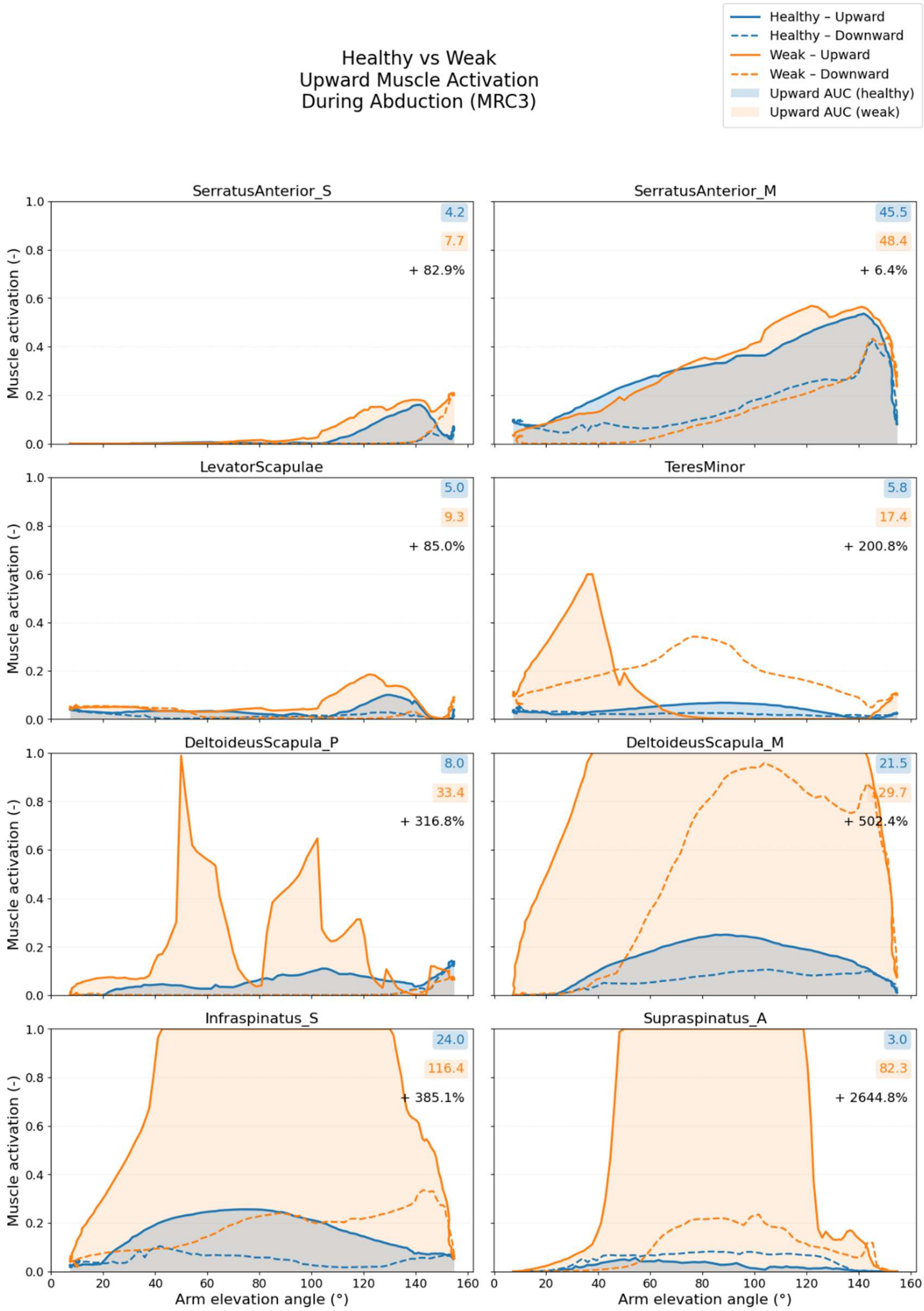
Het exoskelet biedt ondersteuning bij zijwaarts en voorwaarts heffen van de arm tot ongeveer 90 graden, oftewel tot schouderhoogte.

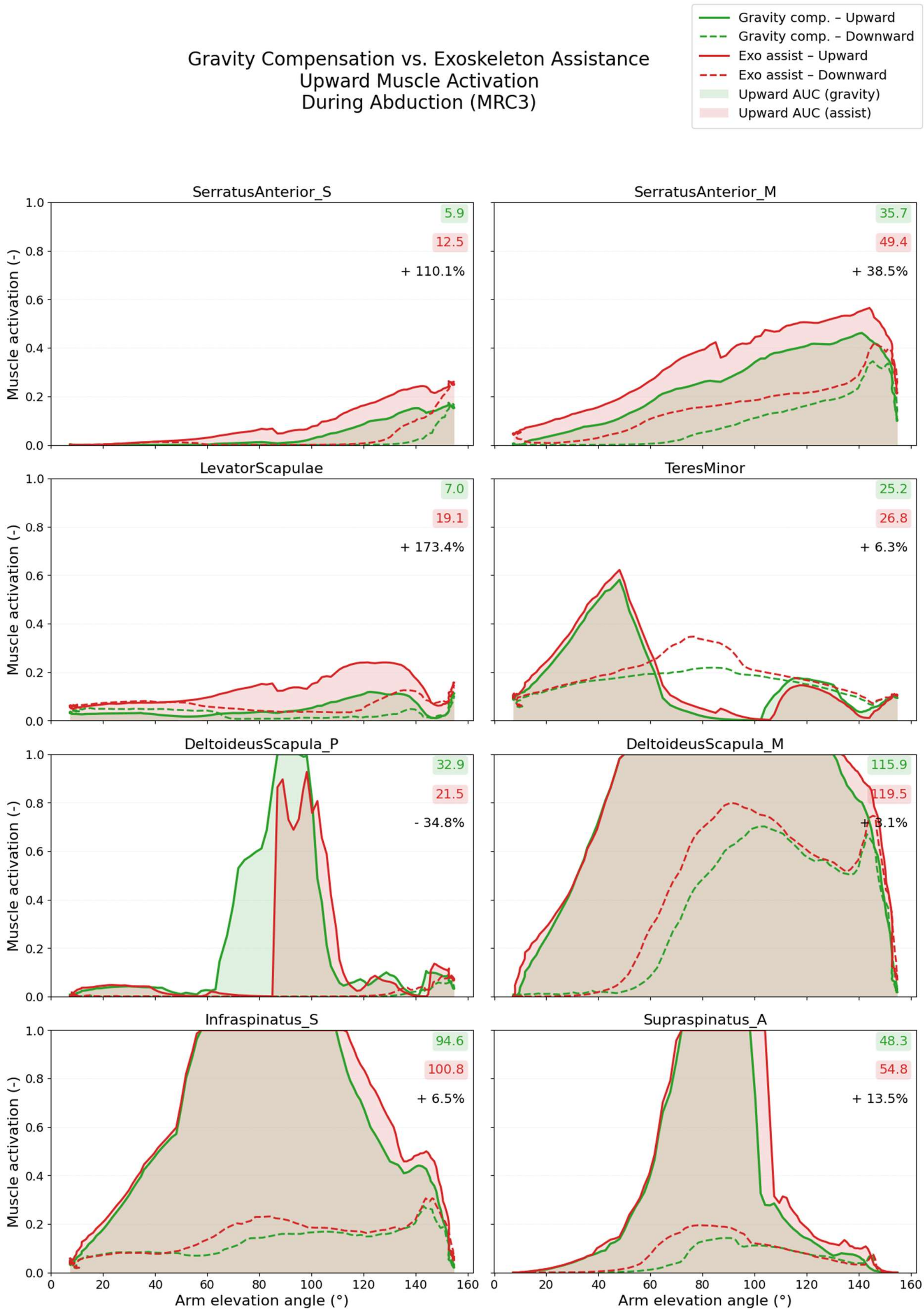
Tot nu toe is het exoskelet getest bij gezonde proefpersonen. Uit deze testen is gebleken dat een ondersteuning van 20 tot 30% als het meest prettig wordt ervaren binnen dit ontwerp. Bij deze mate van ondersteuning bleef het exoskelet comfortabel in gebruik en werd tegelijkertijd een significante afname van spieractiviteit waargenomen."



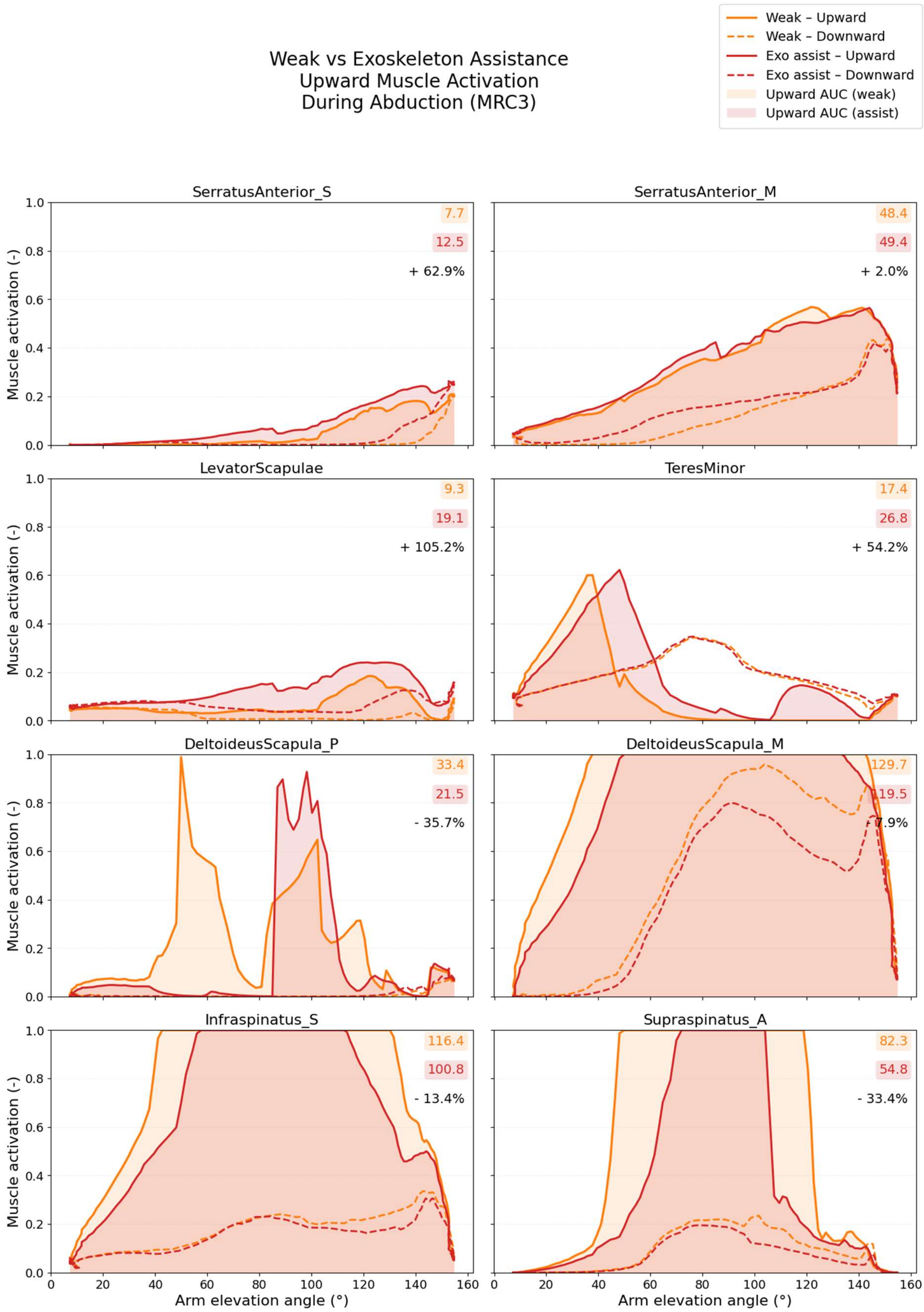


Appendix B.1: MRC3 Abduction Muscle Activation

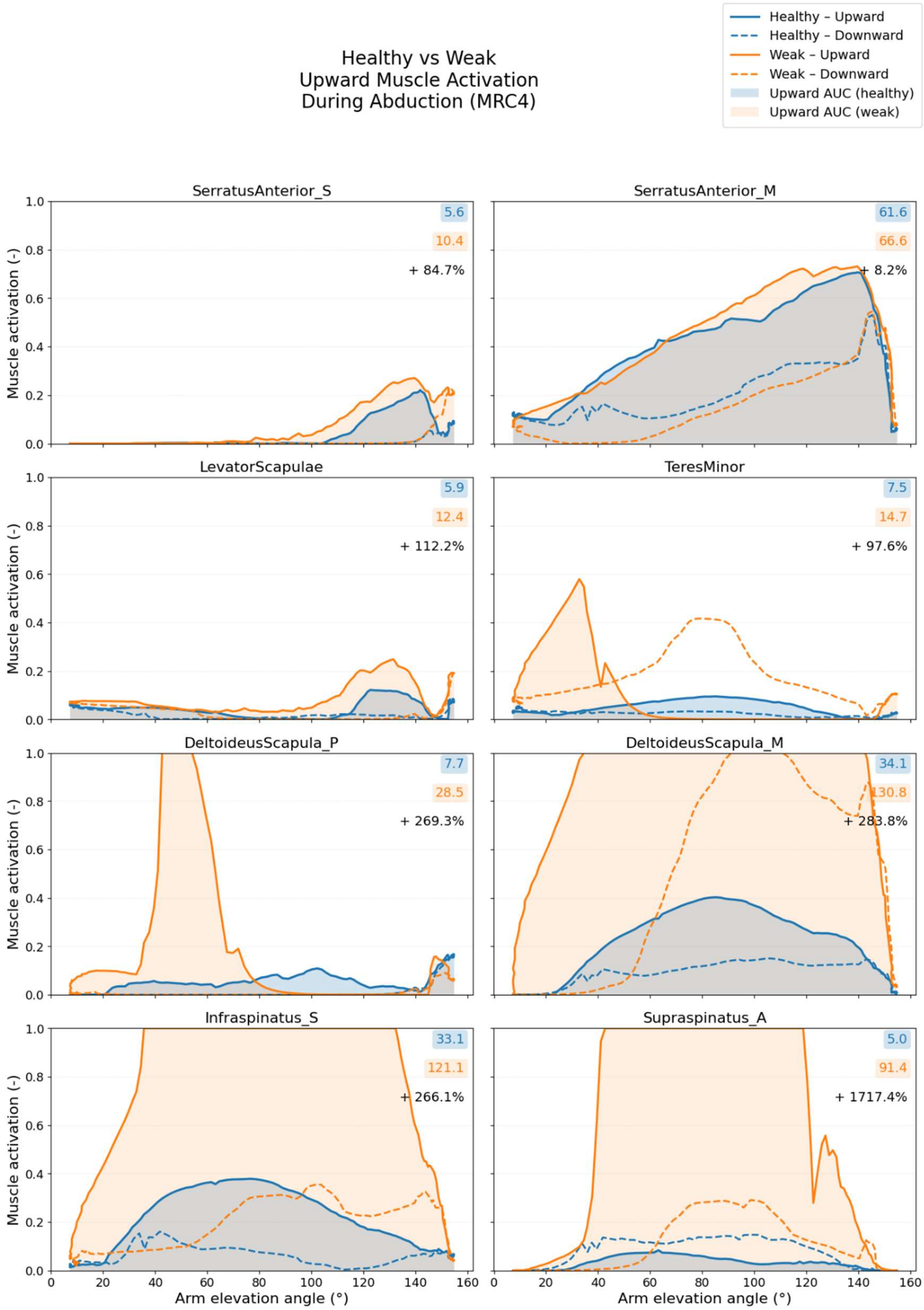


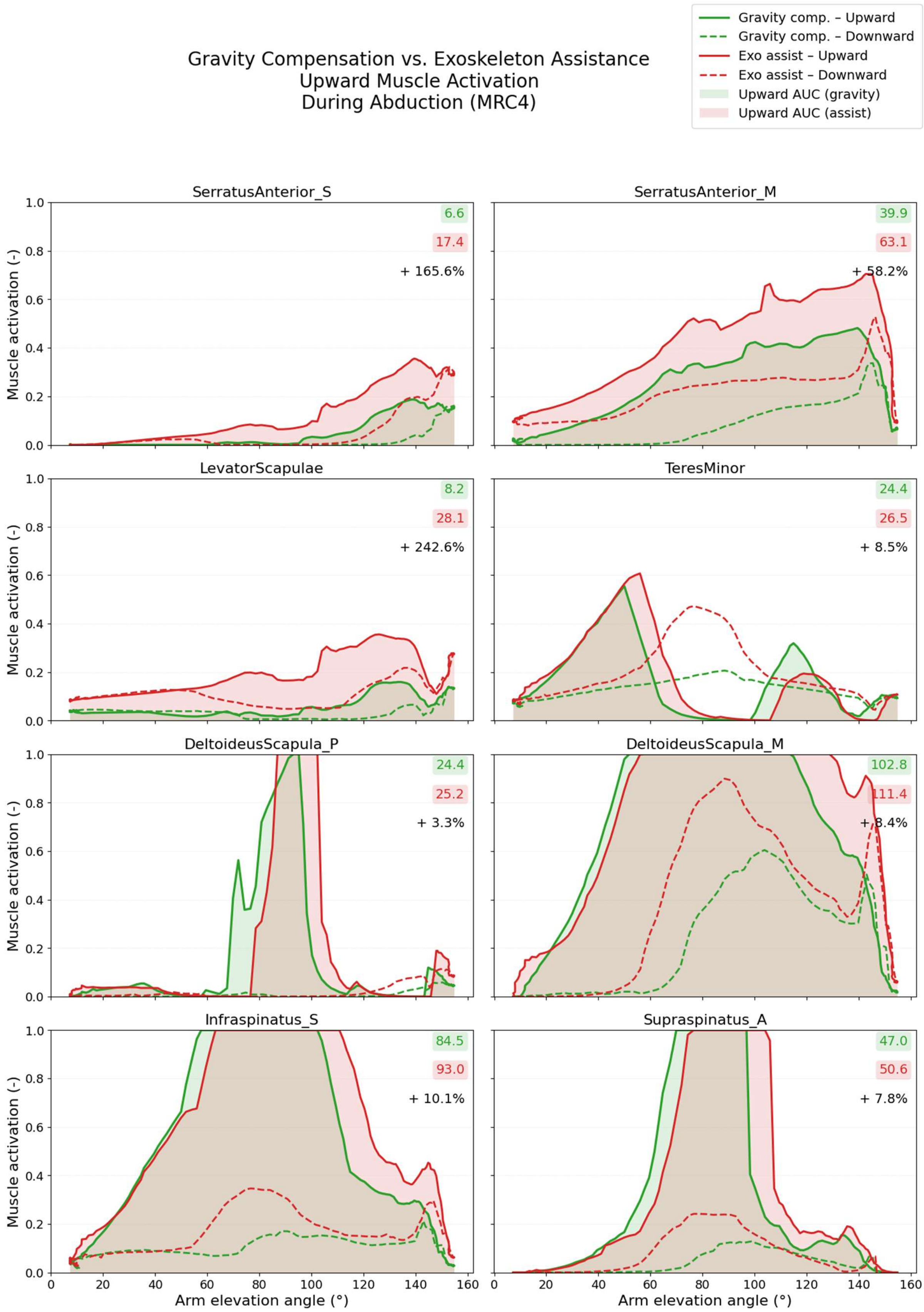


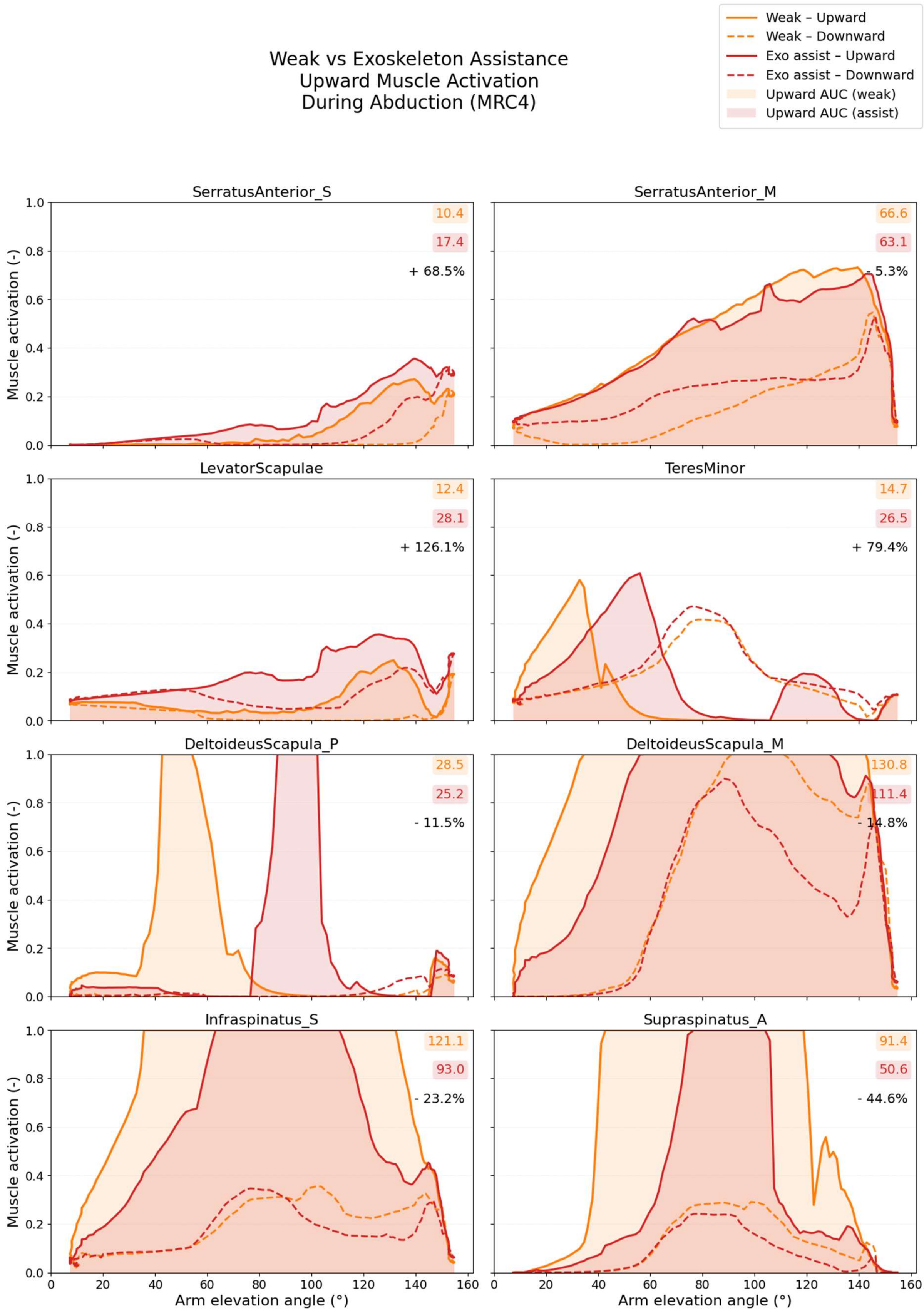
Gravity Compensation vs. Exoskeleton Assistance
Upward Muscle Activation
During Abduction (MRC3)
Gravity comp. – Upward
Gravity comp. – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (gravity)
Upward AUC (assist)
SerratusAnterior_S
5.9
12.5
+ 110.1%
SerratusAnterior_M
35.7
49.4
+ 38.5%
LevatorScapulae
7.0
19.1
+ 173.4%
TeresMinor
25.2
26.8
+ 6.3%
DeltoideusScapula_P
32.9
21.5
- 34.8%
DeltoideusScapula_M
115.9
119.5
+ 3.1%
Infraspinatus_S
94.6
100.8
+ 6.5%
Supraspinatus_A
48.3
54.8
+ 13.5%
Muscle activation (-)
Arm elevation angle (°)


Weak vs Exoskeleton Assistance
Upward Muscle Activation
During Abduction (MRC3)
Weak – Upward
Weak – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (weak)
Upward AUC (assist)
SerratusAnterior_S
7.7
12.5
+ 62.9%
SerratusAnterior_M
48.4
49.4
+ 2.0%
LevatorScapulae
9.3
19.1
+ 105.2%
TeresMinor
17.4
26.8
+ 54.2%
DeltoideusScapula_P
33.4
21.5
- 35.7%
DeltoideusScapula_M
129.7
119.5
- 7.9%
Infraspinatus_S
116.4
100.8
- 13.4%
Supraspinatus_A
82.3
54.8
- 33.4%
Muscle activation (-)
Arm elevation angle (°)

Appendix B.2: MRC4 Abduction Muscle Activation

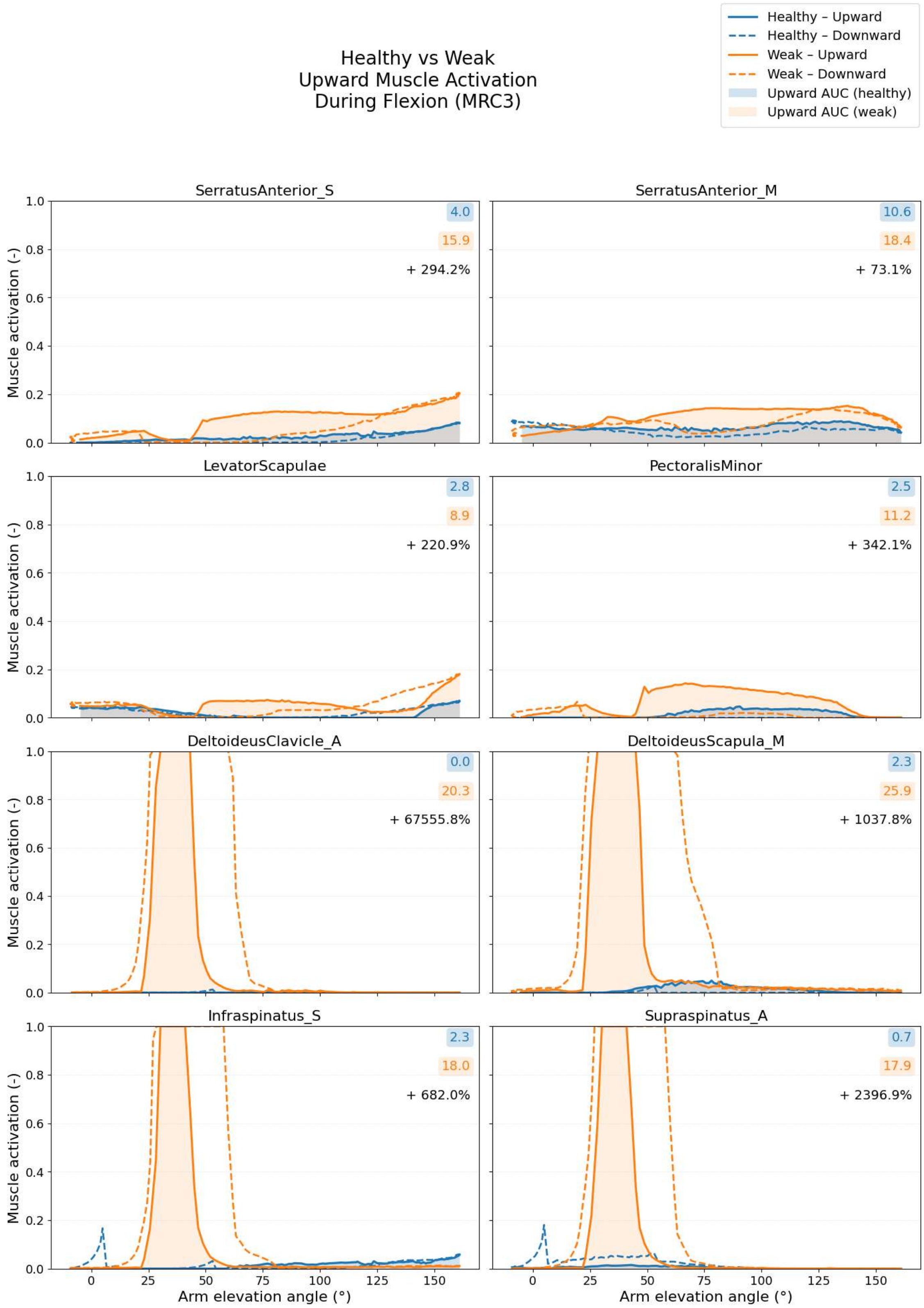


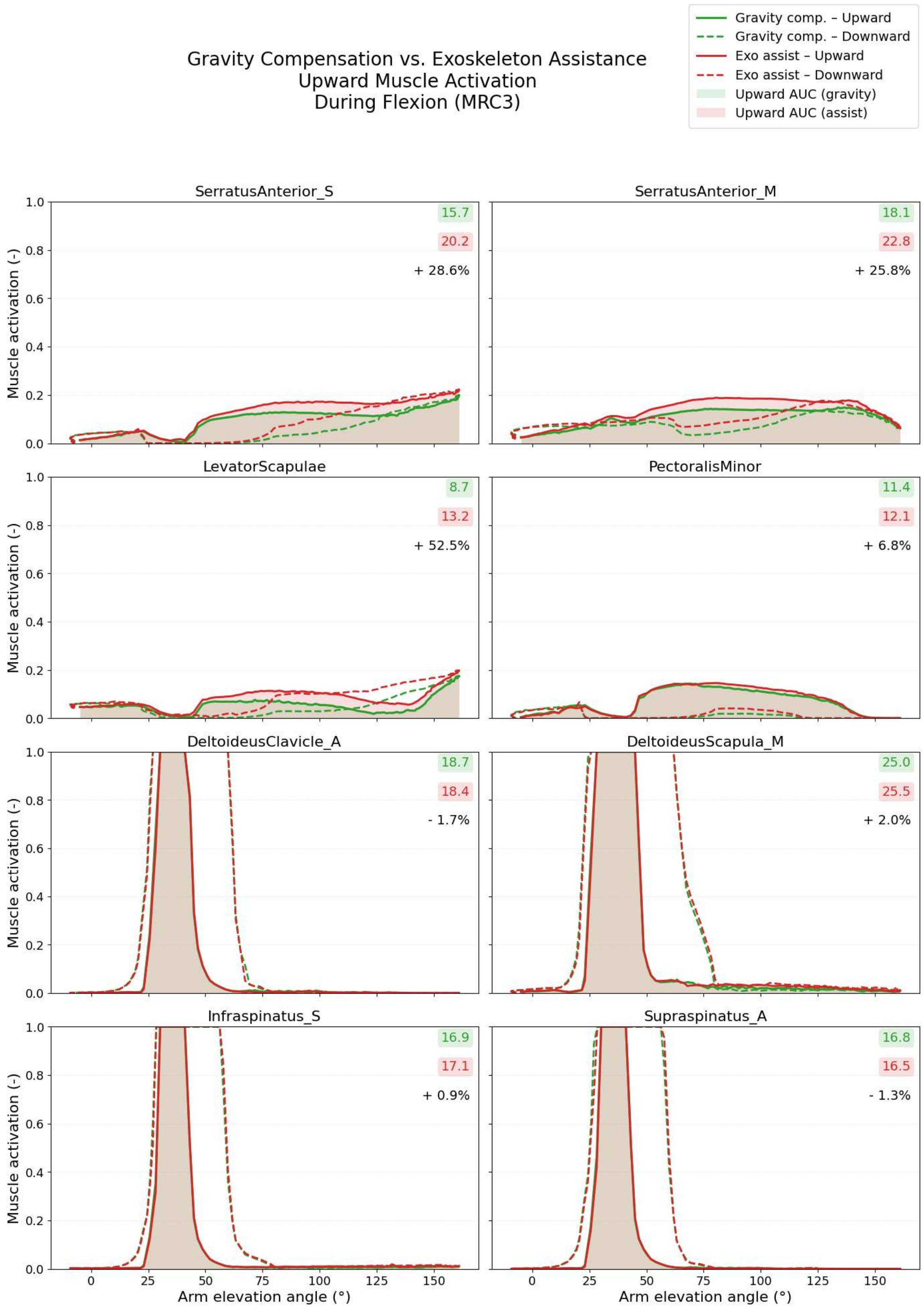

Gravity Compensation vs. Exoskeleton Assistance
Upward Muscle Activation
During Abduction (MRC4)
Gravity comp. – Upward
Gravity comp. – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (gravity)
Upward AUC (assist)
SerratusAnterior_S
6.6
17.4
+ 165.6%
SerratusAnterior_M
39.9
63.1
+ 58.2%
LevatorScapulae
8.2
28.1
+ 242.6%
TeresMinor
24.4
26.5
+ 8.5%
DeltoideusScapula_P
24.4
25.2
+ 3.3%
DeltoideusScapula_M
102.8
111.4
+ 8.4%
Infraspinatus_S
84.5
93.0
+ 10.1%
Supraspinatus_A
47.0
50.6
+ 7.8%
Muscle activation (-)
Arm elevation angle (°)



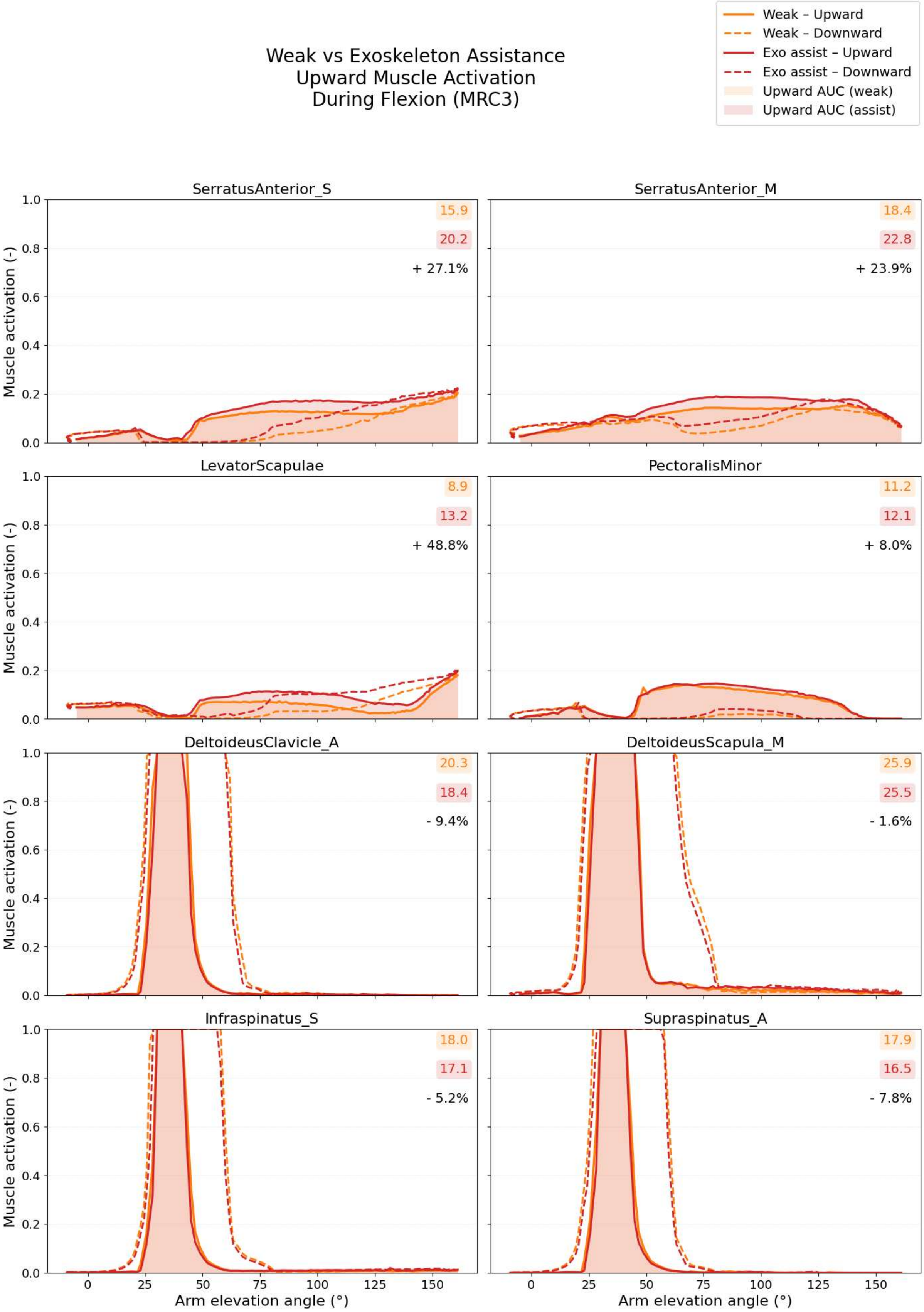
Weak vs Exoskeleton Assistance
Upward Muscle Activation
During Abduction (MRC4)
Weak – Upward
Weak – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (weak)
Upward AUC (assist)
SerratusAnterior_S
10.4
17.4
+ 68.5%
SerratusAnterior_M
66.6
63.1
- 5.3%
LevatorScapulae
12.4
28.1
+ 126.1%
TeresMinor
14.7
26.5
+ 79.4%
DeltoideusScapula_P
28.5
25.2
- 11.5%
DeltoideusScapula_M
130.8
111.4
- 14.8%
Infraspinatus_S
121.1
93.0
- 23.2%
Supraspinatus_A
91.4
50.6
- 44.6%
Muscle activation (-)
Arm elevation angle (°)

Appendix B.3: MRC3 Flexion Muscle Activation



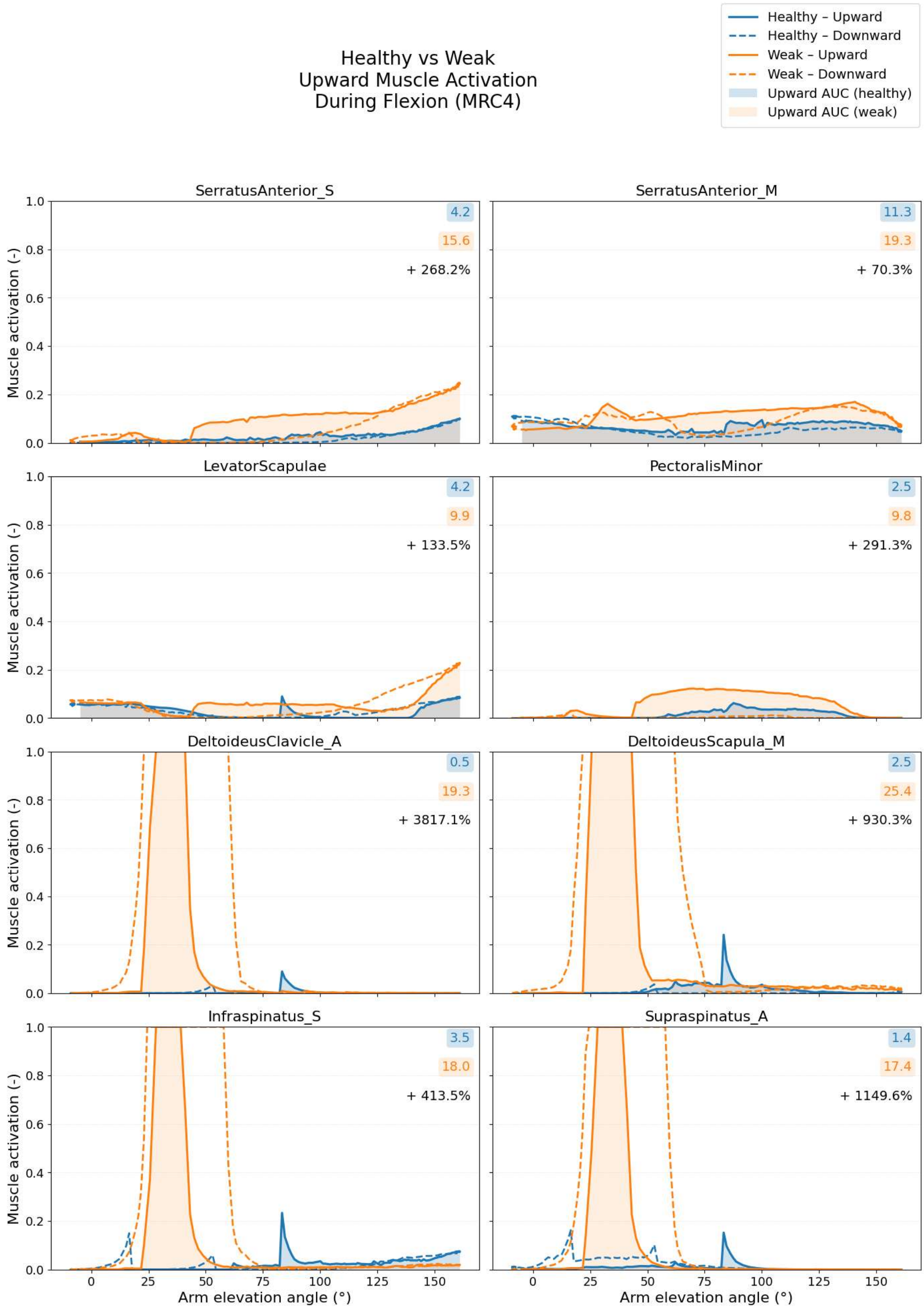


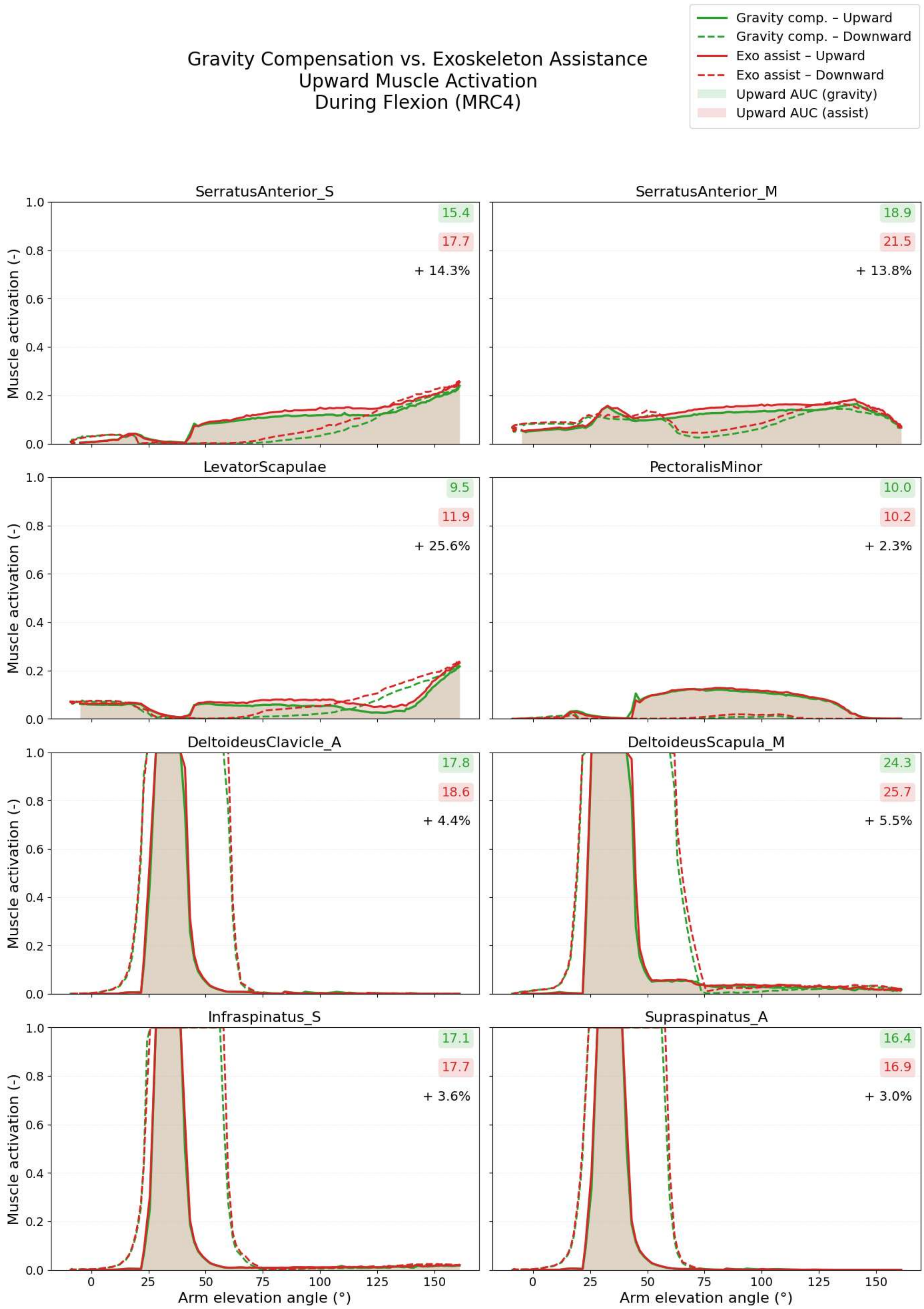
Gravity Compensation vs. Exoskeleton Assistance
Upward Muscle Activation
During Flexion (MRC3)
Gravity comp. – Upward
Gravity comp. – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (gravity)
Upward AUC (assist)
SerratusAnterior_S
15.7
20.2
+ 28.6%
SerratusAnterior_M
18.1
22.8
+ 25.8%
LevatorScapulae
8.7
13.2
+ 52.5%
PectoralisMinor
11.4
12.1
+ 6.8%
DeltoideusClavicle_A
18.7
18.4
- 1.7%
DeltoideusScapula_M
25.0
25.5
+ 2.0%
Infraspinatus_S
16.9
17.1
+ 0.9%
Supraspinatus_A
16.8
16.5
- 1.3%
Muscle activation (-)
Arm elevation angle (°)

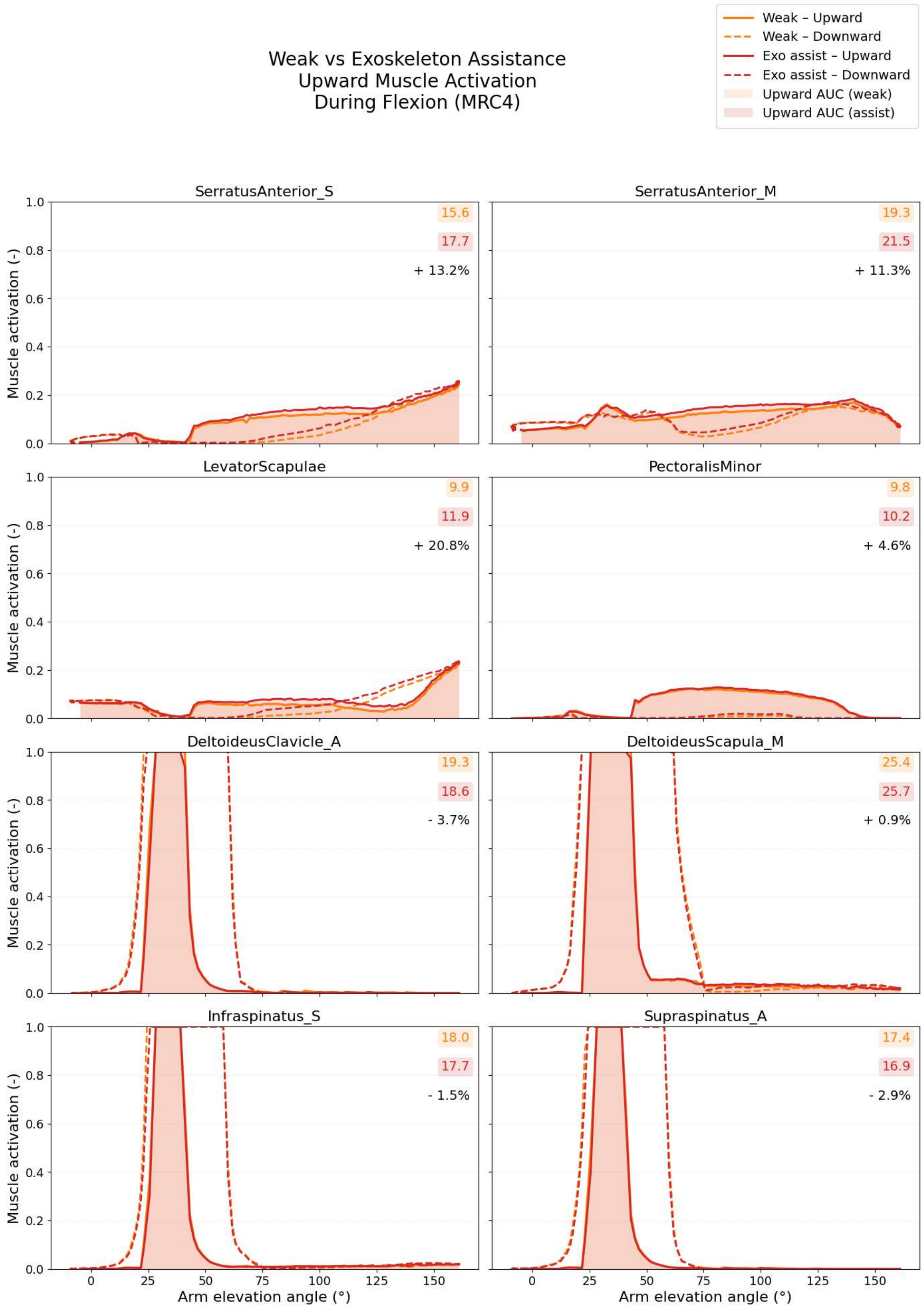


Weak vs Exoskeleton Assistance
Upward Muscle Activation
During Flexion (MRC3)
Weak – Upward
Weak – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (weak)
Upward AUC (assist)
SerratusAnterior_S
15.9
20.2
+ 27.1%
SerratusAnterior_M
18.4
22.8
+ 23.9%
LevatorScapulae
8.9
13.2
+ 48.8%
PectoralisMinor
11.2
12.1
+ 8.0%
DeltoideusClavicle_A
20.3
18.4
- 9.4%
DeltoideusScapula_M
25.9
25.5
- 1.6%
Infraspinatus_S
18.0
17.1
- 5.2%
Supraspinatus_A
17.9
16.5
- 7.8%
Muscle activation (-)
Arm elevation angle (°)

Appendix B.4: MRC4 Flexion Muscle Activation




Gravity Compensation vs. Exoskeleton Assistance
Upward Muscle Activation
During Flexion (MRC4)
Gravity comp. – Upward
Gravity comp. – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (gravity)
Upward AUC (assist)
SerratusAnterior_S
15.4
17.7
+ 14.3%
SerratusAnterior_M
18.9
21.5
+ 13.8%
LevatorScapulae
9.5
11.9
+ 25.6%
PectoralisMinor
10.0
10.2
+ 2.3%
DeltoideusClavicle_A
17.8
18.6
+ 4.4%
DeltoideusScapula_M
24.3
25.7
+ 5.5%
Infraspinatus_S
17.1
17.7
+ 3.6%
Supraspinatus_A
16.4
16.9
+ 3.0%
Muscle activation (-)
Arm elevation angle (°)
0.0
0.2
0.4
0.6
0.8
1.0
0
25
50
75
100
125
150


Weak vs Exoskeleton Assistance
Upward Muscle Activation
During Flexion (MRC4)
Weak – Upward
Weak – Downward
Exo assist – Upward
Exo assist – Downward
Upward AUC (weak)
Upward AUC (assist)
SerratusAnterior_S
15.6
17.7
+ 13.2%
SerratusAnterior_M
19.3
21.5
+ 11.3%
LevatorScapulae
9.9
11.9
+ 20.8%
PectoralisMinor
9.8
10.2
+ 4.6%
DeltoideusClavicle_A
19.3
18.6
- 3.7%
DeltoideusScapula_M
25.4
25.7
+ 0.9%
Infraspinatus_S
18.0
17.7
- 1.5%
Supraspinatus_A
17.4
16.9
- 2.9%
Muscle activation (-)
Arm elevation angle (°)