

UNIVERSITÄTSKLINIKUM HAMBURG-EPPENDORF

Zentralinstitut für Arbeitsmedizin und Maritime Medizin

Univ.-Prof. Dr. med. Volker Harth, MPH

Biomechanical Effectiveness of Occupational Shoulder Exoskeletons in Reducing Musculoskeletal Load During Tasks at or Above Shoulder Level

Dissertation

zur Erlangung des Doktorgrades Dr. rer. biol. hum.
an der Medizinischen Fakultät der Universität Hamburg.

vorgelegt von:

Mirko Kaufmann
aus Koblenz

Hamburg 2025

(wird von der Medizinischen Fakultät ausgefüllt)

**Angenommen von der
Medizinischen Fakultät der Universität Hamburg am:** 29.06.2026

**Veröffentlicht mit Genehmigung der
Medizinischen Fakultät der Universität Hamburg.**

Prüfungsausschuss, der/die Vorsitzende: Prof. Dr. Volker Harth

Prüfungsausschuss, zweite/r Gutachter/in: Prof. Dr. Albert Nienhaus

Table of Contents

1. Work-related musculoskeletal disorders	1
2. The shoulder complex – anatomy and function	4
3. Biomechanics of arm elevation	9
4. Work-related musculoskeletal disorders in the shoulder	13
5. Preventive measures to reduce musculoskeletal shoulder load	16
6. Potential of occupational exoskeletons in reducing work-related musculoskeletal load	18
6.1. Occupational exoskeletons for the upper extremities	19
6.2. Influence of occupational shoulder exoskeletons on musculoskeletal shoulder load	22
7. Aim of the thesis	27
8. Quantification of the support moment of three occupational shoulder exoskeletons	28
8.1. Measurement setup and preparations	28
8.2. Measurement session	34
8.3. Data processing	38
8.4. Results	43
8.5. Discussion and limitations	51
8.6. Conclusion	58
9. Influence of a shoulder exoskeleton on mechanical load and muscle demand during static arm elevation	59
9.1. Participants	59
9.2. Experimental design and task description	59
9.3. Measurement instrumentation and data acquisition	60
9.4. Data processing and dependent variables	65
9.5. Statistics	69
9.6. Results	70
9.7. Discussion and limitations	75

9.8. Conclusion	79
10. Influence of a shoulder exoskeleton on mechanical load and muscle tissue oxygenation during static arm elevation	80
10.1. Participants	80
10.2. Experimental design and task description	80
10.3. Measurement instrumentation and data acquisition	82
10.4. Data processing and dependent variables	86
10.5. Statistics	87
10.6. Results	88
10.7. Discussion and limitations	90
10.8. Conclusion	94
11. Influence of shoulder exoskeletons on mechanical load and muscle demand during aircraft manufacturing tasks	95
11.1. Participants	95
11.2. Workplace replica and task description	96
11.3. Measurement instrumentation and data acquisition	99
11.4. Data processing and dependent variables	103
11.5. Statistics	105
11.6. Results	106
11.7. Discussion and limitations	112
<i>11.7.1. Influence of the OSEs during the drilling task</i>	113
<i>11.7.2. Influence of the OSEs during the riveting task</i>	117
<i>11.7.3. Influence of the OSEs during the screwing task</i>	119
<i>11.7.4. Differences in the influence between the OSEs</i>	122
<i>11.7.5. Summary</i>	123
<i>11.7.6. Limitations</i>	125
11.8. Conclusion	127
12. Discussion and limitations of the thesis	128

13. Conclusion of the thesis	138
14. Zusammenfassung	140
15. Abstract	141
16. List of Figures	142
17. List of Tables	149
18. List of Abbreviations	150
19. List of References	153
20. Danksagung	163
21. Lebenslauf Lebenslauf entfällt aus datenschutzrechtlichen Gründen	166
22. Eidesstattliche Versicherung	168

1. Work-related musculoskeletal disorders

Work-related musculoskeletal disorders (WMSDs) are listed as the most common occurring diseases in the European Union causing significant health problems in various countries and work sectors (EU-OSHA, 2019). In 2015, 58% of European workers (n = 31,612; 28 countries) reported suffering from at least one (or more) musculoskeletal disorder (MSD), with 16 countries reporting prevalence rates exceeding 60%, according to the European Agency for Safety and Health at Work (EU-OSHA; EU-OSHA, 2019). Thus, WMSDs remain the leading health issue, ranking above other conditions such as mental disorders (e.g. stress and depression), pulmonary or cardiovascular disorders and infectious diseases (BAuA, 2014; BAuA, 2024; EU-OSHA, 2019). According to the Federal Institute for Occupational Safety and Health (BAuA), in 2023, WMSDs accounted for 20.8% of all days of incapacity to work in Germany – surpassing mental disorders (13.7%), respiratory diseases (17.4%), and circulatory system diseases (6.4%; Figure 1; BAuA, 2024). Compared with 2013, this percentage has only slightly decreased in 2023 (from 23.2% in 2013; Figure 1; BAuA, 2014), with an average of 16.5 days in the vehicle manufacturing industry (BAuA, 2024).

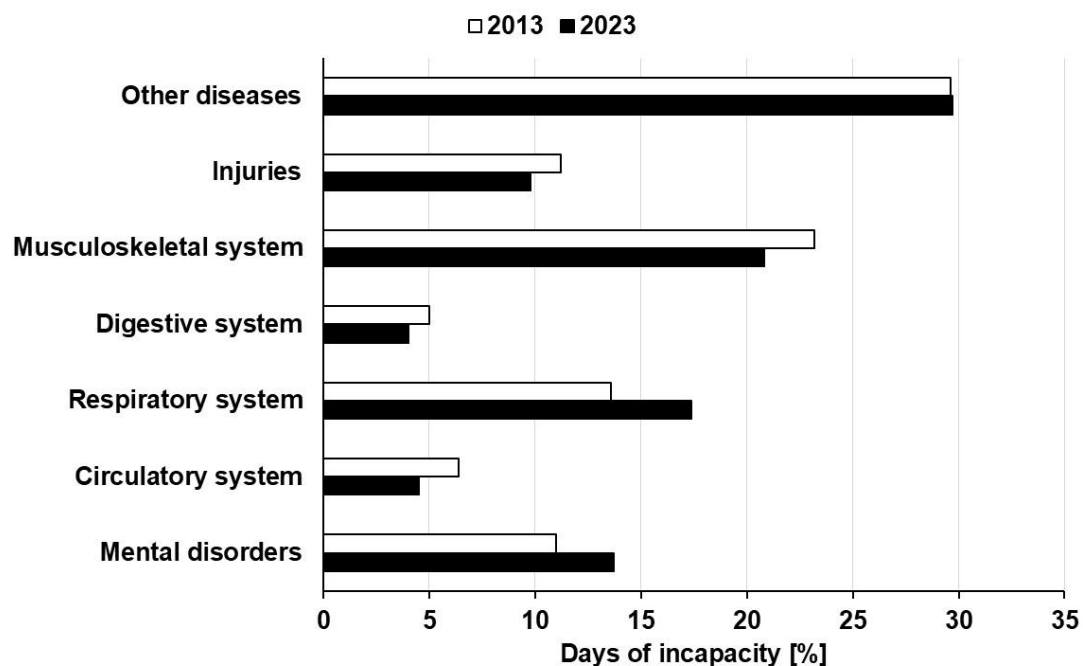


Figure 1: Days of incapacity to work by diagnostic groups in Germany for the years 2013 and 2023. The data were derived from BAuA, 2014 and BAuA, 2024.

If not prevented by interventions or treated at an earlier stage, WMSDs can progress into chronic conditions that affects not only the physiological health of the worker, but also their mental state (EU-OSHA, 2019). An increase in mental health problems, such as higher levels of anxiety, sleeping disturbances and general fatigue, has been linked to a higher prevalence of WMSDs (EU-OSHA, 2019). As well as impacting the workers physical and mental health, WMSDs also result in a loss of productivity and increased social costs (EU-OSHA, 2019). In the German construction industry in 2023, days of incapacity due to WMSDs resulted in a loss of production of around €1.80 billion, as well as a loss of gross value added of €2.89 billion (BAuA, 2024). Furthermore, due to demographic changes and an ageing population, it is predicted that the costs related to WMSDs will only increase in the future (EU-OSHA, 2019), highlighting the importance of preventive measures to reduce their occurrence.

In general, MSDs are defined as impairments or abnormalities of the musculoskeletal system that affect rigid body structures (e.g. bones), joints, or soft tissues (e.g. muscles, ligaments and tendons; Aptel et al., 2002; EU-OSHA, 2019). They result from high physical load/stress on the musculoskeletal system (Aptel et al., 2002; EU-OSHA, 2019). Furthermore, WMSDs are a subcategory of health issues arising from musculoskeletal load (MSL) in occupational or work-related contexts that develop as a result of physical strain experienced at the workplace (EU-OSHA, 2019; Forde et al., 2002). Several occupational risk factors that contribute to the development of WMSDs by increasing MSL have been identified (BMAS, 2021; DGUV, 2023; EU-OSHA, 2019; Forde et al., 2002), including:

- 1) Working in awkward body postures, resulting in sustained static muscle load (e.g. with high trunk flexion or with the hands at or above shoulder height);
- 2) Performing highly, repetitive movements;
- 3) Handling of manual materials (e.g. during lifting);
- 4) Performing work requiring high force exertion; and
- 5) Being exposed to vibrations resulting from hand tools (either to specific body segments or the whole body).

Musculoskeletal pain in the upper limbs and shoulders is one of the most commonly reported WMSD by workers, alongside back pain (Aptel et al., 2002; BAuA, 2024; EU-OSHA, 2019).

The BAuA reported that 46.1% of the 18,625 workers questioned experience pain in the neck and the shoulder, with 43.8% of these workers (n = 1,800) reporting pain in these regions for more than 48 hours (BAuA, 2024). Furthermore, health insurance companies identified lesions of the shoulder as the most common cause of days of incapacity in Germany in 2023, behind back pain or damage to the intervertebral discs, with a rate of 19.04% (1.1% cases per 100 members; Grobe & Bessel, 2024; Klemm et al., 2024). Thereby, WMSDs in the shoulder, e.g. impingement syndrome or rotator cuff tendinitis, are often related to increased MSL during industrial tasks with elevated arms (Grieve & Dickerson, 2008; Hagberg, 1984; Kadefors et al., 1976).

2. The shoulder complex – anatomy and function

The shoulder complex consists of four (main) joints and three bones (Figure 2); the clavicle; the scapula and the humerus; and connects the arm to the torso via the rib cage and sternum, enabling postural changes in the arm through motion of the clavicle, the scapula, and the humerus (Echtermeyer & Sangmeister, 1996; Gohlke, 2002; Peat, 1986). In the frontal plane, the medial end of the clavicle is connected to the sternum via the sternoclavicular joint (*articulatio sternoclavicularis*, SCJ; Figure 2), which acts as ball-and-socket joint, allowing the clavicle to move in three degrees of freedom relative to the sternum: upward and downward (elevation/depression), forward and backward (protraction/retraction) and rotation about its longitudinal axis (Gohlke, 2002; Peat, 1986; Singleton, 1966). At the lateral end of the clavicle, the acromioclavicular joint (*articulatio acromioclavicularis*, ACJ; Figure 2) attaches the clavicle to the scapula at the medial border of the acromion (Culham & Peat, 1993; Singleton, 1966; Terry & Chopp, 2000). The ACJ therefore provides the scapula with movement on the clavicle in three planes (around the coronal, sagittal and vertical axes), while the two main motion directions for the scapula are forward/backward and upward/downward (elevation/depression) gliding over the clavicle (Peat, 1986).

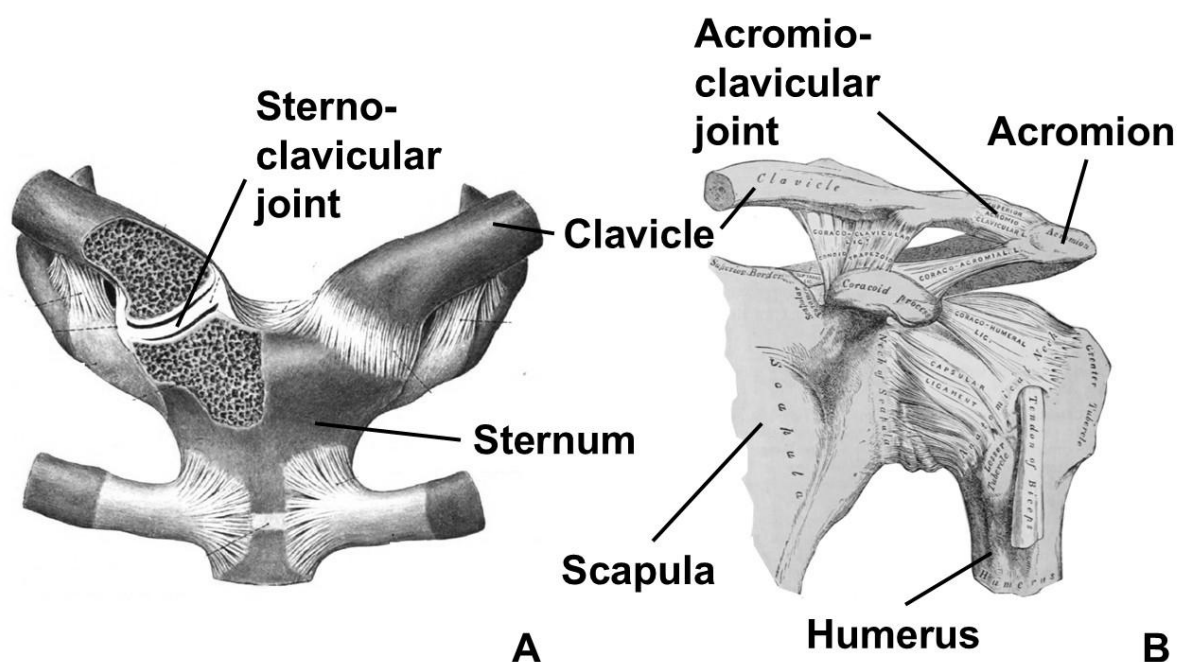


Figure 2: Anterior view of the upper part of the sternum (A) and the shoulder complex (B). Both graphics were extracted and modified from Sobotta (1909; A; p. 120) and Gray and Lewis (1918; B; p. 316). Public domain.

The scapulothoracic joint (STJ) is located between the scapula and the thorax and is characterized by the gliding movement of the scapula over the rib cage and, unlike other shoulder joints, is not a true anatomical joint, as it lacks cartilaginous joint surfaces between the articulating structures (Echtermeyer & Sangmeister, 1996; Peat, 1986; Singleton, 1966). Lastly, the humerus is connected to the scapula via the glenohumeral joint (articulatio humeri, GHJ), with the head of the humerus placed in the glenoid cavity of the scapula (Figure 3; Culham & Peat, 1993; Peat, 1986; Singleton, 1966). The glenoid cavity is surrounded by cartilage (the glenoid labrum) to increase its contact area with the humeral head (Figure 3; Echtermeyer & Sangmeister, 1996; Singleton, 1966). However, this contact area remains smaller than that of the humeral head (Figure 3; Echtermeyer & Sangmeister, 1996; Culham & Peat, 1993; Singleton, 1966; Terry & Chopp, 2000), and the surfaces of both the humeral head and the glenoid cavity are generally asymmetrical and not congruent with each other (Culham & Peat, 1993; Peat, 1986). These features contribute to joint instability (Culham & Peat, 1993), but also ensure a high mobility of the arm due to a movable joint center and joint axes (Peat, 1986; Singleton, 1966; Terry & Chopp, 2000).

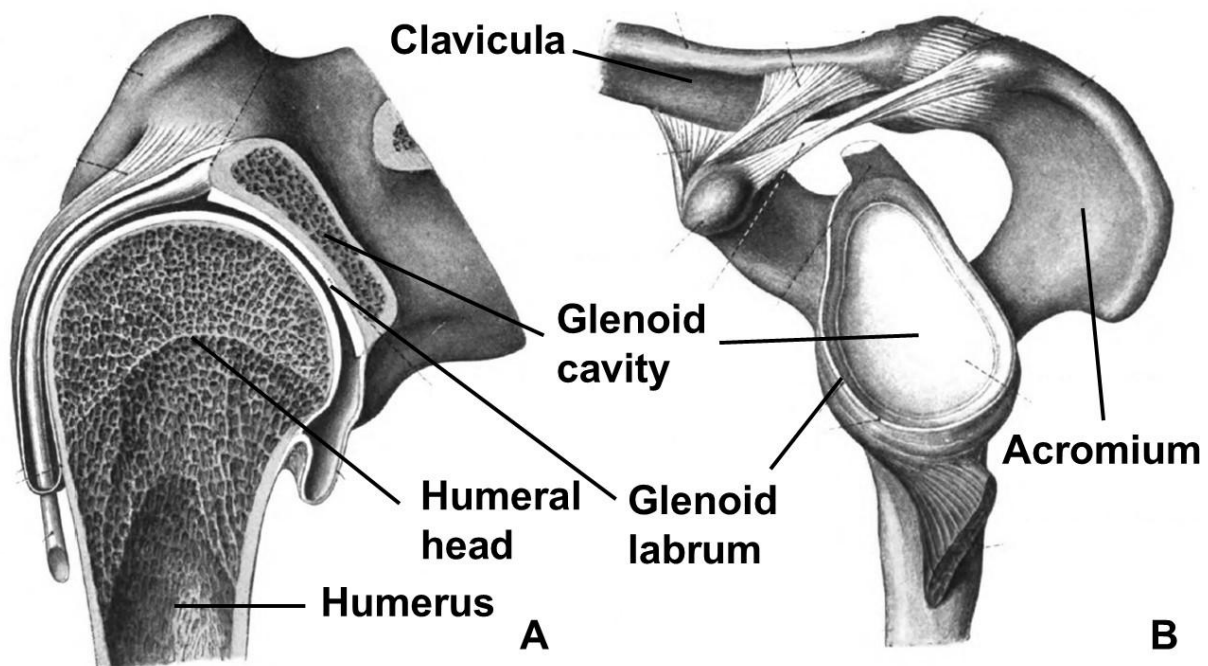


Figure 3: Glenohumeral joint in anterior (sliced view A) and sagittal (B) view. Both graphics (A & B) were extracted and modified from Sobotta (1909; p. 120). Public domain.

The multiaxial ball-and-socket characteristic of the GHJ allows the humerus to move in three directions: 1) flexion and extension in the sagittal plane, 2) abduction and adduction in the frontal plane and 3) rotation around the longitudinal axis (Echtermeyer & Sangmeister, 1996; Peat, 1986; Singleton, 1966). This permits a high range of motion due to the mobility of the joint axes and the joint center in three directions (Echtermeyer & Sangmeister, 1996; Peat, 1986; Singleton, 1966). The arm can therefore be abducted up to 120° before the humerus reaches the acromion, which blocks higher abduction angles (Echtermeyer & Sangmeister, 1996; Singleton, 1966). Various muscles around the GHJ and at the scapula act in the form of different force couples to elevate the humerus and provide stability to the GHJ (Figure 4; Culham & Peat, 1993; Peat, 1986; Singleton, 1966; Terry & Chopp, 2000). The largest muscle contributing to arm elevation is the musculus deltoideus, which consists of three parts that circle the GHJ (Figure 4 A & C; Lee & An, 2002). The anterior part (the musculus deltoideus pars claviculæ, AD) originates from the anterior and superior surfaces of the outer third of the clavicle and the anterior acromion, and is primarily responsible for the anteversion (flexion) of the arm (Figure 4 A). The medial part of the deltoideus (musculus deltoideus pars acromialis, MD) is responsible for abducting the arm, with its origin at the lateral margin of the acromion (Figure 4 A). Lastly, the posterior part of the deltoideus (musculus deltoideus pars spinalis, PD) conducts the retroversion (extension) of the arm, originating from the scapular spine (spinae scapulae, Figure 4 C; cf. Echtermeyer & Sangmeister, 1996; Lee & An, 2002). Four muscles are attached to the humerus at the tuberculum majus or minus to stabilize the GHJ position and contributing to arm elevation by pulling the humerus head against the glenoid cavity (Figure 4 B; Culham & Peat, 1993; Page, 2011; Peat, 1986; Singleton, 1966):

- 1) the musculus infraspinatus (INF);
- 2) the musculus supraspinatus (SUP);
- 3) the musculus teres minor (TM); and
- 4) the musculus subscapularis (SCA).

These are referred to as the rotator cuff muscles.

Further abduction of the arm beyond 120° (up to 180°) requires the scapula to externally rotate by 60° over the STJ, which is partially performed by the upper part of the trapezius muscle (the musculus trapezius pars descendens, UT) elevating the scapula at its lateral part (Figure 4 C; Echtermeyer & Sangmeister, 1996; Terry & Chopp, 2000). The lower part (the musculus trapezius pars ascendens, LT) and the middle part (the musculus trapezius pars horizontalis, MT) of the trapezius muscle are also required to move the scapula in lower or medial direction, respectively (Figure 4 C; Echtermeyer & Sangmeister, 1996).

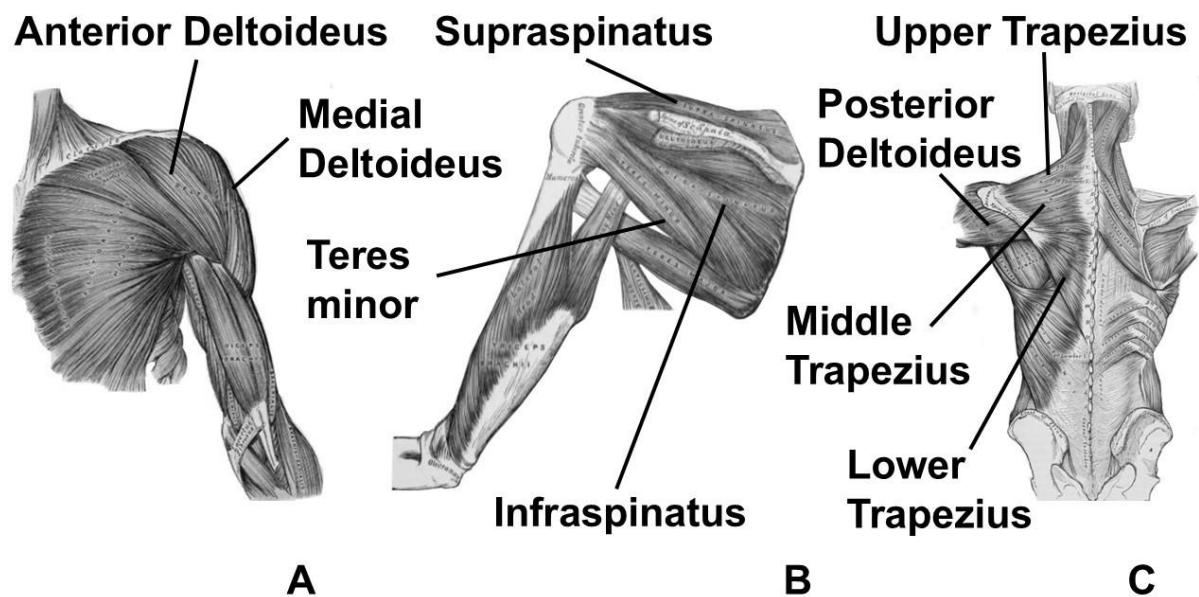


Figure 4: Anterior (A) and posterior view (C) on various shoulder muscles in addition with a sliced posterior view of the rotator cuff muscles (B). All three graphics were extracted and modified from Gray and Lewis (1918; p. 436, p. 441 & p. 433). Public domain.

Contraction of the deltoideus muscles generates an upward/outward force on the humerus, resulting in a translatory movement of the humeral head towards the acromion (Figure 5; Culham & Peat, 1993; Page, 2011; Peat, 1986; Singleton, 1966). In contrast, the infraspinatus, the teres minor and the subscapularis muscles of the rotator cuff depress the humeral head with a downward/horizontal force, which could cause a subluxation of the GHJ (Figure 5; Culham & Peat, 1993; Page, 2011; Peat, 1986; Singleton, 1966). The two forces are initiated in different parts on the humeral head: the force generated by the deltoideus is initiated below the center of rotation of the GHJ, while the force generated by the three rotator cuff muscles is initiated above it (Peat, 1986; Singleton, 1966). As these two forces act in parallel, opposite directions and are non-collinear to each other, they form a force couple that generates a rotation around the GHJ, resulting in elevation of the arm (Culham & Peat, 1993; Peat, 1986; Robertson et al., 2014; Singleton, 1966).

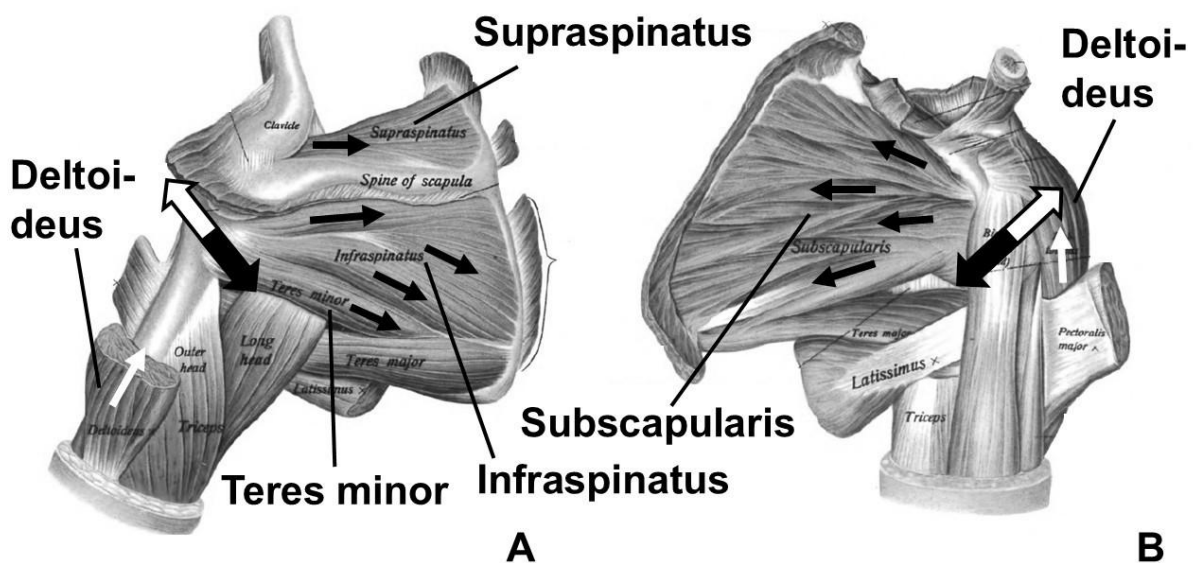


Figure 5: Glenohumeral joint with the surrounding muscles in posterior (A) and anterior (B) view. Thin arrows indicate force direction during contraction for the rotator cuff muscles (black) and the deltoideus parts (white). Thick arrows indicate the direction the humeral head is moving caused by the respective muscle contraction. Based on Singleton (1966). Both graphics were extracted and modified from Sobotta (1909; p. 186). Public domain.

3. Biomechanics of arm elevation

In order to elevate the arm, the forces acting on the humerus generate a rotational moment around the GHJ (the gravitational shoulder joint moment), which is equal to the gravitational moment generated by the gravity acting on the center of gravity (CoG) of the arm. The gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$) can be calculated as follows:

$$\tau_{grav}^{shoulder} = F_{grav}^{arm} * r_{CoG}^{arm} \quad (3.1)$$

where F_{grav}^{arm} is the gravitational force of the arm and r_{CoG}^{arm} is defined as the distance between the GHJ and the CoG of the arm (Figure 6). The position of the CoG changes with the position of the center of mass (CoM), which depends on the mass distribution of the individual arm segments (the upper arm, the lower arm, and the hand), and their angular positions relative to each other and to the GHJ, i.e. the joint angles. For example, when the arm is elevated at 90° angle at the shoulder, fully extended at the elbow (elbow flexion at 0°), with the wrist in a neutral position and all fingers extended (open hand), the CoG reaches its maximal position in relation to the GHJ at this shoulder angle (Figure 6). Increasing the elbow flexion angle (moving the hand and forearm towards the GHJ) shifts the CoG closer to the GHJ, thereby reducing the distance between the two points, and conversely r_{CoG}^{arm} . Consequently, in this case, the relative increment of the elbow flexion angle decreases $\tau_{grav}^{shoulder}$. In contrast, adding an external load (e.g. a dumbbell or a tool) to the hand shifts the CoG in opposite direction to the GHJ, thereby increasing F_{grav}^{arm} and resulting in a higher $\tau_{grav}^{shoulder}$.

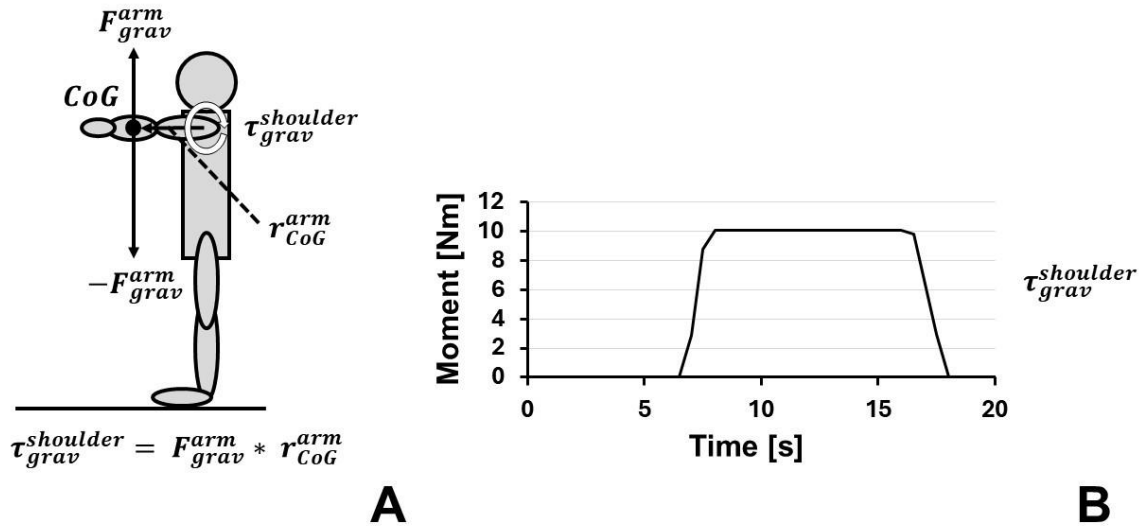


Figure 6: Calculation of the gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$; A) as a product of the gravitational force (F_{grav}^{arm}), acting at the center of gravity (CoG), and the distance between the glenohumeral joint center and the CoG (r_{CoG}^{arm}). In addition, $\tau_{grav}^{shoulder}$ is presented for an exemplary participant performing an arm elevation in 90° shoulder flexion (arm fully extended, wrist in neutral position) for five seconds (B).

During static arm elevation, it has been stated that $\tau_{grav}^{shoulder}$ is a valid indicator of the mechanical load at the GHJ, as a strong linear relationship can be expected between the shoulder joint moment and the joint compression forces acting on the GHJ surface (Praagman et al., 2000). Consequently, an increase in $\tau_{grav}^{shoulder}$ (e.g. when working with a heavy tool) would also increase the joint compression forces, potentially causing damage to the joint surfaces, particularly during prolonged arm elevation (Praagman et al., 2000). As previously mentioned, the shoulder muscles (primarily the deltoideus muscles, the rotator cuff muscles and the trapezius muscles) contract to elevate the arm, resulting in higher internal muscle forces (Hisaeda et al., 2001), which further increases intramuscular pressure (IMP; Järvholm et al., 1988a; Järvholm et al., 1988b; Järvholm et al., 1989; Järvholm et al., 1991a; Palmerud et al., 2000). Järvholm and colleagues (1988a; 1988b; 1989; 1991a) and Palmerud et al. (2000) reported an increase in IMP during static arm elevation in the supraspinatus, the infraspinatus, the descending trapezius and the anterior deltoideus.

Furthermore, they revealed a higher level of increase in IMP in these muscles when.

- a) the arm elevation angle increased, reaching a maximum at 90°;
- b) the external load was increased;
- c) an external force was applied to the arm; and/or
- d) the arm was fully extended (elbow flexion at 0°) compared to when the forearm was flexed at 90°

(Järvholm et al., 1988a; Järvholm et al., 1989; Järvholm et al., 1991a; Palmerud et al., 2000). These factors would increase the required $\tau_{grav}^{shoulder}$ to maintain the arm elevation, as well as the necessary muscle forces. For static arm elevation, higher muscle forces can be interpreted as an increased number of activated muscle fibres, resulting in an higher muscle activity (Robertson et al., 2014). Therefore, a direct positive relationship can be assumed between $\tau_{grav}^{shoulder}$, muscle activity (i.e. muscle forces) and IMP for static arm elevation. This assumption is supported by the results of Järvholm et al. (1989, 1991a), who found a linear relationship between these three parameters in their studies. Furthermore, they demonstrated that muscle blood flow in the supraspinatus muscle is already impaired due to an increased IMP at an arm elevation angle of 30°, suggesting a relationship between IMP and muscle blood flow (Järvholm et al., 1988b).

In addition to potential impairments in blood flow, the increased muscle forces required for arm elevation also result in higher metabolic demands due to increased energy turnover (Perrey et al., 2010a). Consequently, impaired muscle blood flow and higher metabolic demands decrease local muscle tissue oxygenation (MTO), shifting muscle metabolism towards anaerobiosis (de Ruiter et al., 2007; Perrey et al., 2010a). Inadequate MTO due to low oxygen availability has been reported to cause muscle fatigue in the deltoideus or the rotator cuff muscles (Grieve & Dickerson, 2008; Hagberg, 1981). Increasing $\tau_{grav}^{shoulder}$ (e.g. through the weight of a working tool or when an external force is applied) could also contribute to a faster progression of shoulder muscle fatigue (SMF; Ferguson et al., 2013; Grieve & Dickerson, 2008; Järvholm et al., 1991a; Palmerud et al., 2000). Herberts and Kadefors (1976) and Kadefors et al. (1976) found an increased muscle demand in the supraspinatus in workers performing welding with their hands above shoulder level resulting in muscle fatigue. Furthermore, Kadefors et al. (1976) reported muscle fatigue in the deltoideus and trapezius muscles when overhead welding was performed by inexperienced welders.

Hagberg (1981) found that muscle fatigue in the supraspinatus occurred after five minutes of arm elevation at 90° of both anteversion and abduction. An increased SMF, particularly in the rotator cuff muscles, resulting from repetitive arm elevation has been linked to changes in the kinematics of the shoulder joints and further instability of the glenohumeral head (Chen et al., 1999; Chopp et al., 2010b; Ebaugh et al., 2006; Grieve & Dickerson, 2008). Decreased forces in the rotator cuff muscle due to fatigue may lead to increased superior translation of the GHJ caused by the upward force initiated by the deltoideus muscles (Chen et al., 1999; Chopp et al., 2010b; Page, 2011). This can result in a decrease in the subacromial space and further to contact between the humeral head and the coracoacromial arch, and consequently, in impingement of soft tissues (e.g. the supraspinatus tendon; Chen et al., 1999; Chopp et al., 2010b; Culham & Peat, 1993; Grieve & Dickerson, 2008; Page, 2011). As demonstrated by Flatow et al. (1994) and Burns and Whipple (1993), the subacromial space is already compromised during arm elevation (at 60° of arm elevation), even in a non-fatigue state, resulting in a contact between the supraspinatus tendon and the acromion. Chen et al. (1999) and Chopp et al. (2010b) reported increased superior motion of the humeral head during arm elevation in their studies after performing a task designed to fatigue the rotator cuff muscles which could potentially cause subacromial impingement. Consequently, changes in the glenohumeral joint kinematics and the superior translational movement of the humeral head due to SMF might compromise the subacromial space further during movements or tasks performed at or above shoulder level (Chen et al., 1999; Chopp et al., 2010b; Grieve & Dickerson, 2008; Page, 2011). Therefore, industrial tasks requiring prolonged arm elevation can potentially result in the accelerated development of subacromial impingement syndrome induced by muscle fatigue (Chopp et al., 2010b; Grieve & Dickerson, 2008).

4. Work-related musculoskeletal disorders in the shoulder

In 2025, rotator cuff muscle lesions in the shoulder were officially recognized as a new occupational disease in Germany, following a scientific evaluation conducted by the Federal Ministry of Labour and Social Affairs (BMAS; BMAS, 2021). Four potential sources of hazard have been identified, where prolonged and intensive exposure to any one of these factors during work can result in the development of this occupational disease due to an increased MSL (BMAS, 2021):

- 1) working with the hands at or above shoulder level;
- 2) repetitive movement of the upper arm in the shoulder joint;
- 3) application of forces in the shoulder area, particularly during load handling;
and/or
- 4) vibrations on the hand-arm system.

Specific occupational groups and tasks associated with the above-mentioned exposures have also been identified, e.g. welding or assembly tasks in the metal and electronics construction industry, and overhead tasks in the automotive industry (BMAS, 2021). Working with the arms at or above shoulder level has been associated in various industrial sectors and countries with an increased risk of the development of shoulder disorders, such as rotator cuff tendinitis or lesions (particularly in the supraspinatus tendon) and subacromial impingement syndrome (Bodin et al., 2012a; Grieve & Dickerson, 2008; Herberts et al., 1984; Miranda et al., 2005; Punnett et al., 2000; Seidler et al., 2011). These disorders can cause restricted (often painful) movement in the arm and shoulder and often require surgical procedures to restore the functionality (Bodin et al., 2012a; Grieve & Dickerson, 2008; Herberts et al., 1984; Miranda et al., 2005; Punnett et al., 2000; Seidler et al., 2011).

Herberts et al. (1984) reported a higher prevalence of a diagnostic supraspinatus tendinopathy (18%) in a group of 131 Swedish ship craft welders who frequently worked with their arms above shoulder level, compared to a control group of 57 workers. Similarly, a high prevalence of 16.2% was observed in a group of 188 shipyard plate-workers compared to the same control group (Herberts et al., 1984). Furthermore, supraspinatus tendinopathy developed at an earlier age in the welding group than in the plate-working group (Herberts et al., 1984). This was assumed to be due to the static nature of the welder's work compared to the plate-worker's dynamic work characteristics (Herberts et al., 1984).

In a study by Miranda et al. (2005), a positive relationship was reported between the duration for which workers performed tasks above shoulder height (at least one hour per day) and the prevalence of a clinically diagnosed rotator cuff tendinitis in 4,071 Finnish workers. Based on these findings, the researchers concluded that workers exposed to over shoulder work for 14 to 23 years were 4.7 times more likely to develop rotator cuff tendinitis (Miranda et al., 2005). They even found an increased risk (more than three times higher) of chronic rotator cuff tendinitis in men with shorter exposure times (1-3 years), whereas for women, a longer exposure time was expected (Miranda et al., 2005). Similar findings were reported by Bodin et al. (2012a), who found an increased risk of a rotator cuff syndrome or general shoulder pain (factor 2.4) in French male workers ($n = 3,710$) after working with the arms abducted over 90° for at least two hours a day. No increased risk was found for the female participants (Bodin et al., 2012a). Bodin et al. (2012b) reported similar results in an additional prospective study with increased incidence of clinically diagnosed rotator cuff syndrome (by a factor of 3.7) among 839 French men who worked above shoulder level for at least two hours a day over the period of five years. In contrast to the previous study, female workers ($n = 617$) also showed an increased incidence (by a factor of 3.3) when repeatedly performing tasks involving arm elevation between 60° and 90° (Bodin et al., 2012b).

In their case-control study, Punnett et al. (2000) compared 93 workers (the case group) from an American automobile manufacturer who had reported shoulder disorders over a ten-month period, with a control group of 125 workers from the same company who had not reported any shoulder problems. The case group exhibited significantly higher levels of shoulder movement above 90° shoulder abduction than the control group (Punnett et al., 2000). Furthermore, the prevalence of a diagnosed shoulder disorder was higher in the case group than in the control group (Punnett et al., 2000). Punnett et al. (2000) also concluded that the risk increased with longer exposure durations. However, even exposure for as little as 10% (or more) of the work cycle could be sufficient to cause a chronic or recurrent shoulder disorder (Punnett et al., 2000). Seidler et al. (2011) found similar results in the form of a dose-response relationship in a case-control study of German workers. Workers who had performed overhead work for a cumulative time of more than 3,195 hours over the course of their working lives were twice as likely to develop a supraspinatus tendon lesion (483 male workers with the disorder compared to 300 healthy men; Seidler et al., 2011).

Based on their cohort study results (13.3 million Danish workers), Dalbøge et al. (2014) derived a dose-response relationship with an increased risk of surgery due to a rotator cuff lesion when working with elevated arms over many years (factor 2.1 when exposed for over ten years; Dalbøge et al., 2014).

5. Preventive measures to reduce musculoskeletal shoulder load

In summary, there is a clear relationship between the cumulative performance of industrial tasks at or above shoulder level, increased MSL in the shoulder, and the development of shoulder disorders. Therefore, preventive measures must be implemented to eliminate or to reduce shoulder MSL in order to preserve the musculoskeletal health of the workers. However, if MSL in the shoulder cannot be avoided when performing industrial tasks at or above the shoulder level, targeted preventive measures must be taken to minimize MSL according to the STOP principle (DGUV, 2023). The actions mentioned in the acronym must be applied in a hierarchical order: **S**ubstitution, **T**echnical controls, **O**rganizational controls, and **P**ersonal protective control (Figure 7; Steinhilber et al., 2020).

Substitutional measures aim to replace the work steps that impose increased load on the musculoskeletal system with non-hazardous work steps. This approach benefits all workers by offering a low degree of individual variability and a high level of safety (Figure 7). If substitution is not feasible, the load must be minimized as much as possible. One example of a substitutional measure is changing the bonding technique from overhead riveting to adhesive bonding, in order to avoid imposing additional load on the shoulders due to the weight of the tools (DGUV, 2023). Technical controls summarize all measures aimed at reducing MSL through workplace adjustments (e.g. adjusting the working height to reduce the shoulder elevation angle) or lifting aids (Figure 7; DGUV, 2023; EU-OSHA, 2019; Punnett et al., 2000; Steinhilber et al., 2020). Previous studies have demonstrated that overhead task require less muscles effort when the workstation is adapted to the individual anthropometry of the worker (compared to a fixed working height) or when the hand is closer to the body (Anton et al., 2001; Chopp et al., 2010a; Maciukiewicz et al., 2016). Organizational controls focus on changes to the work process, e.g. reducing load weight by using smaller containers, optimizing breaks, or applying job rotation (Figure 7; DGUV, 2023; EU-OSHA, 2019; Steinhilber et al., 2020).

If technical or organizational controls are not applicable, personal protective controls can contribute to reducing MSL including occupational healthcare programs and personal protective equipment (PPE; Figure 7; DGUV, 2023; Steinhilber et al., 2020). However, these are often limited to individual solutions, particularly wearable devices such as occupational exoskeletons (high individuality, low safety level; Figure 7).

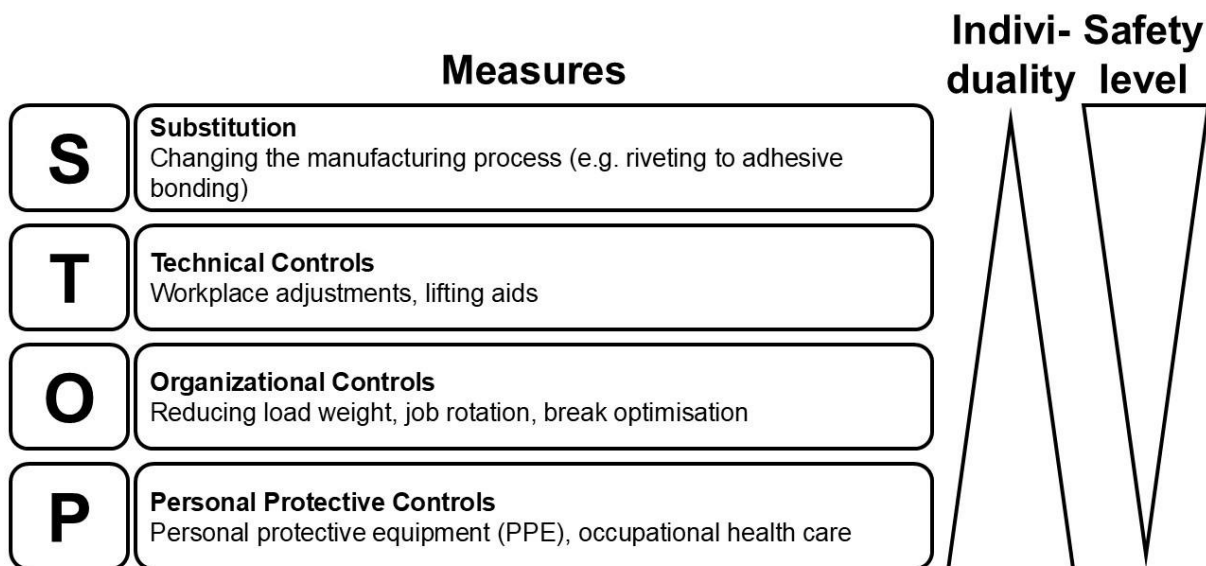


Figure 7: STOP – principle for the prevention or reduction of musculoskeletal load at the workplace (DGUV, 2023; Steinhilber et al., 2020).

6. Potential of occupational exoskeletons in reducing work-related musculoskeletal load

In recent years, the use of occupational exoskeletons has been evaluated as a potential ergonomic intervention to reduce the musculoskeletal load (MSL) during various working tasks (e.g. work performed at or above shoulder level), with the aim of decreasing WMSDs (Bär et al., 2021; de Looze et al., 2016; de Vries & de Looze, 2019; McFarland & Fischer, 2019; Moeller et al., 2022; Theurel & Desbrosses, 2019). In general, occupational exoskeletons are defined as wearable, assistive devices that mechanically support the user in performing work tasks by reducing the mechanical load at specific body parts (e.g. the shoulder or the spine; de Looze et al., 2016; de Vries & de Looze, 2019; Steinhilber et al., 2020; Theurel & Desbrosses, 2019). They provide a support force to the assisted body part, reducing the joint moment that has to be generated by the muscles to maintain a static posture or to move the body part (de Looze et al., 2016; de Vries & de Looze, 2019). Consequently, the required muscle forces, and thus the internal load on the joints, ligaments, tendons and soft tissues (e.g. the intervertebral discs), decrease (de Looze et al., 2016; de Vries & de Looze, 2019).

Occupational exoskeletons can be classified based on their energy supply, i.e. how the support moment is generated, into active, passive or semi-passive systems (Bär et al., 2021; de Looze et al., 2016; de Vries & de Looze, 2019; McFarland & Fischer, 2019; Moeller et al., 2022; Theurel & Desbrosses, 2019). Active exoskeletons have force/energy-generating elements (actuators), such as electric motors, pneumatic muscles, or hydraulic cylinders, which provide additional energy and improve the user's performance (de Looze et al., 2016; Gopura & Kiguchi, 2009; Steinhilber et al., 2020; Toxiri et al., 2018). Passive systems use structures such as springs or elastic materials (e.g. expanders) to store the force/energy generated by the user's motion (static and dynamic) which is then released back to the supported body part to help maintain certain postures (e.g. forward bending or static arm elevation) and to provide support during motion (e.g. lifting an object or working overhead; Bär et al., 2021; de Looze et al., 2016). Semi-active exoskeletons incorporate both active and passive elements implemented into their design (Otten et al., 2018).

Occupational exoskeletons can also be categorized based on the body part or joint they support and relieve (e.g. the upper or lower extremities, the trunk, or the full body; Bär et al., 2021; de Looze et al., 2016; McFarland & Fischer, 2019; Moeller et al., 2022; Steinhilber et al., 2020). Finally, exoskeleton designs can be divided into anthropomorphic exoskeletons, which are designed according to human anthropometry and include rotational axes aligned with human joint axes, and non-anthropomorphic exoskeletons which are generally simpler designed and optimized for specific tasks (de Vries & de Looze, 2019; de Looze et al., 2016; Steinhilber et al., 2020).

6.1. Occupational exoskeletons for the upper extremities

Occupational exoskeletons for the upper extremities have been developed to support or relieve the upper limbs during tasks that require elevated arm positions, such as at shoulder level or overhead (Bär et al., 2021). These devices aim to reduce the muscle demand on the shoulder muscles, such as the deltoideus and the rotator cuff muscles, by decreasing the muscle force required to generate the shoulder joint moment necessary for arm elevation (de Vries & de Looze, 2019). This reduction in muscle demand may help to reduce or prevent the development of shoulder muscle fatigue during these industrial tasks (Del Ferraro et al., 2020; McFarland & Fischer, 2019), thereby potentially minimizing muscle-fatigue-induced changes in glenohumeral joint kinematics and superior humeral head translation (Chen et al., 1999; Chopp et al., 2010b; Ebaugh et al., 2006; Grieve & Dickerson, 2008; Page, 2011). Furthermore, this could result in a decrease of WMSDs in the shoulder (de Vries & de Looze, 2019; de Looze et al., 2016; McFarland & Fischer, 2019) and reduce the days of incapacity to work. As previously mentioned, upper extremity exoskeletons can be categorized by their energy supply as either active or passive systems (de Vries & de Looze, 2019; McFarland & Fischer, 2019). Passive systems are lighter, primarily due to the absence of an additional external supply (de Vries & de Looze, 2019), and are therefore preferred to active systems, and have been analyzed in various studies (Bär et al., 2021). Passive exoskeletons for the upper extremities include, based on their design, shoulder systems and tool support systems with supernumerary limbs (de Vries & de Looze, 2019; McFarland & Fischer, 2019; Moeller et al., 2022). The latter systems (e.g. the Fortis exoskeleton) feature an additional mechanical arm onto which the tool required for a specific work task can be mounted (McFarland & Fischer, 2019).

This arm is fixed to the hip or to the torso, transferring the tool's weight to these body parts (McFarland & Fischer, 2019; Moeller et al., 2022). However, it has been shown that these systems can increase the spinal load and the muscle activity in the torso extensor and lower back muscles due to the additional weight and inertial properties of the mechanical arm (Alabdulkarim & Nussbaum, 2019a; Weston et al., 2018) and are therefore less preferred. In contrast, shoulder systems support the arms directly during elevation by applying a support force via arm cuffs to the arm, e.g. at the upper arm or at the wrist (Figure 8; McFarland & Fischer, 2019; Moeller et al., 2022; Theurel et al., 2018).

Passive occupational shoulder exoskeletons (OSEs) aim to reduce the gravitational shoulder joint moment $\tau_{grav}^{shoulder}$ with a support moment τ^{exo} generated via passive, elastic, force-storing structures, such as carbon fibers or elastic expanders, to obtain the net shoulder joint moment (NSM):

$$NSM = \tau_{grav}^{shoulder} - \tau^{exo}. \quad (6.1)$$

When the arm is lowered, this structure extends, generating a retraction force F_E to restore its initial length (Figure 8). As the passive structure is connected to the arm segment of the OSE (where the arm cuff is attached), this retraction force creates a support moment around a rotational hinge joint (exoskeleton joint center; JC_{Exo}) over an internal lever arm $r_{internal}$:

$$\tau^{exo} = F_E * r_{internal}. \quad (6.2)$$

This support moment is then transmitted as a force F_A over an external lever arm $r_{external}$ to the upper arm via the arm cuff, thereby supporting the muscles during arm elevation (Figure 8):

$$F_E * r_{internal} = F_A * r_{external}. \quad (6.3)$$

The support force, and consequently the support moment, depend on the elevation angle of the arm and can be manually adjusted to achieve an optimal fit according to the task demands (de Vries et al., 2019; Musso et al., 2022; Watterworth et al., 2023).

Lastly, the exoskeleton is fixed at pelvis height above the gluteal muscle to transfer the weight of the arms onto the hips (Figure 8).

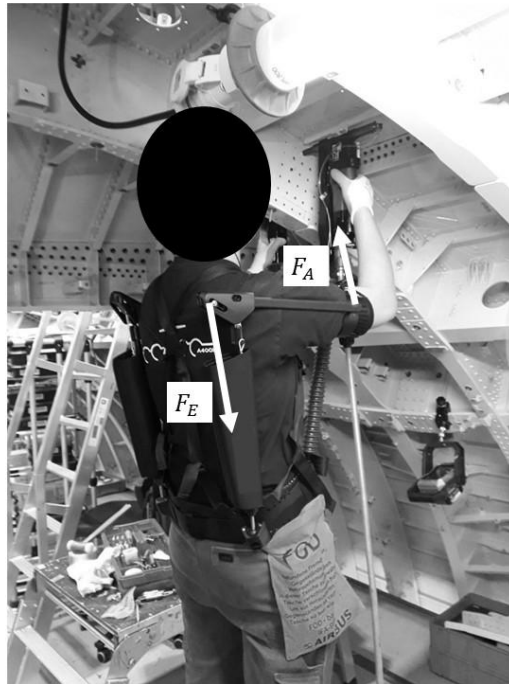


Figure 8: Example of the functional principle of an occupation shoulder exoskeleton (OSE). The exoskeleton generates a force F_E , which in turn generates a corresponding support moment over the internal lever arm of the exoskeleton. This support moment then transmits a corresponding force F_A to the upper arm via an external lever arm. This causes a reduction in the shoulder joint moment, which is intended to reduce the demand on the shoulder muscles (modified photo: © Airbus Defence & Space GmbH, 2021).

6.2. Influence of occupational shoulder exoskeletons on musculoskeletal shoulder load

Previous laboratory studies have already revealed the benefiting effects of OSEs on the musculoskeletal load in the shoulder by reducing the shoulder muscles demand, either during static arm elevation (Brunner et al., 2023; Desbrosses et al., 2021; de Vries et al., 2019; Tetteh et al., 2022) or during simulated industrial tasks performed overhead or at/around shoulder level, such as riveting (Jorgensen et al., 2022a), sealing (Jorgensen et al., 2022b), drilling (Alabdulkarim & Nussbaum, 2019a; Alabdulkarim et al., 2019b; de Bock et al., 2022; Kim et al., 2018; Kim & Nussbaum, 2019; Kong et al., 2023; Schmalz et al., 2019), screwing and tracing with a drill (van Engelhoven et al., 2019), screwing with heavier tools (Brunner et al., 2023; Pinho & Forner-Cordero, 2022), screwing by hand or light tools (Brunner et al., 2023; Musso et al., 2024; Schmalz et al., 2019), wiring (de Bock et al., 2022; Kim et al., 2018; Kim & Nussbaum, 2019; van der Have et al., 2022) and mounting with a nutrunner (Ojelade et al., 2023). In particular, studies have shown that wearing an OSE during static arm elevations and industrial overhead/shoulder tasks reduces the shoulder muscle activity for both commercially available and prototype OSEs (Brunner et al., 2023; Desbrosses et al., 2021; de Vries et al., 2019; Pacifico et al., 2020). De Vries et al. (2019) reported a reduction in shoulder joint moment and muscle activity (in percentage to the maximal activity of the respective muscle, obtained during reference trial with a maximal isometric voluntary contraction – MVC) in the anterior deltoideus (AD) and upper trapezius (UT) muscles when wearing an OSE (Skelex) during static arm elevations. Similarly, Desbrosses et al. (2021), Brunner et al. (2023) and Tetteh et al. (2022) found reduced muscle activity in the shoulder muscle (from 8.6 %MVC to 21.6 %MVC for the AD, 3.45 %MVC to 9.6 %MVC for the UT and 1.3 %MVC to 10.8 %MVC for the medial deltoideus [MD]) when performing different static arm postures using the Skelex, the Delta Suit and the Airframe exoskeletons, respectively.

Various studies have found beneficial effects in reducing muscle demands in the shoulder, as indicated by reduced muscle activity in AD, MD and UT, when performing a simulated drilling task in the laboratory with an OSE at shoulder height or overhead (both standardized and non-standardized) under different load conditions (i.e. tool weights; Alabdulkarim & Nussbaum, 2019a; Alabdulkarim et al., 2019b; de Bock et al., 2022; Kim et al., 2018; Kong et al., 2023; Schmalz et al., 2019).

However, some of these studies also reported adverse effects on muscle demand in the arms, with a higher activity in the triceps brachii during drilling with an OSE (Alabdulkarim & Nussbaum, 2019a; de Bock et al., 2022). Further laboratory studies have shown that wearing an OSE can also reduce shoulder muscle activity during similar tasks (here: screwing), even when the external load resulting from the weight of the tool differs (Brunner et al., 2023; Musso et al., 2024; Pinho & Forner-Cordero, 2022; Schmalz et al., 2019; van Engelhoven et al., 2019). Brunner et al. (2023), Musso et al. (2024) and Schmalz et al. (2019) all found reduced muscle activity in the anterior and the medial deltoideus (up to 45% for the AD) when screwing overhead with the hands or a light tool. Brunner et al. (2023) also reported lower muscle activity in the AD, MD and UT when using their OSE (DeltaSuit) to screw with a power tool (1.8 kg). Although the magnitude of this reduction was lower in all three muscles compared to the condition without the OSE (e.g. mean difference for the AD = 15.6 %MVC when screwing with the hands, mean difference for the AD = 13.3 %MVC when screwing with the power tool; Brunner et al., 2023). Similarly, van Engelhoven et al. (2019) found reduced muscle activity for the anterior deltoideus and the upper trapezius when using two tool weights (0.45 kg and 2.25 kg) to screw overhead with an OSE (ShoulderX). This reduction increased up to 18 %MVC in AD by increasing the support moment provided by the OSE (from 5 Nm to 15 Nm; van Engelhoven et al., 2019). However, the muscle activity of the antagonist muscles (posterior deltoideus, latissimus dorsi, and triceps long head) increased with the highest support moment (15 Nm), independent of the tool weight or the performed task (van Engelhoven et al., 2019). Therefore, it is preferable to adjust the OSE's support moment according to the specific task demands (van Engelhoven et al., 2019).

When working with heavier tools, it can be concluded that a higher support moment is required to effectively reduce shoulder muscle activity (in AD and MD). This assumption is supported by the results of Pinho and Forner-Cordero (2022), who reported a reduction in AD and MD muscle activity when working with a pneumatic screwdriver (5.02 kg) in different arm postures. A higher support moment is provided from their OSE to the user with a peak support moment of approx. 20 Nm (as shown in the figures of Pinho & Forner-Cordero, 2022) compared to the study of van Engelhoven et al. (2019) with a support moment of 15 Nm.

For industrial overhead and shoulder tasks with low task demands, such as wiring without an additional tool, decreasing the support moment of the OSE can still be beneficial in reducing musculoskeletal shoulder load (e.g. shoulder joint moment and muscle activity; de Bock et al., 2022; van der Have et al., 2022). However, the effect of different OSEs can vary for the same task, even when the general functionality (i.e. initiation of a support force onto the upper arm) is identical (Jorgensen et al., 2022a; Jorgensen et al., 2022b; Ojelade et al., 2023). Jorgensen et al. (2022a; 2022b) and Ojelade et al. (2023) both reported different effects on shoulder muscle activity when using three different OSEs to perform the identical industrial task. These findings are important to consider when implementing OSEs in the field, i.e. at the workplace.

Beneficial effects of OSEs in reducing shoulder muscle demand have also been found when used at the workplace (field studies) across various sectors including the automotive industry (Gillette & Stephenson, 2019; Gillette et al., 2022; Iranzo et al., 2020), a gas company (Pentenga et al., 2024), multiple construction sites (Mänttari et al., 2024), a slaughterhouse (Dalbøge et al., 2024) and in the surgery room (Cha et al., 2024). In addition, similar effects were reported when using an OSE in real-world work scenarios performing different tasks such as plastering (de Vries et al., 2021), cleaning surgical instruments (Arnoux et al., 2023), manual material handling (Schrøder Jakobsen et al., 2023), wool textile processing (Coccia et al., 2024) and window and ceiling cleaning (Pacifico et al., 2023). Dalbøge et al. (2024) found an average reduced bilateral muscle activity when working in a slaughterhouse with an OSE (ShoulderX) in the AD and the UT by 1.4 %MVC (28.1%) and 1.8 %MVC (21.6%), respectively. They also evaluated the OSE against a technical measure, a lifting aid (a lifting glove), which showed a higher reduction in right AD muscle activity than the OSE (36.6%; Dalbøge et al., 2024). However, they also found an increased activity in the left AD (15.6%) when using the lifting glove compared to when working without an assistive device (Dalbøge et al., 2024). Field evaluations of OSEs generally showed lower reductions in the shoulder muscle activity, mostly between 1.4 %MVC (Dalbøge et al., 2024) and 6.4 %MVC (Arnoux et al., 2023), than evaluations in the laboratory (up to 18 %MVC; van Engelhoven et al., 2019). Pacifico et al. (2022) reported a difference in the magnitude, finding higher reductions in total shoulder muscle activity when using an OSE during similar work activities performed in a simulated laboratory environment compared to a field setting.

Furthermore, de Bock et al. (2021) also reported discrepancies in the effects of two passive OSEs, with beneficial effects on muscle demand when tested under controlled laboratory conditions during isolated work activities, but only limited effects when used in a real work scenario. Therefore, while laboratory studies are necessary to acquire valuable information concerning the general effectiveness of OSEs in reducing musculoskeletal load in the shoulder during shoulder/overhead work, the transferability of these findings to real workplaces is clearly limited. Consequently, a combination of laboratory and field studies, such as conducted by de Bock et al. (2021) and Pacifico et al. (2022), will be better for evaluating the effectiveness and implementation of OSE at the workplace.

To further quantify the biomechanical effectiveness of OSEs in reducing MSL in the shoulder, it is necessary, to estimate their effect on the shoulder joint moment, as these results would provide further insight into their influence on the mechanical load at the GHJ (Praagman et al., 2000). Previous studies have demonstrated the beneficial effects of OSEs in reducing the shoulder joint moment and, conversely, mechanical load during static arm elevation (de Vries et al., 2019) and industrial tasks performed at or above shoulder level (Fritzsche et al., 2021; Musso et al., 2022; Seiferheld et al., 2022; Sylla et al., 2014; van der Have et al., 2022). However, these studies were either primarily conducted in a laboratory setting with a standardized measurement setup (e.g. individual working heights or arm postures), or with a prototype OSE (de Vries et al., 2019; Fritzsche et al., 2021; Sylla et al., 2014; van der Have et al., 2022). Furthermore, some studies did not collect kinematic data when working with the OSE (Seiferheld et al., 2022), while others simulated the use case for the OSE for one person only (Musso et al., 2022). Therefore, the effect of OSEs on the shoulder joint moment in real working scenarios including tasks performed at or above shoulder height needs to be further analyzed. In addition, to estimate the effect of an OSE on the shoulder joint moment, the support moment provided by the OSE has to be known. This can be done by either directly measuring the moment (or the support forces) when the OSE is applied to the user (e.g. through integrated force sensors in the arm cuff; de Vries et al., 2019) or indirectly by estimating the moment-angle curve in a standardized measurement setup (e.g. with the OSE fixed to a dynamometer; Watterworth et al., 2023). In the latter case, the support moment is simulated during post-processing (Musso et al., 2022; van der Have et al., 2022), based on the kinematics of the user (e.g. the shoulder elevation angle).

Finally, the effect of an OSE on local shoulder muscle tissue oxygenation (MTO) has only been analyzed in one study (Weston et al., 2022), meaning that it is not fully understood yet. A deeper analysis would surely provide more sensitive knowledge on its effect on shoulder muscle demand (Del Ferraro et al., 2020; Ferguson et al., 2013; McFarland & Fischer, 2019; Weston et al., 2022).

7. Aim of the thesis

Occupational shoulder exoskeletons (OSEs) represent a promising ergonomic intervention to reduce work-related musculoskeletal disorders (WMSDs) of the shoulder, such as rotator cuff lesions, impingement syndrome or tendonitis. These disorders often result from increased musculoskeletal load (MSL), i.e. increased muscle demand and mechanical load at the glenohumeral joint (GHJ), during tasks performed at or above shoulder level. The OSE counteracts this load by reducing the shoulder joint moment required to elevate the arm, thereby reducing shoulder muscle forces and consequently decreasing shoulder muscle demand.

While the effects of OSEs on shoulder muscle demand have been extensively investigated in both laboratory and field studies, there remains a lack of understanding their influence on the mechanical load at the GHJ and local muscle tissue oxygenation (MTO). These factors could contribute to a deeper insight into the effect of OSEs on shoulder muscle forces and provide valuable information on their influence on MSL. The overall aim of this thesis was therefore to conduct an exploratory analysis of the biomechanical effectiveness of OSEs in reducing MSL in the shoulder during tasks performed at or above shoulder level.

To address this aim, four separate studies were conducted with the following specific objectives:

- 1) Quantification of the support moment generated by three occupational shoulder exoskeletons across their respective support range and for all available support levels.
- 2) Exploratory analysis of the influence of an occupational shoulder exoskeleton on the mechanical load at the GHJ and shoulder muscle demand during static arm elevation.
- 3) Exploratory analysis of the influence of an occupational shoulder exoskeleton on the mechanical load at the GHJ and muscle tissue oxygenation during static arm elevation.
- 4) Exploratory analysis of the influence of three occupational shoulder exoskeletons on the mechanical load at the GHJ and shoulder muscle demand during aircraft manufacturing tasks in a real working scenario.

8. Quantification of the support moment of three occupational shoulder exoskeletons

The aim of this study was to quantify the support moment generated by three different occupational shoulder exoskeletons (OSE) across their respective support range and for all available support levels (SL). Furthermore, the data obtained was used to calculate the support moment as a function of the exoskeleton angle in order to further estimate the support moment provided by the OSE to the user in laboratory settings and under real working conditions. The OSEs analyzed in this study and subsequently the thesis were the Paexo Shoulder (V1, Ottobock SE & Co. KGaA, Duderstadt, Germany), the Skelex 360° (Version 1.1, Skel-ex B.V., Den Haag, Netherlands) and the Levitate Airframe (Levitate Technologies, Inc., San Diego, California). It should be noted that only the support levels 3 and 4 of the Airframe were available and therefore included in the study.

8.1. Measurement setup and preparations

To estimate the support moment under a standardized measurement protocol, a collaborative robot (COBOT; UR5 CB3, Universal Robots A/S; Odense, Denmark; Figure 9) was used to perform repetitive dynamic arm elevations at a predefined rotational velocity of 10 °/s across the full support range (determined individually for each exoskeleton). This velocity was derived from an investigation conducted by the Institute for Occupational Safety and Health of the German Social Accident Insurance (IFA) using biomechanical measurement systems, which involved analyzing various work tasks that increase musculoskeletal shoulder load. The COBOT arm thus served as a dummy to replicate the user's arm (Figure 9). In contrast to previous studies where the support moment was either measured directly using a dynamometer (Watterworth et al., 2023) or estimated using mathematical modelling (Musso et al., 2022), the approach in this study was to measure the contact forces at the arm cuff, similar to de Vries et al. (2019) and Hartmann et al. (2021). However, instead of integrating a force sensor into the arm cuff, as done by de Vries et al. (2019), the force sensor was mounted to a ring that could be moved along the COBOT arm allowing the sensor's position relative to the arm cuff to be adjusted individually for each exoskeleton (Figure 9 A).

Furthermore, in contrast to de Vries et al. (2019) and Hartmann et al. (2021), a 6D force sensor (1000 Hz; FTN-Mini-45, Schunk GmbH & Co. KG, Lauffen/Neckar, Germany) was used to measure not only the vertical forces in the arm cuff but also the horizontal (shear) forces. The angle between the force sensor and the ring could be adjusted to ensure vertical alignment of the sensor with the arm cuff during dynamic elevations of the COBOT arm (Figure 9 A).

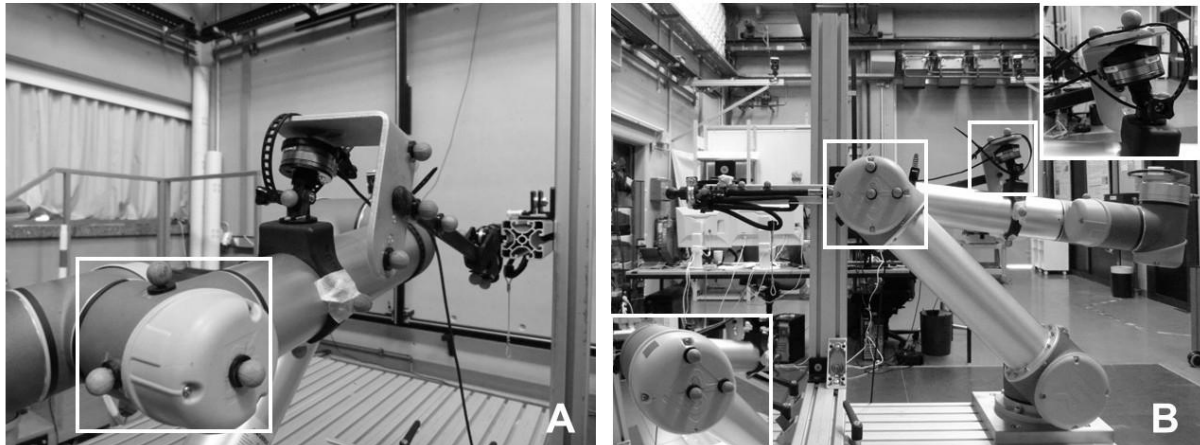


Figure 9: Measurement setup of the collaborative robot (COBOT) with an attached force sensor (B; white square, right upper corner), retro-reflective markers placed the arm (A; white square) and shoulder segment (B, white square, left lower corner) and an exemplary exoskeleton.

The top of the sensor was fitted with a self-constructed cone to minimize the contact point between the sensor and the arm cuff (Figure 10), and the tip of the cone was oriented in the same direction as the vertical axis of the force sensor (z-axis; Figure 10 A & C). In addition, four notches were integrated in a circular pattern at the bottom of the top piece, with two notches facing each other and aligned parallel to the horizontal axis of the force sensor (x and y axes; Figure 10 B & D).

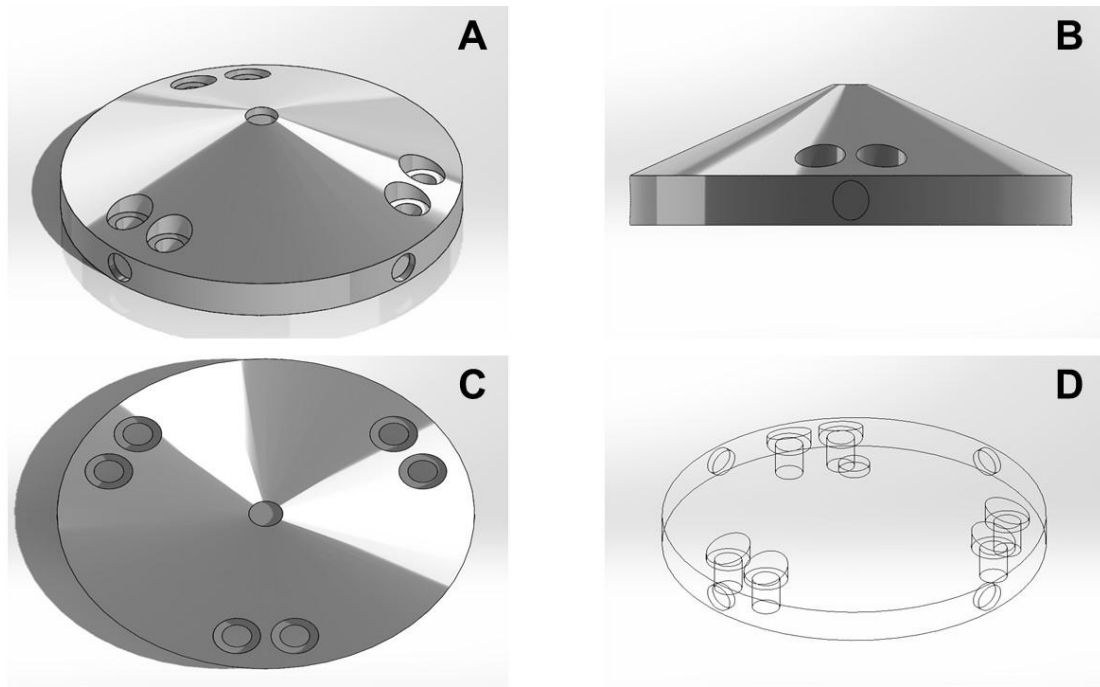


Figure 10: Illustrations of the self-constructed top of the force sensor (visualized in Solidworks; version 2022, Dassault Systemes Deutschland GmbH, Stuttgart, Germany) in isometric view as full (A) and see through model (D) as well as in side (B) and top view (C).

Prior to a measurement session, each OSE was disassembled into two parts: the lower part (the “arm segment”) with the arm cuff connecting the COBOT arm to the OSE, and the upper part fixed to the measurement table (the “shoulder segment”). During this process, it was ensured that only components not required (1) to generate the support moment or (2) to transmit the support forces to the arm, such as the hip belt or the shoulder straps, were disassembled. Therefore, each OSE was visually inspected by the study personnel to identify the passive force-generating structures, the upper and lower part and the rotational hinge joint connecting the two parts.

For example, the Airframe was disassembled at the rotational joint above the hinge joint, and the metal rod leading to the hinge joint was identified as the upper part, i.e. the “shoulder segment” (Figure 11 A). The Paexo and the Skelex were disassembled only at the hip belt (Figure 11 B & C). The passive force- generating structures were identified as a carbon leaf fiber for the Skelex and two expander cables for the Paexo (Figure 11 B & C). The structures for the Airframe could not be visually identified as they were concealed inside a black cover (Figure 11 A). This black cover (“cassette”) was not dissembled, as it was unclear whether a disassembly could affect the internal mechanism that generates the support moment. In the case of the Paexo, the protective cover around the passive force-generating structures, i.e. the expanders, was also removed in order to facilitate a more secure fixation of the OSE. In addition, the arm cuffs for these OSEs were replaced with a self-produced metal bracket because (a) the arm cuff for the Skelex only functions when the arm straps are applied (exerting pressure on the force sensor causing inaccurate detection of the support force), and (b) the bottom of the Paexo arm cuff is uneven making it impossible to reliably estimate the point of application of the support force (Figure 11).

Measurements were conducted for only one side of each OSE, assuming identical moment angle profiles for both sides. After disassembly, the upper part of each OSE was fixed in a horizontal position parallel to the measuring table, and the position of the force sensor and the angle relative to the arm cuff were adjusted (as described above). The rotational axes of the hinge joint of the OSE and the COBOT were aligned in order to reduce the horizontal (shear) forces generated by the OSE during the measurements. For the Paexo and the Skelex, an additional strap was placed around the arm cuff and below the force sensor, as the arm cuff would lose contact with the COBOT arm if no support was provided at the lowest exoskeleton angle. To ensure that the strap did not influence the measured contact forces during the dynamic elevations which would lead to incorrect calculations of the support moment, the strap was attached as loosely as possible (Figure 11 C). Prior to each measurement, the setup was visually inspected by the study personnel to ensure that there was no contact between the force sensor and the arm cuff.

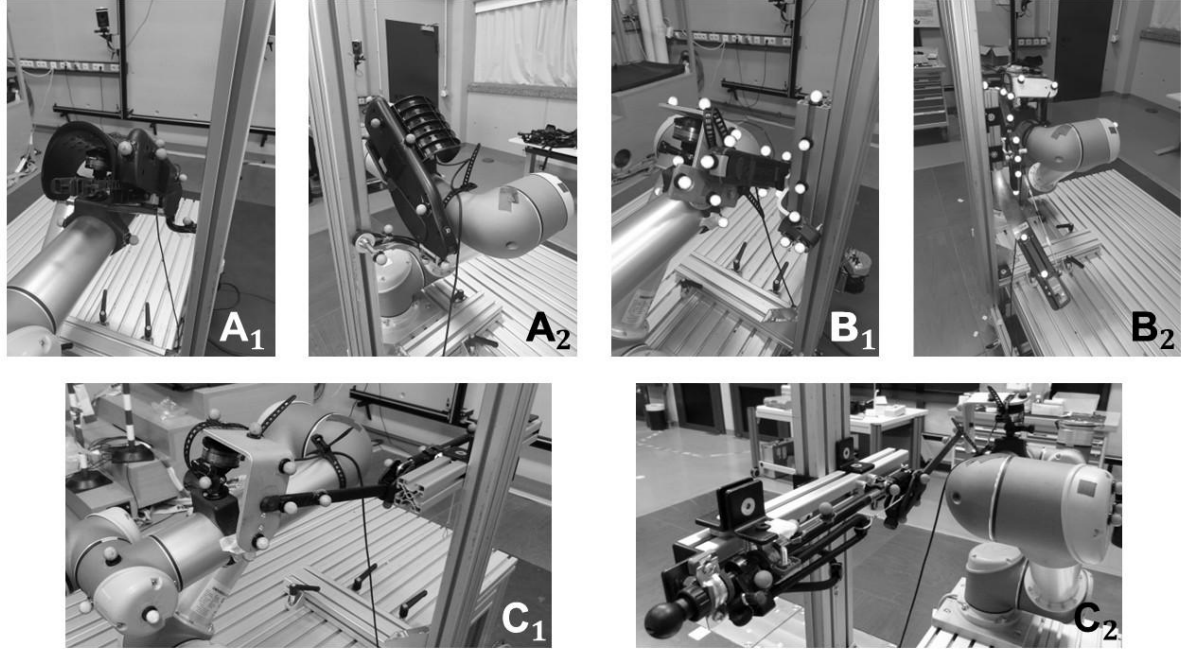


Figure 11: Measurement setup with the collaborative robot (COBOT) for the three tested occupational shoulder exoskeletons (OSE); Levitate Airframe (A), Skelex 360 (B) and Paexo Shoulder (C); in front and back view. Each exoskeleton and the COBOT were equipped with retro-reflective markers (at least three per segment).

A motion protocol was defined in the COBOT software, specifying the maximum (T_{max}) and minimum angular positions (T_{min}) of an elevation performed by the COBOT. T_{max} and T_{min} were defined as the maximum and minimum support angles of each exoskeleton, respectively (Figure 12). An initial angular position (T_{int}) at (approximately) 5° above T_{max} was also defined and included in the protocol to ensure that no contact occurred between the OSE and the sensor prior to a measurement (Figure 12). An additional reference angular position was defined where the angle between the upper and lower part of the OSE was (approximately) 90° (T_{90} ; Figure 12). After implementing the motion protocol, the elevation was performed repeatedly while the contact between the sensor and the arm cuff was visually inspected. If a clear contact was detected exclusively between the top of the sensor and the arm cuff, the motion protocol was approved. To measure the kinematics of the COBOT and the OSE, preparations were completed by placing retro-reflective markers (diameter 14 mm) on the COBOT's arm (3 markers), on the adjacent segment of the COBOT (3 markers; COBOT's "shoulder" segment), on the ring where the sensor was attached to (4 markers), and on the upper and lower part of the OSE (at least 3 markers per part).

The 3D trajectories of the markers were measured with 12 Vicon cameras (100 Hz; Vero, Vicon Motion Systems, Inc., Oxford, UK) and the force sensor data was synchronized with the marker data using the software ViconNexus (Version 11, Vicon Motion Systems, Inc., Oxford, UK).

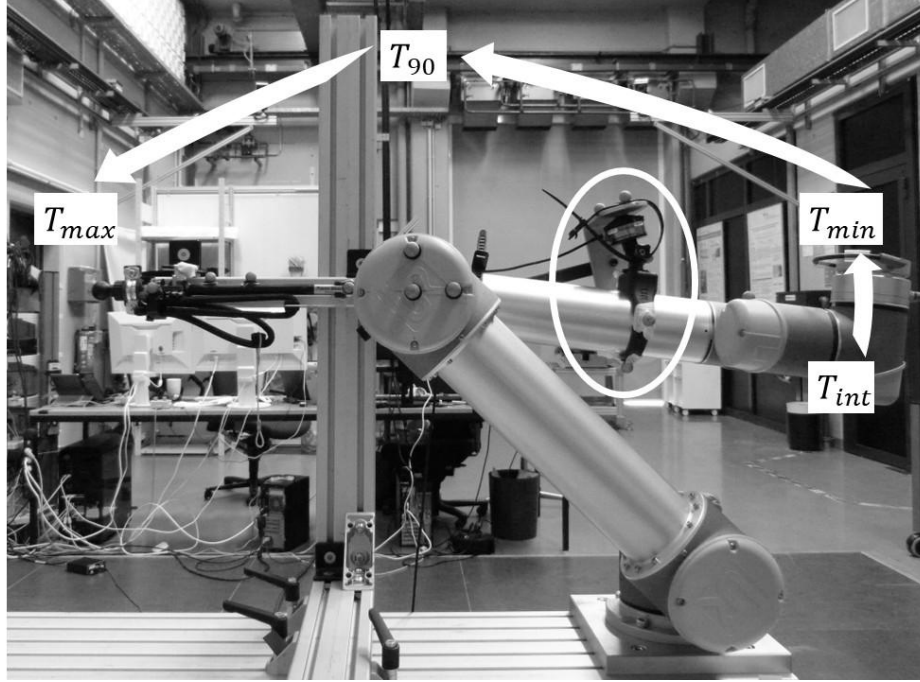


Figure 12: Measurement setup of the collaborative robot (COBOT) with an attached force sensor (white circle), an exemplary exoskeleton and the four angular positions (T_{int} , T_{min} , T_{90} and T_{max}) which were defined in each measurement session.

8.2. Measurement session

Each measurement session for an OSE began with a reference trial in which ten repetitive dynamic elevations were performed from T_{max} to T_{min} according to the motion protocol without the exoskeleton (“NoExo” trial). This reference trial was performed to estimate the offset forces resulting from the sensor mass and its moment of inertia, as these would influence the forces measured at the arm cuff during the OSE trials. For the Paexo, an additional reference trial was conducted after attaching the OSE to the measurement setup, with both expanders disconnected, i.e. no support force was generated by the OSE (“NoSupport” trial). This enabled the measurement of strap-induced forces acting on the arm cuff and the sensor. However, this procedure was not applicable to the Skelex. After completing the reference trials, additional calibration trials were carried out to estimate specific points on the OSE and the force sensor that could not be tracked by the cameras during dynamic elevation trials, either because the area above these points was too small for a marker or because it was covered during a measurement. It was concluded that the global positions of these points were essential for calculating the support moment and angle of the OSE. These points included:

- (a) the top of the force sensor, i.e. the point of force application between the sensor and the arm cuff and aligned with the z-axis of the sensor;
- (b) the four notches at the bottom of the top of the force sensor, aligned with the positive and negative x- and y-axes of the sensor; and
- (c) two points outside of the rotational joint center of the OSE located opposite to each other and collinear to the exoskeleton joint center.

To reconstruct the global positions of these points during a measurement, their local positions relative to the local coordinate system of the respective rigid bodies on which they were placed on were obtained during the calibration trials (Figure 13, Robertson et al., 2014). These rigid bodies were identified as the COBOT ring (for the five points at the top of the force sensor) and the upper part of the OSE (for the two points outside of the OSE rotational joint center). Therefore, the tip of a rigid self-constructed pointer equipped with four reflective markers was successively placed on these points for ten seconds while the study personnel moved as “little as possible” (Figure 13).

Prior to the calibration trials, the global position of the pointer tip was estimated during another calibration trial, so that its global position could be estimated in the later calibration trials using the reference system constructed from the four markers placed on the pointer (Figure 13; Robertson et al., 2014). Since both the ring on the COBOT arm and the upper part of each OSE were equipped with at least three markers, each of the seven described global points (where the pointer tip was placed) could be reconstructed during data processing (Figure 13).

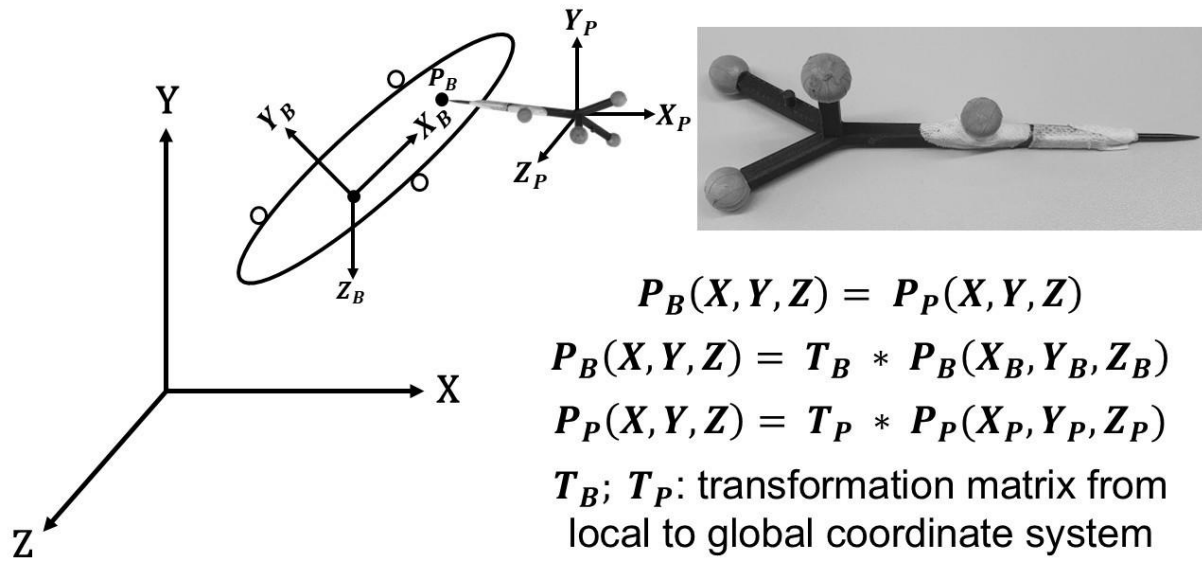


Figure 13: Exemplary illustration of the mathematical approach to reconstruct the global coordinates (X, Y and Z) of a point (P_B) on a rigid body equipped with three reflective markers (indicated by circles) from its local coordinates (X_B, Y_B, Z_B). The local coordinates were estimated by the global coordinates (X, Y and Z) of the top of the pointer (P_P). The global coordinates of the pointer top P_P were calculated from its respective local coordinates (X_P, Y_P, Z_P) obtained during a calibration trial.

Data collection continued with ten dynamic arm elevations performed according to the motion protocol for each OSE at a given support level. Prior to an elevation trial, the sensor position was checked to ensure that there was no contact between the sensor and the arm cuff. The force sensor was then reset in the Vicon Nexus software to eliminate offset forces. Each elevation trial then began with the COBOT remaining in its initial position (T_{int}) for ten seconds (baseline) to collect zero values for the force sensor. Afterwards, the COBOT rotated to the lowest angular position (T_{min}) which was defined in the motion protocol.

Then, the ten dynamic elevations were performed between the angular positions T_{min} and T_{max} with a one second pause at each of the two positions, respectively. The elevation trial was completed by rotating the COBOT arm back to its initial position T_{int} , followed by another ten seconds remaining in the initial position. Finally, a reference trial was performed in which the OSE remained in the reference angular position T_{90} for two minutes. Then, the support level (SL) of each OSE was increased and the elevation trial was repeated using the same procedure. For the Skelex and the Paexo, a higher level of support was expected by continuously increasing the distance between the center of the exoskeleton joint (JC_{Exo}) and the point where the retraction force generated by the passive structures (the leaf fiber for the Skelex and the expanders for the Paexo) was initiated (defined as the internal lever arm $r_{internal}$; Figure 14). The different levels of support were defined based on the marks placed around the internal lever arm resulting in five different support levels for the Paexo (PSL11-PSL15; similar to Watterworth et al., 2023; Figure 14 A) and seven for the Skelex (SxSL1-4; three additional levels were defined equally spaced between the four levels marked on the OSE, Figure 14 B). Five additional support levels were defined for the Paexo (PSL21-PSL25), as a second expander can be fitted to further increase the support moment.

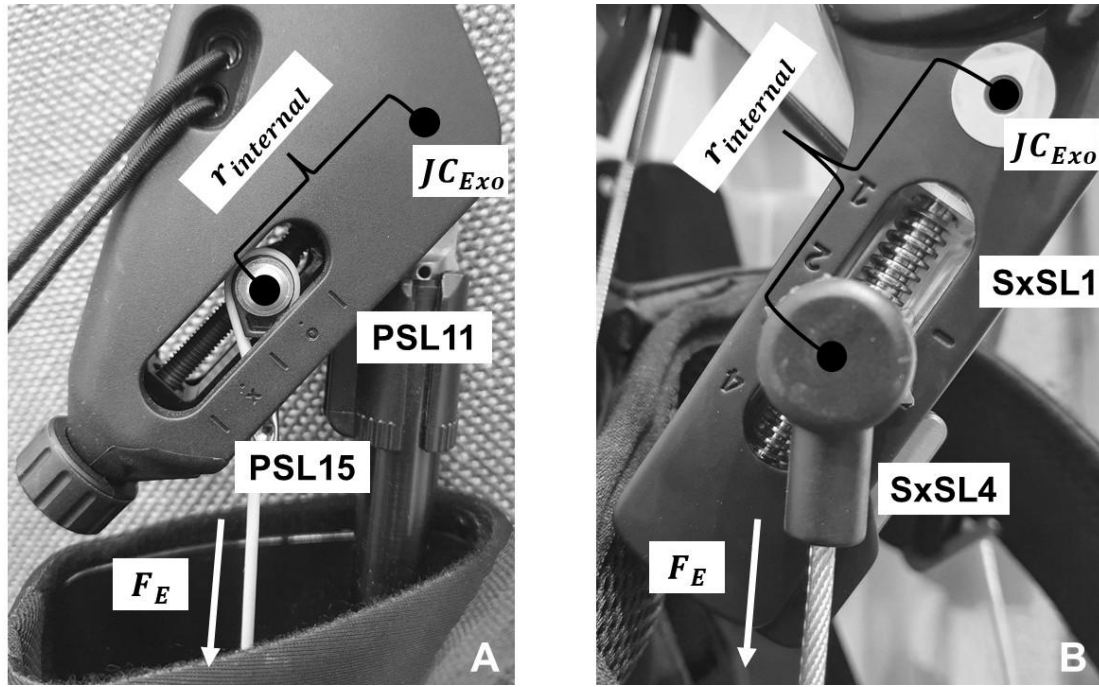


Figure 14: Mechanism to increase the support level (SL) for the exoskeletons Paexo (A) and Skelex (B) by increasing the distance between the point of force application and the rotational joint center of the respective exoskeleton (JC_{Exo}), i.e. the internal lever arm ($r_{internal}$). F_E is the retraction force generated by the passive structures of the exoskeleton (white arrow indicates the force direction). For both exoskeletons, two exemplary support levels are marked (PSL11 and PSL15 for the Paexo and SxSL1 and SxSL4 for the Skelex).

However, in order to increase the support moment of the Airframe from level 3 (ASL3) to level 4 (ASL4), the cassette containing the force generating passive structures had to be exchanged. Therefore, these two support levels were measured in separate measurement sessions, including the described calibration and reference trials.

8.3. Data processing

Data processing began in the ViconNexus software by calculating the rotational center and axes for the OSE and the COBOT for the elevation and reference trials (“NoExo”, “NoSupport” and T_{90}) using the SCoRE and SARA methods (Ehrig et al., 2006; Ehrig et al., 2007; Ehrig & Heller, 2019; Taylor et al., 2010). The results were exported and further data processing was performed using Python (Software Spyder, version 3.12, Python Software Foundation, Wilmington, USA) and the “Kinetic Toolkit” (Chénier, 2021). The 3D trajectories of the markers and the force data for all trials (including the calibration trials) were filtered (Butterworth, low pass, 5 Hz, 4th order) and 3D coordinate systems were defined for the lower (arm segment) and upper parts (shoulder segment) of the OSE and the COBOT arm, shoulder segment and the ring. Thus, the rotational center and the primary axis of rotation (obtained from the SCoRE and SARA methods) were defined as the origin and the first coordinate axis for the shoulder segments of the OSE and the COBOT. The joint kinematics between the arm and shoulder segments were then calculated individually for the OSE and the COBOT, and the joint angles were obtained using an XYZ Euler sequence. The exoskeleton angle was corrected for offset by shifting all values relative to the value obtained during the T_{90} -trial in which the OSE remained in static position. Then, the local coordinates of the seven points on the OSE and the COBOT, measured in the calibration trials, were exported and their global coordinates were reconstructed from their local coordinates. An additional coordinate system was calculated for the force sensor with the origin defined as the midpoint between the four points at the bottom of the sensor top and the first axis defined as the vector between the origin and the tip of the top (z-direction of the force sensor; Figure 15).

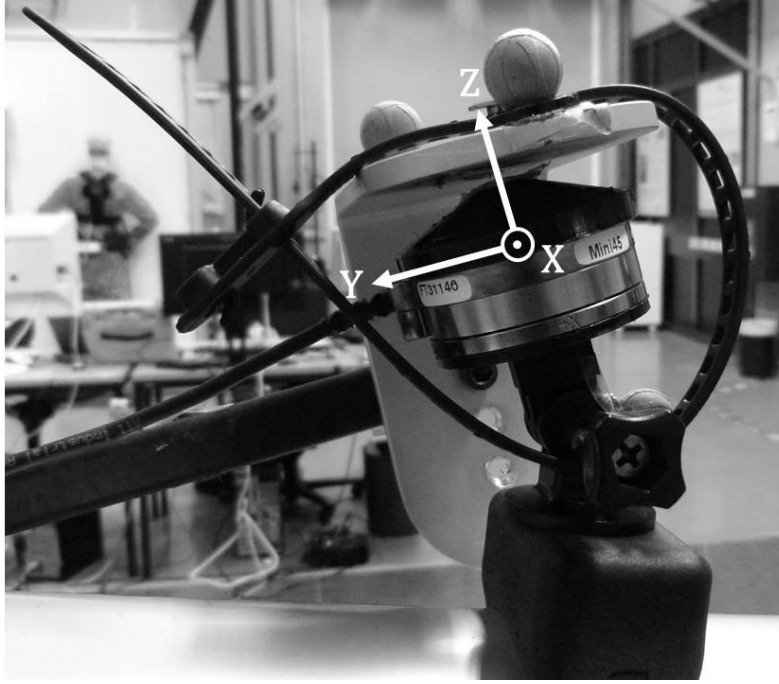


Figure 15: Force sensor attached to the ring on the collaborative robot (COBOT) for the measurement setup of the exoskeleton Paexo and the coordinate system of the sensor (X, Y and Z).

The measured forces were subsequently transformed into the sensor's coordinate system. Then, each repetition of a dynamic elevation and the upward (T_{min} to T_{max}) and downward (T_{max} to T_{min}) phases in the elevation and reference trials were identified using the "detect_cycles" function of the "Kinetics Toolkit" (Chénier, 2021) for the COBOT primary angle (Figure 16 A). The offset forces obtained during the "NoExo" trial were estimated for both phases relative to the COBOT primary angle and were subtracted from the elevation trials to obtain the 3D force vector. A similar approach was conducted for the "NoSupport" trial for the Paexo to eliminate the remaining forces that may have been initiated by the strap around the arm cuff and force sensor. As this procedure could not be performed for the Skelex, the offset in the vertical force for T_{min} (where the support force at the arm cuff should be close to zero) was estimated to be approximately 1 N (< 4% of the maximum force obtained during the lowest level of support SxSL1) and therefore decided to be negligible. The exoskeleton joint center was calculated as the midpoint between the two collinearly aligned global points outside of the joint center and the 3D external lever arm ($r_{external}$) was obtained as the vector pointing from the exoskeleton joint center to the contact point between the arm cuff and the sensor.

Consequently, the 3D support moment of the OSE (τ_{exo}) could be estimated by the cross product of the external lever arm and the 3D force vector (F_A):

$$\tau_{exo} = r_{external} \times F_A \quad (8.1.)$$

In order not to overestimate the support moment provided by the exoskeleton, only the sagittal component (around the X axis) of the 3D moment vector was further analyzed, as it was identified as the moment primarily responsible for the vertical support force in the arm cuff (Figure 16 A & B). The support moments in the remaining two directions (Y and Z) were not further analyzed as they primarily generate horizontal (shear) rather than vertical forces in the arm cuff. To estimate the support moment as a function of the exoskeleton angle for a given support setting in later laboratory and field studies presented in this thesis (see chapters 9 to 11), polynomial regression models were created. However, as already shown by Johns et al. (2024), hysteresis was observed for all three OSEs across all support levels. Consequently, the calculated support moment for a given exoskeleton angle depended on the movement phase (i.e. whether the COBOT arm was lifted upwards or downwards; Figure 16 B). This effect is highly relevant when analyzing the effect of the OSEs on the shoulder joint moment, as the provided support moment is expected to be lower when lifting the arm than when lowering it. Therefore, two separate polynomial regression models would have to be calculated for each of the two movement phases to adequately estimate the support moment for a given OSE and support level. Furthermore, to choose the appropriate model, the previous exoskeleton angle has to be considered, as the change in this angle between the previous and present arm posture is required to identify the movement phase. However, it was predicted that the changes in the exoskeleton angle could be relatively small (below 1°), particularly during static or quasi-static tasks, as minor arm movements can be expected. Even minor changes to the exoskeleton angle would make it difficult to estimate the support moment, as the slightest arm movements would cause the algorithm to detect variations in movement phases and consequently alternate permanently between the two polynomial equations. Therefore, it was decided to create the regression models only for the upward phase of each OSE. Consequently, the support moment provided by an OSE would not be overestimated, as only the minimum possible support moment for a given exoskeleton angle was assumed. The models were fitted separately for each OSE across all support levels.

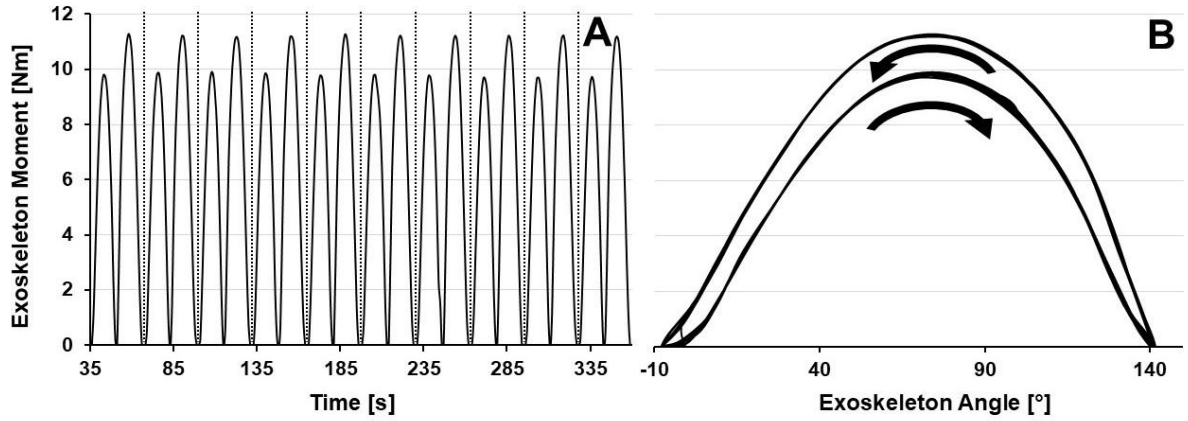


Figure 16: Exoskeleton support moment, estimated from the highest support level for the exoskeleton Paexo, relative to the measurement time (A) and the exoskeleton angle (B) during ten dynamic elevations. Dotted lines (A) indicate the start and end time of each elevation. Arrows (B) indicate the upward (from left to right) and downward phase (from right to left) during the elevations.

For the support force and the external lever arm, the norm of the respective 3D-vector was calculated for every time point to obtain the norm support force and norm external lever arm over time. Then, for the upward phase, the resulting support moment, its corresponding exoskeleton angle, the norm support force and the norm external lever arm were further time-normalized to 100% ($n = 100$ points) for each repetition and the mean and standard error of the mean (SEM) over all ten repetitions was calculated. Additionally, two thresholds T_{low} and T_{high} (relative to T_{min} and T_{max} , respectively) were defined for each exoskeleton angle to limit the range of data points included in the model. It was observed that values outside of these thresholds reduced model performance in predicting the support moment, particularly in the range between 60° and 100° . Therefore, 0° was selected as T_{low} for all OSEs, while T_{high} was selected at 128° for the Airframe, 138° for the Paexo and 143° for the Skelex. These T_{high} values were also determined in accordance with the laboratory and field studies subsequently presented in this thesis, since the maximum shoulder elevation angles observed in all studies were lower than T_{high} . However, it should be noted that the exoskeleton angle measured in this study does not represent the full range of motion of the OSE, but only the range in which OSE was detected. The polynomial regression models were then evaluated for orders from 1 to 12 for each support level separately, using the highest adjusted R^2 value as an indicator of the best model performance.

To estimate the precision of the calculation of the mean support moment and the mean exoskeleton angle for the upward phase, the respective maximum SEM (MaxSEM) of the time-normalized values was obtained. In addition, the maximal support force and mean external lever arm were calculated from the norm support force and the norm external lever arm, respectively. Finally, the peak support moment and its corresponding exoskeleton angle were calculated for each support level of each OSE by performing a one-dimensional optimization of the polynomial regression equations from the lowest to the highest mean exoskeleton angle.

8.4. Results

For the exoskeleton Paexo in the configuration with one expander, the support moment was obtained as a function of the exoskeleton angle for five different support levels (PSL11-15; Figure 17 A). The model equations for these five support settings were estimated from the experimental data using polynomial regression (Figure 17 B).

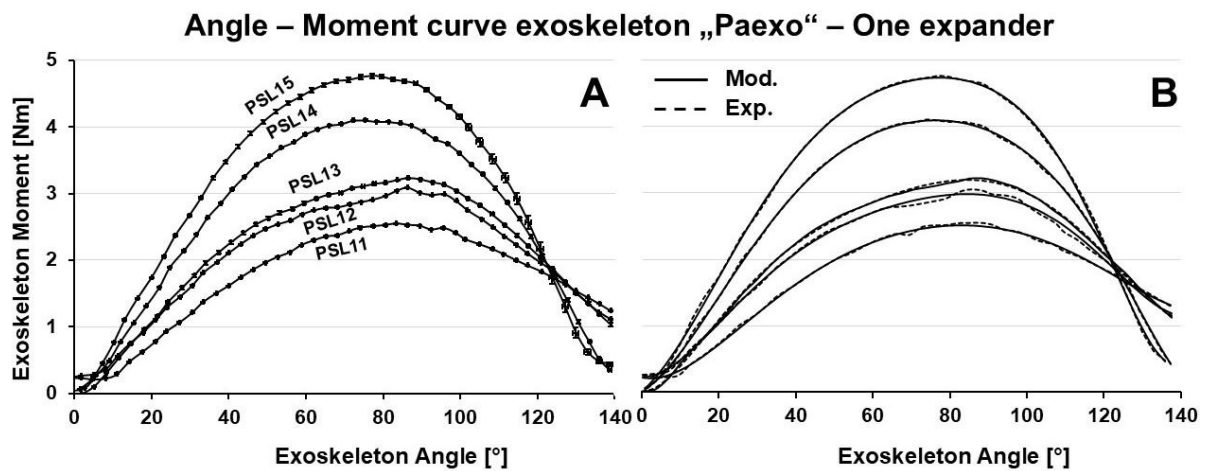


Figure 17: Mean support moment for the five support levels (PSL11-PSL15) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B).

The maximum standard error of the mean (MaxSEM) of the experimental data for the exoskeleton Paexo in the configuration with one expander was below 0.1 Nm for the support moment and ranged from 0.1° to 0.9° for the exoskeleton angle. The highest MaxSEM values for both parameters were observed in the support setting PSL15 (0.1 Nm and 0.9°, respectively; Table 1). When comparing the results of the polynomial regressions, the order with the highest adjusted R^2 values for the different support levels was 6 for the levels PSL11 and PSL12 and 7 for the levels PSL13, PSL14 and PSL15, with an adjusted R^2 value > 0.9 for all five support settings (Table 1). For the estimated peak support moment and maximum support force (3D), an increase was found from the lowest support level PSL11 (2.5 Nm and 10.3 N, respectively) to the highest PSL15 (4.7 Nm and 19.6 N, respectively), while the angle of the peak moment decreased from PSL11 (83.1°) to PSL15 (77.3°, Table 1). The mean external lever arm varied between the support levels from 268.7 mm to 273.6 mm (Table 1).

Table 1: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the five support levels (PSL11-15).

Exoskeleton Paexo – One expander					
	PSL11	PSL12	PSL13	PSL14	PSL15
Order of polynomial	6	6	7	7	7
R^2 -adjusted	0.9	0.9	0.9	0.9	0.9
Estimated Peak Moment [Nm]	2.5	3.0	3.2	4.1	4.7
Angle of Peak Moment [°]	83.1	84.4	84.6	76.3	77.3
Maximum SEM Moment [Nm]	<0.1	<0.1	<0.1	<0.1	0.1
Maximum SEM Angle [°]	0.1	0.1	0.3	0.3	0.9
Maximum Support Force 3D [N]	10.3	12.6	13.6	16.7	19.6
Mean external lever arm 3D [mm]	273.6	273.2	273.4	272.5	268.7

For the exoskeleton Paexo in the configuration with two expanders, the support moment was obtained as a function of the exoskeleton angle for five different support levels (PSL21-25; Figure 18 A). The model equations for these five support settings were estimated from the experimental data using polynomial regression (Figure 18 B).

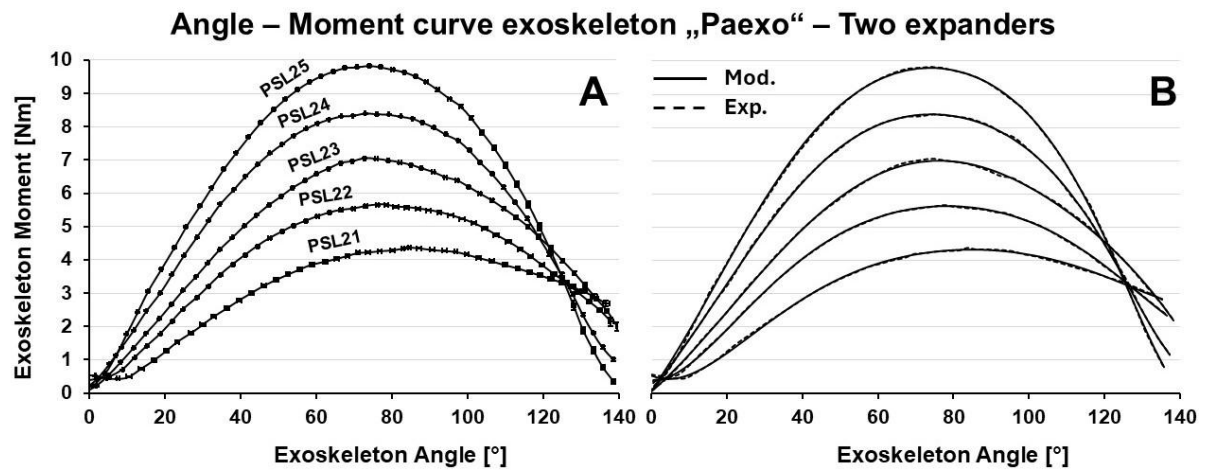


Figure 18: Mean support moment for the five support levels (PSL21-PSL25) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B).

The maximum standard error of the mean (MaxSEM) of the experimental data for the exoskeleton Paexo in the configuration with two expanders ranged from 0.1 Nm to 0.2 Nm for the support moment and from 0.3° to 0.9° for the exoskeleton angle. The highest MaxSEM values for both parameters was observed in support setting PSL21 (0.2 Nm and 0.9°, respectively; Table 2). When comparing the results from the polynomial regressions, the order with the highest adjusted R^2 values for the different support levels was 6 for the levels PSL21, PSL22 and PSL23 and 7 for the levels PSL24 and PSL25, with an adjusted R^2 value >0.9 for all five support settings (Table 2). For the estimated peak support moment and maximum support force (3D), an increase was found from the lowest support level PSL21 (4.3 Nm and 18.1 N, respectively) to the highest PSL25 (9.8 Nm and 39.7 N, respectively), while the angle of the peak moment decreased from PSL21 (84.7°) to PSL25 (74.0°, Table 2). The mean external lever arm varied between the support levels from 270.1 mm to 272.7 mm (Table 2).

Table 2: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the five support levels (PSL21-25).

Exoskeleton Paexo – Two expander					
	PSL21	PSL22	PSL23	PSL24	PSL25
Order of polynomial	6	6	6	7	7
R^2 -adjusted	0.9	0.9	0.9	0.9	0.9
Estimated Peak Moment [Nm]	4.3	5.6	7.0	8.4	9.8
Angle of Peak Moment [°]	84.7	78.3	76.3	74.2	74.0
Maximum SEM Moment [Nm]	0.2	0.2	0.1	0.1	0.1
Maximum SEM Angle [°]	0.9	0.8	0.5	0.3	0.5
Maximum Support Force 3D [N]	18.1	23.0	28.4	33.7	39.7
Mean external lever arm 3D [mm]	271.7	272.3	272.7	272.4	270.1

For the exoskeleton Skelex, the support moment was obtained as a function of the exoskeleton angle for seven different support levels (SxSL1-4; Figure 19 A). The model equations for these seven support settings were estimated from the experimental data using polynomial regression (Figure 19 B).

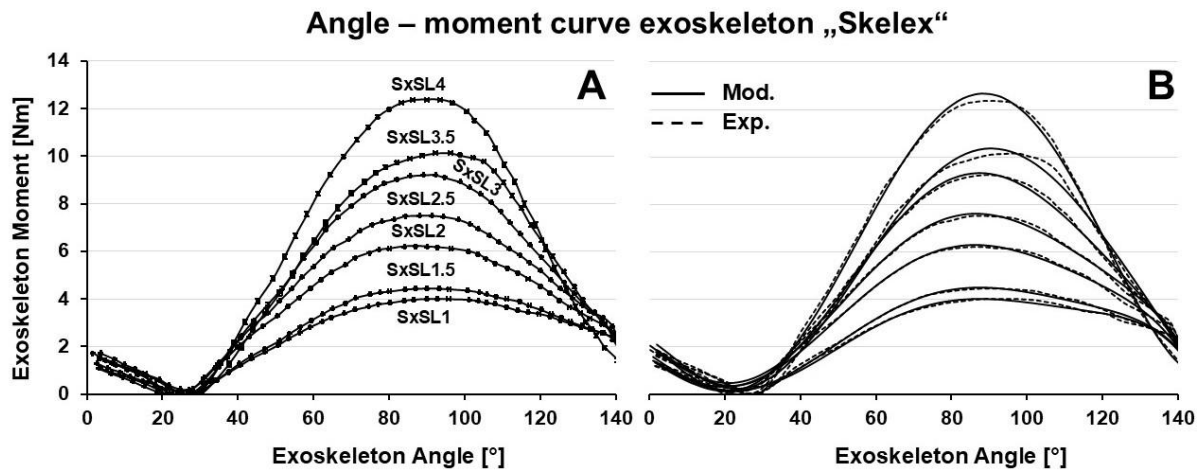


Figure 19: Mean support moment for the seven support levels (SxSL1-SL4) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B).

The maximum standard error of the mean (MaxSEM) of the experimental data for the exoskeleton Skelex was below 0.1 Nm for the support moment and ranged from 0.1° to 0.7° for the exoskeleton angle. The highest MaxSEM values for both parameters were observed in support setting SxSL3.5 (0.1 Nm and 0.7°, respectively; Table 3). When comparing the results of the polynomial regressions, the order with the highest adjusted R^2 values for the different support levels was 6 for all seven support levels with an adjusted R^2 value >0.9 (Table 3). For the estimated peak support moment and maximum support force (3D), an increase was found from the lowest support level SxSL1 (4.0 Nm and 25.1 N, respectively) to the highest SxSL4 (12.7 Nm and 77.6 N, respectively), while the angle of the peak moment varied between 87° and 91° (Table 3). The mean external lever arm varied between the support levels from 202.2 mm to 203.9 mm (Table 3).

Table 3: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R²-value for the seven support levels (SxSL1-4).

Exoskeleton Skelex							
	SxSL1	SxSL1.5	SxSL2	SxSL2.5	SxSL3	SxSL3.5	SxSL4
Order of polynomial	6	6	6	6	6	6	6
R ² -adjusted	0.9	0.9	0.9	0.9	0.9	0.9	0.9
Estimated Peak Moment [Nm]	4.0	4.5	6.3	7.6	9.3	10.4	12.7
Angle of Peak Moment [°]	87.4	87.4	86.3	86.5	87.5	90.5	88.5
Maximum SEM Moment [Nm]	0.1	0.1	0.1	<0.1	<0.1	0.1	0.1
Maximum SEM Angle [°]	0.5	0.5	0.5	0.1	0.1	0.7	0.5
Maximum Support Force 3D [N]	25.1	28.0	38.5	46.4	57.3	64.8	77.6
Mean external lever arm 3D [mm]	203.9	203.7	203.3	203.4	202.6	202.7	202.2

For the exoskeleton Airframe, the support moment was obtained as a function of the exoskeleton angle for two support levels (ASL3-4; Figure 20 A). The model equations for these two support settings were estimated from the experimental data using polynomial regression (Figure 20 B).

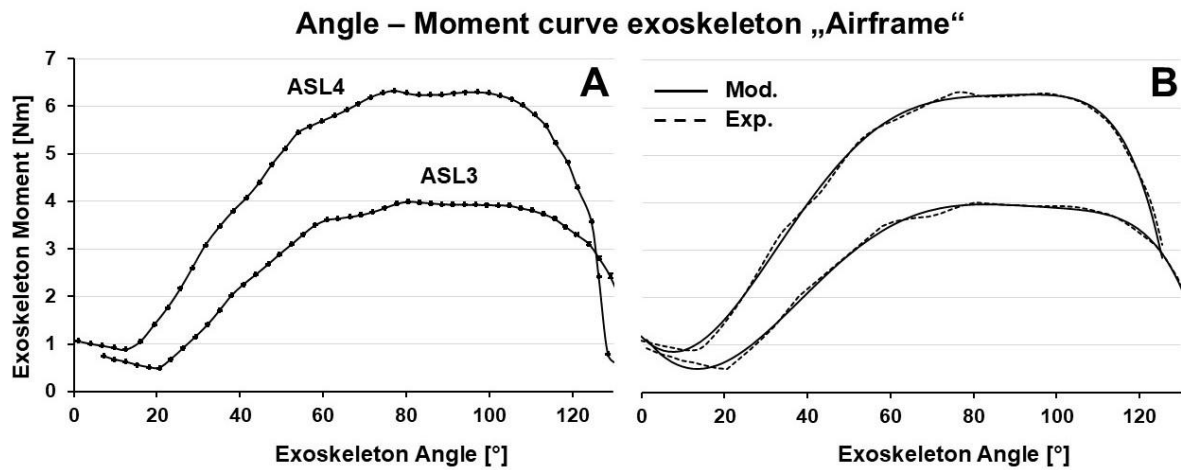


Figure 20: Mean support moment for the two support levels (ASL3 and ASL4) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B).

The maximum standard error of the mean (MaxSEM) of the experimental data for the exoskeleton Airframe was 0.1 Nm and below 0.1 Nm for the support moment and 0.3° and 0.1° for the exoskeleton angle for support levels ASL3 and ASL4, respectively (Table 4). When comparing the results of the polynomial regressions, the order with the highest adjusted R^2 values for the two support levels was 6 for both levels with an adjusted R^2 value >0.9 (Table 4). For the estimated peak support moment and maximum support force (3D), an increase was found from ASL3 (4.0 Nm and 25.3 N, respectively) to ASL4 (6.3 Nm and 32.9 N, respectively) with the angle of the corresponding peak moments at 82.8° and 94.0°, respectively (Table 4). The mean external lever arm was 218.5 mm for ASL3 and 214.7 mm for ASL4 (Table 4).

Table 4: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the two support levels (ASL3 and ASL4).

Exoskeleton “Airframe”		
	ASL3	ASL4
Order of polynomial	6	6
R^2 -adjusted	0.9	0.9
Estimated Peak Moment [Nm]	4.0	6.3
Angle of Peak Moment [°]	82.8	94.0
Maximum SEM Moment [Nm]	0.1	<0.1
Maximum SEM Angle [°]	0.3	0.1
Maximum Support Force 3D [N]	25.3	32.9
Mean external lever arm 3D [mm]	218.5	214.7

8.5. Discussion and limitations

The aim of this study was to quantify the support moment generated by three different occupational shoulder exoskeletons (OSE) across their respective full range of support for all available support levels (SL) during repetitive dynamic elevations performed by a collaborative robot (COBOT) under a standardized measurement protocol. Therefore, the support force at the arm cuffs and the kinematics of the OSE and the COBOT were recorded using a force sensor (attached to the COBOT) and a 3D marker-based motion capture system, respectively. The collected data were then used to calculate the support moment as a function of the exoskeleton angle using polynomial regression enabling further estimation of the support moment provided by the OSE to the user in laboratory settings and under real working conditions.

For all three OSEs, a visual increase in peak support moment was observed with increasing support level (Figure 17 to Figure 20), which is consistent with the findings reported by Hartmann et al. (2021), Musso et al. (2022) and Watterworth et al. (2023). Furthermore, all three OSEs visually demonstrated a clear support effect (greater than 1 Nm) within an elevation range of approximately 40° to 120°, extending up to 140° for certain support level configurations (Figure 17 to Figure 20). The increase in the support moment can be attributed to a rise in the maximum (3D) support force acting on the arm (similar to observations by Musso et al., 2022) from 10.3 N to 39.7 N for the Paexo, from 25.1 N to 77.6 N for the Skelex and from 25.3 N to 32.9 N for the Airframe. Similarly, only small differences were found between the support levels of an OSE when comparing the mean external lever arms, i.e. the distance between the exoskeleton joint center and the point of application of the support force in the arm cuff. In particular, a maximum difference of 4.9 mm was calculated in the mean external lever arm between the different support levels for the Paexo (average mean external lever arm over all support levels = 272.1 mm, 3D), 0.8 mm for the Skelex (average mean external lever arm over all support levels = 201.1 mm, 3D) and 3.9 mm for the Airframe (average mean external lever arm over both support levels = 216.6 mm, 3D), which represents for all OSEs a difference of less than 2% compared to the respective average mean external lever arm. Therefore, these results support the conclusion that the increase in peak support moment is primarily attributable to an increase in the support forces acting on the COBOT arm from the arm cuff, rather than a change in the external lever arm (Musso et al., 2022).

To increase the support level, i.e. the support moment, for the Skelex and the Paexo, the distance between the exoskeleton joint center and the point of application of the support force was increased (Musso et al., 2022, Watterworth et al., 2023; Figure 14), resulting in a larger internal lever arm (Figure 14). This point was identified as where the retraction force resulting from the deformation of the passive structures (the leaf spring for the Skelex and the expanders for the Paexo) was initiated. In addition, for the Paexo, a second expander can be connected, which further increases the support moment, as demonstrated by the results of this study and by Watterworth et al. (2023). In contrast, for the Airframe, the lower part of the exoskeleton, i.e. the cassette, was changed from level 3 to 4 in order to change the support level. Whereas for the Skelex and the Paexo the mechanism for transferring the force generated by the deformation of the passive structures to the arm cuff and the arm is visible, for the Airframe, this mechanism and the passive energy storage structures are concealed inside the cassette. As the presented method did not require further disassembly of the cassette to measure the support moment, it was unclear prior to the measurement how the support moment was increased for this OSE. The increase could be achieved either by adjusting the internal lever arm or by varying the material properties of the passive, energy storing structures. In addition, it was intended to further analyze the effect of the Airframe on the musculoskeletal shoulder load in further studies. Therefore, the cassettes were not disassembled after the measurements, as it was unclear whether a disassembly could affect the internal mechanism that generates the support moment. Furthermore, the author contacted the manufacturer but received no additional information on the passive structures.

The maximum standard error of the mean (MaxSEM) for the support moment and exoskeleton angle (obtained as an indicator of the measurement error of these parameters) ranged from <0.1 Nm to 0.2 Nm and from 0.1° to 0.9° , respectively, across all support levels of the three OSEs. Therefore, the precision in estimating the mean exoskeleton angle from the measurements was considered sufficient, as the maximum SEM values indicate an estimation error of less than 1° for all data points. For the support moment, the maximum SEM was less than 0.1 Nm for both support levels of the Airframe, for nine (out of ten) of the support levels for the Paexo and for two support levels for the Skelex. For the remaining support levels, the maximum SEM was between 0.1 Nm to 0.2 Nm.

However, for all the support levels of the three OSEs, these values are less than 4.2 % of their respective peak support moment (estimated from the polynomial equations). Therefore, it was decided that the errors in the calculation of the mean support moment over the ten repetitions were acceptable and consequently, it was concluded that the measurement setup was sufficiently accurate to estimate the support moment and the exoskeleton angle.

The peak support moments (and its respective angles) estimated by the polynomial equations ranged from 2.5 Nm to 9.8 Nm for the Paexo (74° to 85°), from 4.0 Nm to 12.7 Nm for the Skelex (86° to 91°), and from 4.0 Nm to 6.3 Nm for the Airframe (82.8° for ASL3 and 94.0° for ASL4). For an earlier version of the Skelex, de Vries et al. (2019) reported a peak support moment of 6.1 Nm (intermediate level of support, de Vries et al., 2019) at 90° shoulder flexion by measuring the vertical force at the arm cuff with an integrated force sensor under the cuff while participants performed static arm elevations at this angle. In addition, Musso et al. (2022) calculated the maximum support force at the arm cuff for the same (newer) version of the Skelex between 85°-90° shoulder flexion. Therefore, the estimated exoskeleton angles of the peak support moment for the Skelex are consistent with the findings of both de Vries et al. (2019) and Musso et al. (2022). Furthermore, this study found a peak support moment of 6.3 Nm for the newer version of the Skelex in the support level 2 (SxSL2, range of SxSL1 – SxSL4), which is similar to the peak moment reported by de Vries et al. (2019, small variations can be attributed to the differences in the exoskeleton version and the measurement method). In contrast, Watterworth et al. (2023) measured higher peak support moments for the Paexo at all support levels in both expander configurations with a mean difference of 1.2 Nm. To compare the values between both studies, the peak support moments and angles from Watterworth et al. (2023) were estimated in RStudio based on the equations and the optimization algorithm published in their study. These differences in peak support moment between this study and Watterworth et al. (2023) may be, to a lesser extent, due to the difference in measurement method (arm cuff support force measurement versus dynamometer support moment measurement). Watterworth et al. (2023) also applied a different approach for estimating the order of polynomial regression models, choosing a model that better balanced goodness of fit and complexity by using a lower 4th-order polynomial for the Paexo (Watterworth et al., 2023).

In contrast, the approach in this study was to identify the perfect fit of the measured support moments and angles to the estimated model data in order to reduce the error in calculating the support moment of these specific exoskeletons provided to the user in laboratory settings and under real working conditions. Finally, although the date the Paexo was purchased by Watterworth et al. (2023) is unclear, the Paexo tested in this study was purchased in 2019 and has already been used in previous studies and field trials. Therefore, the advanced age of the OSE in combination with the extended use, resulting in repeated mechanical loading by expander deformation, may have led to an advanced material degradation, potentially contributing to a continuous decrease in the retraction force of the expanders. Environmental factors, e.g. temperature, exposure to light and air humidity, may also have negatively affected this degradation. To support those assumptions, the author contacted the manufacturer who confirmed that primarily the advanced age of the purchased Paexo, and consequently, the integrated expanders, could be a valid explanation for the differences in the estimated peak support moments between this study and Watterworth et al. (2023).

Watterworth et al. (2023) further reported generally higher estimated peak support moment angular positions (from 91.0° to 107.7°) compared to this study (mean difference between studies of 19.5°). Similar to the conclusion for the peak support moments, variations were expected due to the different approaches used to estimate the model equations. However, differences in the measurement setup for the Paexo could also have contributed to the variations in the estimated angles for the peak support moments. In this study, the protective cover around the expanders of the Paexo was removed and the exoskeleton angle was defined as the angle between the upper part (where the lower part of the expanders were attached; shoulder element) and the lower part (where the upper part of the expanders were attached; arm element; Figure 21 A & B). This step was considered necessary to better secure the exoskeleton to the measurement setup and it was assessed that removing the protective cover would not affect the retraction force of the expanders. In contrast, Watterworth et al. (2023) fixed the Paexo to their measurement setup with the flat side of the protective cover facing downwards (as inferred from the figures of Watterworth et al., 2023; Figure 21 C). Between the upper part of the Paexo (defined in this study) and the flat side of the protective cover, an angle of approximately $6-7^{\circ}$ was measured by a goniometer.

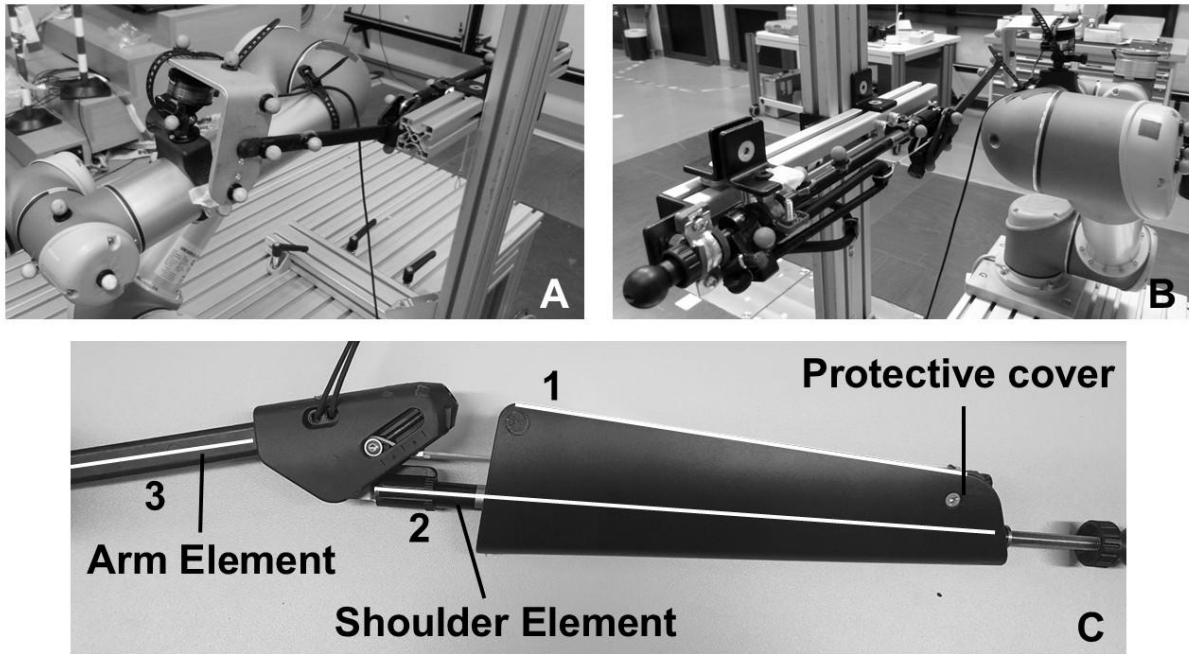


Figure 21: Measurement setup for the Paexo Shoulder in front (A) and back view (B). To attach the Paexo to the measurement table, the protective cover around the expanders (C) was removed. The exoskeleton angle was defined between the arm element (line 3 in C) and the shoulder element (line 2 in C) in this study. For Watterworth et al. (2023), it was assumed that the exoskeleton angle was defined between the flat side of the cover (line 1 in C) in relation to the arm element (line 3 in C).

Therefore, it is reasonable to assume that these differences in exoskeletons fixation may have additionally contributed to the variation in the estimated peak support moment angles. However, this assumption cannot be proven because only a general description of the exoskeleton angle across all of the four exoskeletons tested by Watterworth et al. (2023) is available. Ultimately, this study and Watterworth et al. (2023) both found a reduction in peak moment angle of 8.3° and 10.7° on average from the lowest to the highest support level in both expander configurations. These results indicate that increasing the internal lever arm for the Paexo does not only increase the peak support moment, but also shifts it to a lower angle position, which could be a relevant piece of information when implementing this OSE at the workplace.

Finally, for the Airframe, the results for the peak support moment and angle for the support level 4 were consistent when compared to Watterworth et al. (2023) who reported a peak support moment of 6.5 Nm at 97.2° (estimated from the equations for the configuration “Range 90” in Watterworth et al., 2023). In contrast, for the support level 3, Watterworth et al. (2023) reported a higher peak support moment of 5.6 Nm (4.0 Nm in this study) at a higher exoskeleton angle of 97.3° (82.8° in this study). Similarly, for the Paexo, the variation in peak support moment could be due to differences in the measurement setup, the approach for estimating the respective polynomial regression equations and/or a decrease in retraction force due to previous extended use. However, this last suggestion is only hypothetical since, as mentioned above, the material of the passive, energy storing structure and its material properties are not known. The use of different materials for the passive, elastic structures of an OSE, e.g. rubber or carbon, with different elastic properties will result in a different behavior over time or when exposed to environmental conditions (e.g. heat, sunlight). This assumption can be further supported by comparing the passive structures of the Skelex and the Paexo. While the passive structures for the Skelex are made of carbon fibers, the expanders built into the Paexo are likely based on polymer material, such as rubber, and could be more prone to a decrease in the retraction force over time (i.e. advanced age), as confirmed by the manufacturer. However, no information of the passive structures inside the Airframe could be provided even after contacting the manufacturer.

Finally, since all values of the adjusted R^2 were greater than 0.9, it can be concluded that the polynomial equations based on the measured data are sufficiently accurate to estimate the support moment from the exoskeleton angle for a given level of support. Therefore, these equations can be used in future studies to calculate the support moment of the OSE and to analyze its effect on the shoulder joint moment.

Several limitations of this study must be considered when interpreting the results. First, as mentioned above, all three OSEs had been used and analyzed in previous laboratory studies and field tests. Therefore, the duration of previous use, advanced age and the exposure to different environmental conditions (e.g. heat, sunlight) could not be assessed, particularly during the field tests. Consequently, it remains unclear whether these factors could have negatively affected the retraction force of the passive, elastic structures, a concern associated with the Paexo. Newer versions of these OSE, manufactured more recently, may provide a higher support moment to the user.

Second, although the approach of estimating the polynomial equations for each support level in each configuration of the OSE was considered sufficient to calculate the support moment in the studies subsequently presented in this thesis, these equations are limited to the specific versions of the OSE tested in this study. Therefore, a direct transfer of these equations to other versions of the OSEs has to be applied with caution, especially for newer models (already being on the market for the Airframe and the Paexo). If applicable, the author suggests that own measurements should be conducted to quantify the support moment when analyzing newer versions of these OSEs (either recently manufactured or newer models). In addition, it has to be noted that the polynomial regression models were only estimated for the upward phase in order to quantify the support moment for a given exoskeleton angle when performing static and quasi-static task. During tasks requiring dynamic movements, e.g. cleaning a ceiling, plastering or manual material handling between varying heights, a higher support moment may be provided from the OSEs due to the hysteresis effect (Johns et al., 2024). However, if own measurements are not applicable, the equations presented in this study provide a good first overview of how the support moments of the different OSEs behave across the range of different elevation angles, even to dynamic movements as only the minimum support moment will be applied.

Third, the exoskeleton angle was calculated relative to a reference angle of 90° between the lower and upper part of the exoskeleton. Therefore, the lower and upper parts and the connecting rotational joint of each exoskeleton were individually identified based on a visual inspection of the study personnel. Consequently, a different definition of the exoskeleton angle or the reference position could shift the moment angle curve along the axis of the exoskeleton angle, resulting in different angles for the peak support moment (as assumed for the results for the Paexo). To the best of the author's knowledge, there is currently no standardized definition of the exoskeleton angle, which limits the comparability between studies, even when the same exoskeleton is being tested. Finally, the arm cuffs for the Paexo and Skelex had to be replaced with a self-produced metal bracket since (a) the arm cuff for the Skelex only functions when the arm straps are applied (exerting pressure on the force sensor causing inaccurate detection of the support force), and (b) the bottom of the Paexo arm cuff is uneven making it impossible to reliably estimate the point of application of the support force.

As a result, the forces measured by the sensor on the arm cuff may differ slightly from the forces applied to the user's arms, particularly as the arm cuffs for all three OSEs are normally fitted with a small layer of padding inside the arm cuff to improve the comfort for the user. Therefore, the padding material is expected to have a small dampening effect on the support forces.

8.6. Conclusion

In summary, the support moments of the three occupational shoulder exoskeletons tested were measured for all available support levels over the entire support range of exoskeletons. The resulting data obtained were used to estimate polynomial equations to describe the support moments as a function of the exoskeleton angle, as well as to determine the maximum peak moment and its corresponding angle. These results for peak support moments and angles were generally consistent with findings in the literature (de Vries et al., 2019; Hartmann et al., 2021; Musso et al., 2022; Pinho & Forner-Cordero, 2022; Watterworth et al., 2023) with slight variations attributed to differences in the study design and exoskeleton age. These equations can be applied in future studies to calculate the support moment provided by the exoskeleton under various conditions. By integrating the support moment into a biomechanical model, the effect of the OSE on the shoulder joint moment can be estimated. In addition, if shoulder muscle activity or muscle tissue oxygenation is measured simultaneously, additional insights conclusions can be gained about the overall influence of the OSE on the shoulder muscle forces and further musculoskeletal shoulder load.

9. Influence of a shoulder exoskeleton on mechanical load and muscle demand during static arm elevation

The aim of this study was to conduct an exploratory analysis of the influence of an occupational shoulder exoskeleton (OSE) on mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand during static arm elevation. Therefore, the relative net shoulder joint moment for the right shoulder ($RNSM_{rel}$; variable: $mRNSM_{rel}$) and the muscle activity in the right musculus deltoideus pars clavicularis (right anterior deltoideus, RAD; variable: $mRAD$) and musculus deltoideus pars acromialis (right medial deltoideus, RMD; variable: $mRMD$) were measured. They were selected to serve as indicators for the changes in the mechanical load at the GHJ and shoulder muscle demand. The OSE used in this study was the Levitate Airframe.

9.1. Participants

Ten young, healthy participants (five males, five females; mean $\pm$ standard deviation; 25.1 years $\pm$ 3.3, 1.8 m $\pm$ 0.6, 68.6 kg $\pm$ 7.4) voluntarily participated in this study which was approved by the Ethics Committee of the German Sport University Cologne. Before the measurements were taken, participants gave written informed consent and completed a health questionnaire. All participants reported no history of musculoskeletal diseases prior to the measurement.

9.2. Experimental design and task description

A three-factorial repeated-measures design was applied with the independent variables “exoskeleton (EXO_A)”, “shoulder elevation angle (SEA)” and “shoulder abduction angle (SAA)”. Participants performed bilateral static repetitive arm elevations in an upright position (both knees fully extended) with (EXO_A) and without (NEXO) the OSE for six combinations of SEA and SAA: shoulder elevation of 60°, 90° and 120° and shoulder abduction of 0° and 30° (Figure 22). During the arm elevation, the elbow was fully extended, and the arm was internally rotated with the palms facing in medial direction (Figure 22). In addition, two dumbbells (1.5 kg) were placed in the participant’s hands as an external load to simulate a work tool, such as a drill.

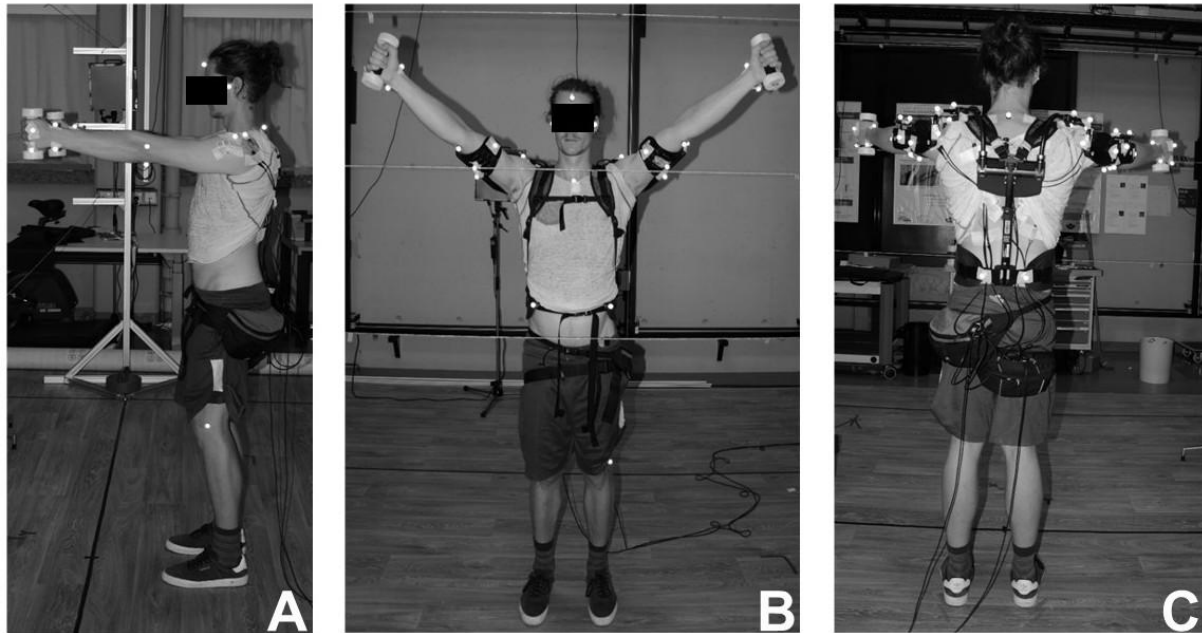


Figure 22: Participant performing arm elevations with (B,C) and without (A) an occupational shoulder exoskeleton in three arm postures (values for SEA and SAA specify the respective shoulder elevation and abduction angle): (A) SEA = 90°, SAA = 0°, (B) SEA= 120°, SAA = 30°; and (C) SEA = 90°, SAA = 30°.

9.3. Measurement instrumentation and data acquisition

First, segment lengths and circumferences of the participant's right arm (upper arm, forearm and fist) and body weight were obtained following the instructions by Hanavan (1964), to calculate inertial properties. The OSE was then adjusted according to the participant's anthropometry and all participants reported a perfect fit. The OSE used in this study was the Airframe with support level 4. After the OSE was adjusted, adhesive electrodes (Ag/Ag-Cl, BlueSensor N, Ambu A/S, Ballerup Sogn, Denmark) were placed on the RAD (Figure 23 C) and the RMD (Figure 23 D) in order to measure electromyographic (EMG) signals from the muscles during the arm elevation. The electrode positions were determined according to the SENIAM guidelines to ensure a stable and standardized measurement of the EMG signal (Hermens et al., 2000). Before applying the electrodes, the skin over the muscles of interest was shaved, pretreated with abrasives and cleaned with alcohol. The EMG signals during the trials were recorded with the BioMed cable-based four-channel system (1000 Hz; BioMed Jena GmbH Biomedizinische Technik, Jena, Germany) and time-synchronized with the ViconNexus software (Version 11, Vicon Motion Systems, Inc., Oxford, UK).

To record participant's kinematics during the trials with 12 Vicon cameras (100 Hz; Vero, Vicon Motion Systems, Inc., Oxford, UK), 16 retro-reflective markers (diameter 14 mm) were placed bilaterally above the acromion, laterally on the upper arm (approximately at the level of the GHJ), on the lateral and medial epicondyles of the elbow, on the radial and ulnar styloids of the wrist, and over the base joint of the middle finger (articulationes metacarpophalangeae III). An additional marker was placed on the lateral epicondyle of the left knee (Figure 23 A & B).

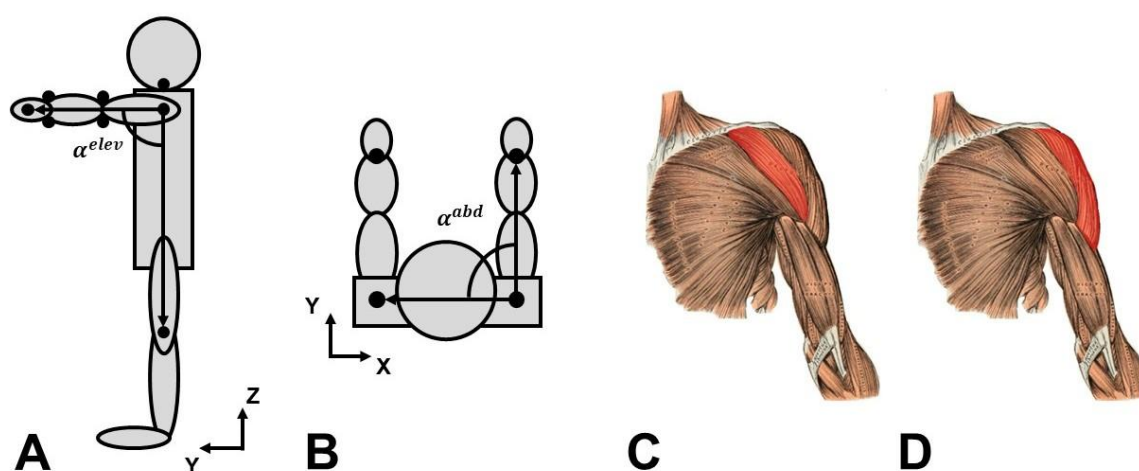


Figure 23: Illustration of the calculation of the shoulder elevation (α^{elev} , A) and shoulder abduction angle (α^{abd} , B) with the respective retro-reflective markers placed on the participants (black circles) and the muscle analyzed in this study: the musculus anterior deltoideus (C, red) and the musculus medial deltoideus (D, red). Graphics C and D were extracted and modified from Gray and Lewis (1918; p. 436).

Participants were then positioned in the measurement area and instructed to raise their arms in the sagittal plane while maintaining an upright posture with fully extended arms and knees until they perceived reaching a shoulder elevation angle of 60°. During this process, the shoulder elevation angle was calculated in real time by the Vicon Nexus software as the angle between the markers placed above the middle finger, lateral to the shoulder and lateral to the knee. Participants received verbal feedback on their shoulder elevation angle and were corrected until the desired angle was achieved. The vertical displacement of the marker above the middle finger was then recorded, as it represents the height to which the participant's hand must be raised to achieve a shoulder elevation angle of 60°. This procedure was repeated for the remaining two shoulder elevation angles.

Prior to a measurement session, two bars were placed on opposite sides of the measurement area and connected vertically by three ropes at different heights (Figure 22). After the initial measurements, the height of each rope was adjusted according to the values obtained for the respective shoulder elevation angles. This ensured that the participant was able to reproduce the required shoulder elevation in each trial by raising their hands until the middle finger reached the height of the respective rope. In addition, the position of the feet was marked on the floor to ensure a constant distance from the ropes. The procedure was then repeated to obtain the two shoulder abduction angles. Therefore, participants were positioned in front of the ropes and asked to raise their arms “parallel” until they reached the first rope (height level for SEA60°). The shoulder abduction angle was then calculated in real time as the angle between the two markers placed above the acromion and the marker placed on the radial styloid of the wrist. The participant’s arm position was further corrected until an abduction angle of 0° was achieved, and two marks were placed on the rope as target points for the elevation trials. This procedure was repeated for the remaining two SEA (90° and 120°) and for the second abduction angle (SAA 30°).

Data collection began with participants performing maximum isometric voluntary contractions (MVC) in each of the six arm postures, i.e. each combination of SEA and SAA, to determine maximum reference values for RAD and RMD muscle activity. Therefore, they were instructed to pull a rope upwards with their hands. The rope was connected to a non-movable weight that acted as a fixed connection to the floor. The height of the rope was adjusted according to the hand heights obtained in the previous section in order to achieve the respective shoulder elevation angles (Figure 24 B, left panel). The MVC trials with a shoulder abduction angle of 0° were performed with both hands and the body parallel to the weight whereas the trials with the shoulder abducted in 30° were performed with only one hand and the body rotated at an angle of 30° to the weight. To ensure a 30° abduction angle during the MVC, the participants were asked to stand in front of the weight before the trial, and they had to lift the rope with their arm fully extended until they reached the designated height. They were then instructed to “rotate the body away from the extended arm” until the instructor reported a 30° shoulder abduction (verified in real-time via ViconNexus) and the feet position was marked on the floor. In addition, the weight was placed on a force plate (1000 Hz, Kistler, Type 9286B, Kistler Instrumente GmbH, Sindelfingen, Germany) in order to obtain the ground reaction forces during a MVC trial (Figure 24 B, left panel).

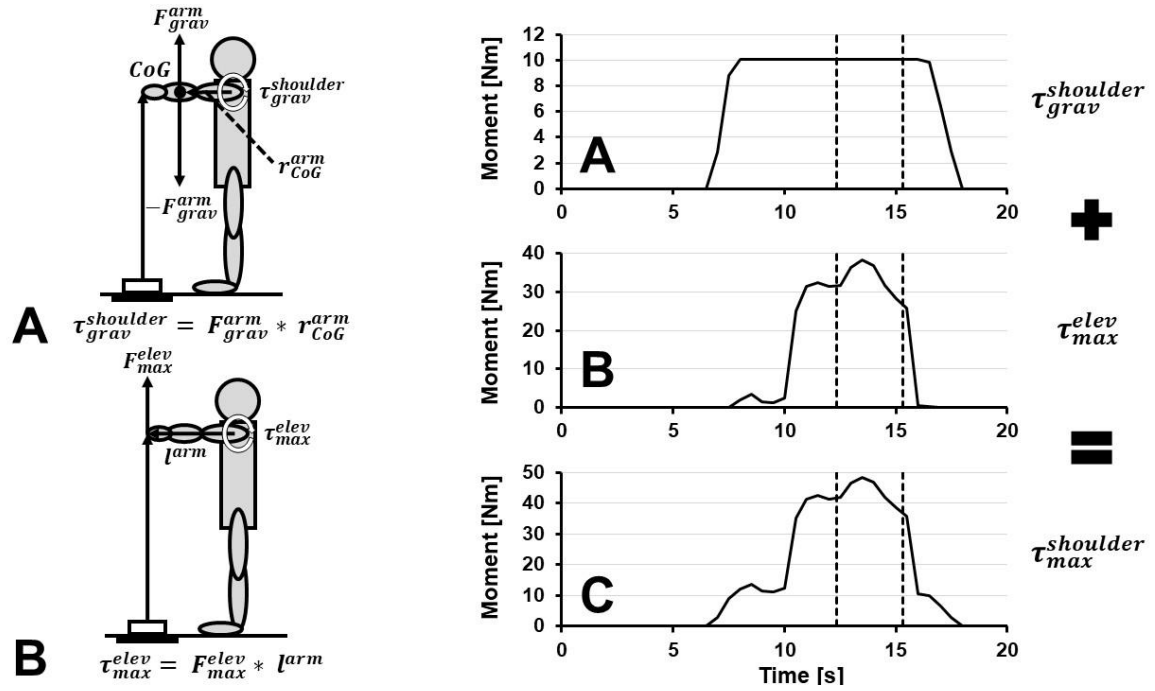


Figure 24: Estimation of the maximum net shoulder joint moment ($\tau_{max}^{shoulder}$, C) for 90° shoulder elevation as the sum of the gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$, A) and the maximum shoulder elevation moment (τ_{max}^{elev} , B), the later obtained by the maximal voluntary contraction trial (MVC). In this posture, $\tau_{grav}^{shoulder}$ was estimated as the product of the gravitational force (F_{grav}^{arm}), acting at the center of gravity (CoG), and the distance between the glenohumeral joint center and the CoG (r_{CoG}^{arm}). τ_{max}^{elev} was estimated as the product of the elevation force (F_{max}^{elev}) and the arm length (l^{arm}).

Prior to an MVC trial, the participants were instructed to find a comfortable, hip-wide stance in front of the force plate, where they could maintain the arm elevation with a straight back and shoulders parallel to the hips (i.e., without shifting the GHJ center in an anterior-posterior direction). Each MVC trial started with a ten-second baseline to collect baseline values from the EMG and force data. The rope was then placed in the participant's hands and the arm was raised until the specific shoulder elevation angle was reached. To prevent injuries, participants were instructed to gradually increase the vertical force to a maximum over two seconds and then maintain this contraction for an additional three seconds. Finally, the force was slowly released and the arm was lowered back to the neutral position. Maximum muscle activity for RAD and RMD were later calculated as the maximum values obtained during the three seconds of maximum contraction.

During the MVC trials, participants received verbal, non-threatening encouragement and were instructed to pull primarily with the shoulder muscles while avoiding “lifting from the back” or “raising the shoulder (GHJ center)”, thereby minimizing the influence of the back or neck muscles. The trials were performed in a randomized order with two minutes rest in between.

It should be noted, that the study protocol was designed to collect bilateral data on muscle activity and shoulder moment. Therefore, the following elevation trials were performed with both arms. In addition, the MVC trials in SAA30° were also performed with the left arm. However, due to errors in the measuring device, which became apparent after completion of the measurements, the data for the left arm could not be further analyzed. This issue was discussed in the research group and no objections were raised regarding its impact on the remaining data for the elevation and MVC trials for the right arm, as the data for both arms were recorded separately. After completion of the MVC trials, participants performed static arm elevations in twelve experimental conditions (all combinations of “EXO_A”, “SEA” and “SAA”). A three-step randomization protocol was used in order to minimize possible sequential effects due to muscle fatigue:

- 1) First, the “EXO_A” factor was counterbalanced by having five participants start with the exoskeleton and five without it. To keep the EMG electrodes and markers in place and minimize the risk of them detaching (particularly when donning and doffing the exoskeleton), each participant completed all combinations of the remaining two factors with the exoskeleton before repeating the procedure without it (and vice versa).
- 2) Second, the SAA factor was counterbalanced for each iteration of NEXO/EXO_A.
- 3) Third, the order of SEA was randomized for each participant. Participants completed all trials for the three SEA for a given SAA before repeating this procedure for the other SAA.

Each measurement block (EXO_A/NEXO) consisted of the following: prior to an elevation trial the dumbbells were placed in the participant’s hands. The participants then began the elevation trial in a T-Pose (to ensure an active EMG signal), followed by ten seconds of rest. After the ten seconds, they performed the elevation in the designated posture (combination of SEA and SAA) for an additional ten seconds. This elevation was repeated five times with a ten second rest between each repetition.

Participants then performed the same procedure for the remaining two elevation angles with a two-minute rest in between. Between each trial, the dumbbells were removed from their hands and participants could rest on a chair, if desired.

9.4. Data processing and dependent variables

Data analysis was performed in Python (Software Spyder, version 3.12.0, Python Software Foundation, Delaware, USA). First, the 3D marker trajectories were filtered (Butterworth, low pass, 5 Hz, 4th order) for all MVC and elevation trials. The shoulder elevation angle (α^{elev}) was then calculated by the dot product ($\cdot$) of the vectors (vec) of the sagittal components (Y-Z-plane) between the three markers placed laterally on the upper arm (UPA, approximately at the level of the GHJ), the lateral epicondyle of the knee (KNE) and over the base joint of the middle finger (FIN, Figure 25).

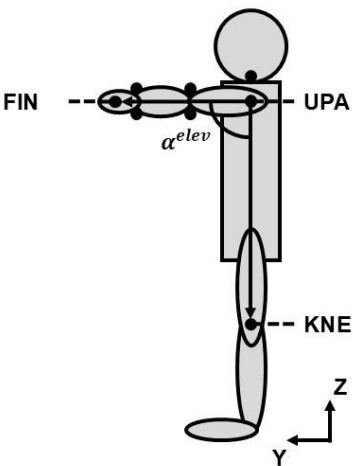
$$\alpha^{elev} = \arccos \left(\frac{vec(FIN - UPA) \cdot vec(KNE - UPA)}{norm(vec(FIN - UPA)) * norm(vec(KNE - UPA))} \right)$$


Figure 25: Estimation of the shoulder elevation angle (α^{elev}) by the dot product ($\cdot$) of the vectors (vec) between the three markers UPA, KNE and FIN, with *norm* indicating the calculation of the norm (i.e. length) of the specific vector and *arccos* indicating the calculation of the angle based on its corresponding arcus cosines value.

The raw EMG signals were amplified (by a factor of 1000) and filtered with a bandpass filter (20-450 Hz) and a notch filter (50 Hz). The start times of each of the five repetitive arm elevations were identified based on the shoulder elevation angle using the “detect_cycles” function of the “Kinetics Toolkit” (Chénier, 2021), and the data were segmented into the individual repetitions. For the EMG data, the following data processing steps were performed separately for each repetition.

The filtered EMG data were offset-corrected by subtracting the mean value calculated over the first second of each repetition, before the arm was raised, ensuring that the participants were in neutral position. The signals were then rectified and the root mean square value (RMS, window length 100 milliseconds) was calculated. The RMS data were offset-corrected a second time by subtracting the minimum value obtained over the first second of a repetition and then normalized with respect to the corresponding maximum reference value for each elevation trial obtained during the MVC trials.

To calculate the gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$), the segment lengths for the right upper and lower arm were determined from the marker data recorded during the first five seconds of an elevation trial, with the participant in a neutral position and their arms at their side. Therefore, the length of each segment was defined as the distance between the global positions of the GHJ, elbow and wrist joint center, respectively. The position of the elbow and wrist joint center were obtained by calculating the midpoint between the markers on the epicondyle and styloid, respectively. To determine the global position of the GHJ, a vertical projection of the marker placed over the right acromion was calculated by subtracting 17% of the distance between the two acromion markers from its vertical position (with the marker diameter subtracted beforehand), as proposed by Rab et al. (2002). Based on their study, Michaud et al. (2016) identified this algorithm as a reliable method to calculate the GHJ position from marker data without using imaging data (e.g. CT or MRI). The inertial properties for the right arm of each participant were calculated based on the measured segment lengths and circumferences as well as the body weight, according to Hanavan (1964). In addition, for the elevation trials, the weight of the dumbbell (1.5 kg) was added to the weight of the hand. Then, the shoulder gravitational joint moment ($\tau_{grav}^{shoulder}$) was calculated as:

$$\tau_{grav}^{shoulder} = F_{grav}^{arm} * r_{CoG}^{arm} \quad (9.1)$$

where is F_{grav}^{arm} the weight force of the arm including the dumbbell and r_{CoG}^{arm} is defined as the product of the sinus of the shoulder elevation angle and the distance between the GHJ and the center of gravity (CoG) of the arm. To calculate the support moment provided by the OSE during arm elevation τ^{exo} , the polynomial equation obtained for the Airframe (support Level 4) in the previous section was applied with the shoulder elevation angle as input variable.

Thus, the absolute net shoulder joint moment for the right shoulder ($RNSM_{abs}$) when using the exoskeleton was defined as the difference of $\tau_{grav}^{shoulder}$ and the support moment provided by the OSE:

$$RNSM_{abs} = \tau_{grav}^{shoulder} - \tau^{exo} \quad (9.2)$$

For all NEXO trials, the net shoulder joint moment ($RNSM_{abs}$) was equal to the gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$).

The force data measured during the MVC trials were also offset-corrected (subtracting the average over the first 5 seconds), filtered (Butterworth, low pass, 5 Hz, 4th order) and the resulting force was calculated as the norm of the force vector. The maximum shoulder elevation moment (τ_{max}^{elev}) was defined as (Figure 24):

$$\tau_{max}^{elev} = F_{max}^{elev} * l^{arm} \quad (9.3)$$

where F_{max}^{elev} is the maximum elevation force and l^{arm} is the product of the sinus of the shoulder elevation angle and the arm length. Therefore, the right maximum shoulder joint moment ($\tau_{max}^{shoulder}$) was estimated as the sum of $\tau_{grav}^{shoulder}$ and τ_{max}^{elev} (Figure 24):

$$\tau_{max}^{shoulder} = \tau_{grav}^{shoulder} + \tau_{max}^{elev} \quad (9.4)$$

The reference value for the respective arm posture ($RNSM_{max}$) was obtained as the mean value over the three-second period of maximum contraction from $\tau_{max}^{shoulder}$.

The relative net shoulder joint moment ($RNSM_{rel}$) in each arm posture was then calculated by further normalizing the right absolute net shoulder joint moment ($RNSM_{abs}$) to the maximal reference value ($RNSM_{max}$) obtained from the respective MVC trial:

$$RNSM_{rel} = \frac{RNSM_{abs}}{RNSM_{max}/100} \quad (9.5)$$

For further data analysis, the static phase in each of the five repetitions during an elevation trial was identified and a five-second window was applied for each repetition in order to obtain values for $RNSM_{rel}$ and the normalized muscle activity for RAD and RMD (Figure 26).

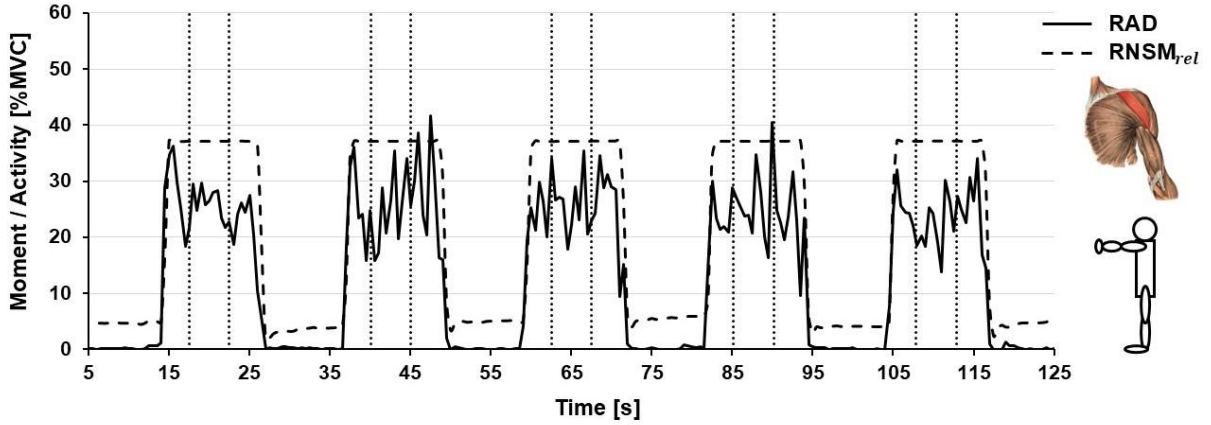


Figure 26: Muscle activity of the right anterior deltoideus (RAD) and right relative net shoulder joint moment ($RNSM_{rel}$) of an exemplary participant for the static arm elevation performed in 90° shoulder flexion with 0° shoulder abduction. Vertical dot lines show the start time and end time for the static phase (five-second window) during an arm elevation. The graphic for the anterior deltoideus (right side) was extracted from Gray and Lewis (1918; p. 436) and modified.

The values obtained from each of the five repetitions were combined and the mean value was calculated over all values (example for RAD):

$$mRAD = \frac{\sum_0^n RAD_0^{rms} + \dots + \sum_0^n RAD_5^{rms}}{n * 5} \quad (9.6)$$

with n being the number of data points in one time window (depending on measurement frequency and data processing steps). This procedure was conducted for $RNSM_{rel}$ ($n = 500$) as well as RAD and RMD ($n = 50$ for both parameters) to obtain mean values for these parameters for further statistical analysis: $mRNSM_{rel}$, $mRAD$ and $mRMD$, respectively.

9.5. Statistics

Statistical analysis was performed using R (version 1.2.5042, RStudio, Inc., Boston, MA). Dependent variables included in the analysis were $mRNSM_{rel}$, $mRAD$ and $mRMD$ as indicators for changes in relative net shoulder joint moment and muscle activity, respectively. A three-way repeated-measures ANOVA was conducted separately for each dependent variable to determine main and interaction effects for all three independent variables: EXO_A , SEA and SAA (2 x 3 x 2 repeated-measures design). Normal distribution of the residuals was visually inspected via Q–Q (quantile-quantile) plots. Boxplots were used to detect outliers and sphericity was analyzed using Mauchly's test with a Greenhouse–Geisser correction applied, if necessary. Statistical significance level was set at 5% and post-hoc pairwise comparisons were carried out using Student's t-test with Bonferroni-Holm correction.

9.6. Results

Throughout the elevation trials, the reduction in the right absolute net shoulder joint moment ($RNSM_{abs}$) between the NEXO to the EXO_A condition ranged from -2.6 to -6.4 Nm with the highest mean reduction reported for $SEA90^\circ$ with -6.4 Nm for $SAA0^\circ$ and -6.2 Nm for $SAA30^\circ$ (Table 5). The lowest reduction for $RNSM_{abs}$ was found for $SEA120$ with a mean reduction of -4.6 Nm for $SAA0^\circ$ and -3.8 Nm for $SAA30^\circ$ (Table 5). For the $SEA60^\circ$ condition, a mean reduction of -5.9 Nm and -5.8 Nm was reported in $mRNSM_{abs}$ for $SAA0^\circ$ and $SAA30^\circ$, respectively (Table 5).

Table 5: Mean value ($\pm$ standard deviation) for the right absolute net shoulder joint moment, obtained from trials with (EXO_A) and without (NEXO) exoskeleton, in all combinations of shoulder elevation (SEA) and shoulder abduction (SAA) angle.

Shoulder angle [°]		Right Absolute Net Shoulder Joint Moment [Nm]		
SEA	SAA	NEXO	EXO_A	$NEXO - EXO_A$
60°	0°	17.3 ($\pm$ 1.6)	11.4 ($\pm$ 1.6)	-5.9 ($\pm$ 0.4)
	30°	16.7 ($\pm$ 1.5)	10.9 ($\pm$ 1.6)	-5.8 ($\pm$ 0.5)
90°	0°	19.4 ($\pm$ 1.9)	13.1 ($\pm$ 1.9)	-6.4 ($\pm$ 0.0)
	30°	19.2 ($\pm$ 2.1)	13.1 ($\pm$ 1.9)	-6.2 ($\pm$ 0.7)
120°	0°	17.2 ($\pm$ 1.9)	12.5 ($\pm$ 1.6)	-4.6 ($\pm$ 0.6)
	30°	16.5 ($\pm$ 1.6)	12.7 ($\pm$ 1.5)	-3.8 ($\pm$ 0.8)

For the dependent variable $mRNSM_{rel}$, repeated measures ANOVA showed a significant two-way interaction effect for $EXO_A \times SEA$, $EXO_A \times SAA$ and $SAA \times SEA$ as well as significant main effects for all three factors (EXO_A , SEA and SAA ; all p – values < 0.05). Further pairwise comparisons revealed a significant reduction in $mRNSM_{rel}$ when wearing the exoskeleton, independent of shoulder elevation or abduction angle, with the highest mean reduction for $SEA90^\circ$ of -13.9 %MVC for $SAA0^\circ$ and -12.0 %MVC for $SAA30^\circ$ (Figure 27 B; all p – values < 0.01). The lowest reduction in $mRNSM_{rel}$ between EXO_A vs NEXO condition was found for $SEA120^\circ$ with a mean reduction of -9.9 %MVC for $SAA0^\circ$ and -9.1 %MVC for $SAA30^\circ$ (Figure 27 C).

For SEA60°, mRNSM_{rel} showed a mean reduction in EXO_A compared to NEXO of -12.0 %MVC and -9.6 %MVC for SAA0° and SAA30°, respectively (Figure 27 A). For the main effect of SEA, pairwise comparisons showed a significant increase in mRNSM_{rel} for SEA90° compared to SEA60° with 7.4 %MVC at 0° shoulder abduction and 9.4 %MVC at 30° shoulder abduction for NEXO, and 5.5 %MVC at 0° shoulder abduction and 7.0 %MVC at 30° shoulder abduction for EXO_A (all p-values < 0.05). Furthermore, a significant increase in mRNSM_{rel} was found between SEA120° to SEA90°, but only for EXO_A in 30° shoulder abduction with a mean increase of 4.8 %MVC (p – value < 0.01). For SEA60°, mRNSM_{rel} showed a significant increase from SAA30° to SAA0° with a mean increase of 7.2 %MVC and 4.8 %MVC for NEXO and EXO_A, respectively (p – value < 0.01). Finally, a significant increase in mRNSM_{rel} was found between SEA60° to SEA120°, but only for 30° shoulder abduction with a mean increase of 11.2 %MVC for NEXO and 11.7 %MVC for EXO_A (both p – values < 0.01).

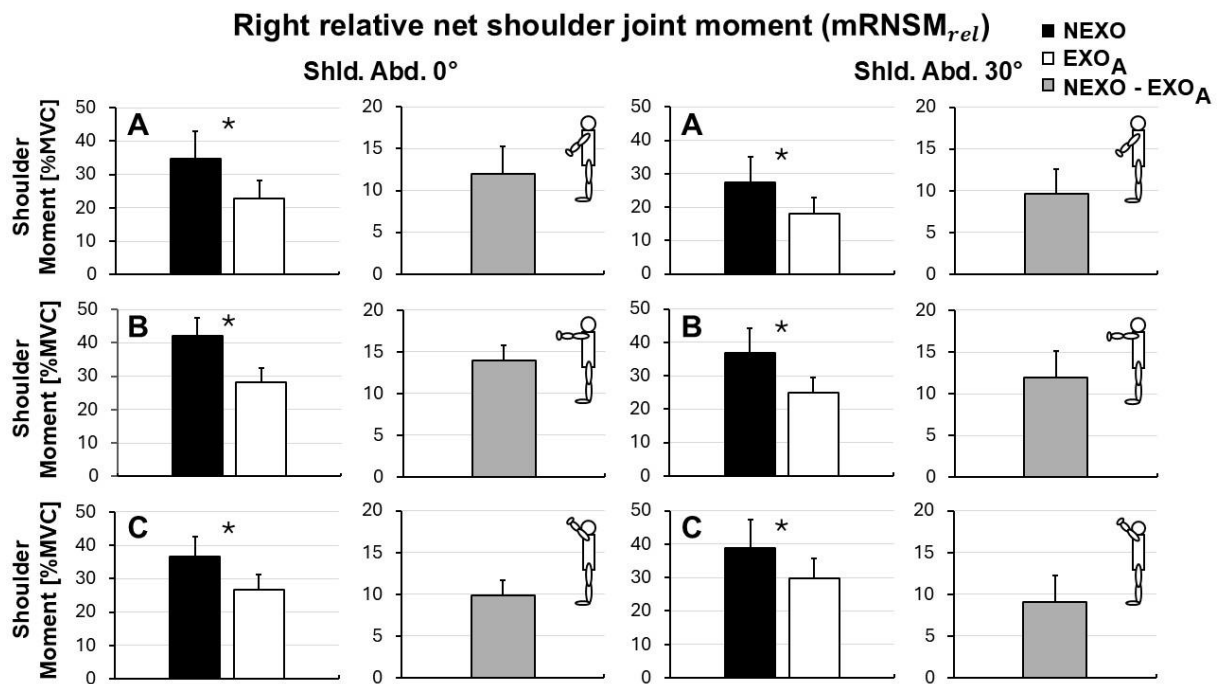


Figure 27: Mean bars of the mean (m) right relative net shoulder joint moment during static arm elevations in 60° (A), 90° (B) and 120° (C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXO_A) and without (baseline, black, NEXO) a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference (p < 0.05) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXO_A conditions.

Statistical analysis showed a significant main effect in mRAD for EXO_A and SEA (both $p - \text{values} < 0.01$). Further pairwise comparisons revealed a significant reduction in mRAD when wearing the exoskeleton ($p - \text{value} < 0.01$), independent of shoulder elevation or abduction angle, with the highest mean reduction for SEA90° with -10.2 %MVC for SAA0° and -7.9 %MVC for SAA 30° (Figure 28 B). For SEA60°, mRAD showed a mean reduction between EXO_A vs NEXO condition of -6.2 %MVC and -6.6 %MVC for SAA0° and SAA30°, respectively (Figure 28 A). Furthermore, a mean reduction was revealed in mRAD EXO_A compared to NEXO for SEA120° with -7.0 %MVC at 0° shoulder abduction and -5.0 %MVC at 30° shoulder abduction (Figure 28 C). For the main effect in SEA, pairwise comparisons showed significant mean increase in mRAD only for SEA120° compared to SEA60° in all combinations of EXO_A x SAA ($p - \text{value} < 0.01$), by 10.9 %MVC at 0° shoulder abduction and by 4.4 %MVC at 30° shoulder abduction for NEXO, and by 10.1 %MVC at 0° shoulder abduction and by 6.1 %MVC at 30° shoulder abduction for EXO_A. No interaction effect and no significant main effect for SAA were found for mRAD (all $p - \text{values} > 0.05$).

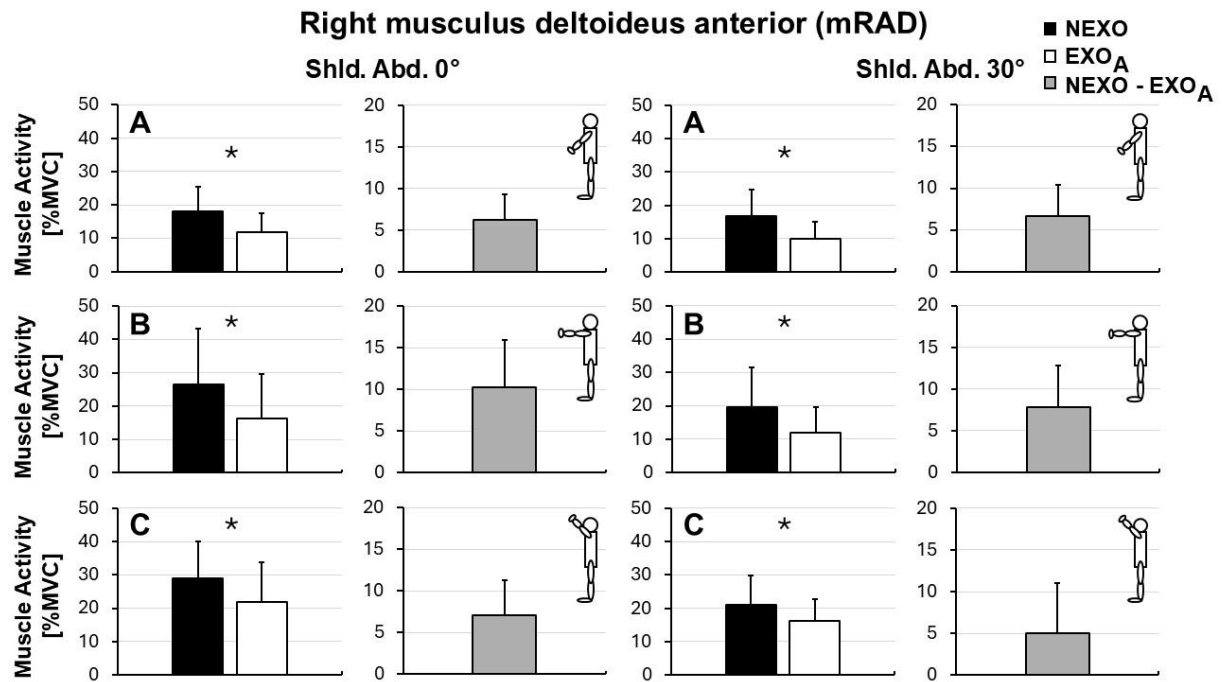


Figure 28: Mean bars of the mean (m) activity for the right anterior deltoid muscle during static arm elevations in 60° (A), 90° (B) and 120°(C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXO_A) and without (baseline, black, NEXO) a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference ($p < 0.05$) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXO_A conditions.

For the dependent variable mRMD, repeated measures ANOVA showed a significant main effect for SEA (p – value < 0.01). Further pairwise comparison revealed significant reduction for SEA60° to both SEA90° and SEA120° independent of the factors EXO_A and SAA (both p – values < 0.01). However, no significant difference was detected between SEA90° and SEA120° (p – value > 0.05). Compared to SEA90°, mRMD showed a reduction in SEA60° by -7.2 %MVC (SAA0°) and -4.6 %MVC (SAA30°) for NEXO and by -5.4 %MVC (SAA0°) and -3.9 %MVC (SAA30°) for EXO_A. A reduction by -7.3 %MVC (SAA0°) and -7.2 %MVC (SAA30°) for NEXO and by -5.1 %MVC (SAA0°) and -6.9 %MVC (SAA30°) for EXO_A was found in mRMD for SEA60° vs SEA120°. No interaction effects between the factors and no significant main effects for EXO_A and SAA were found for mRMD (all p – values > 0.05 ; Figure 29).

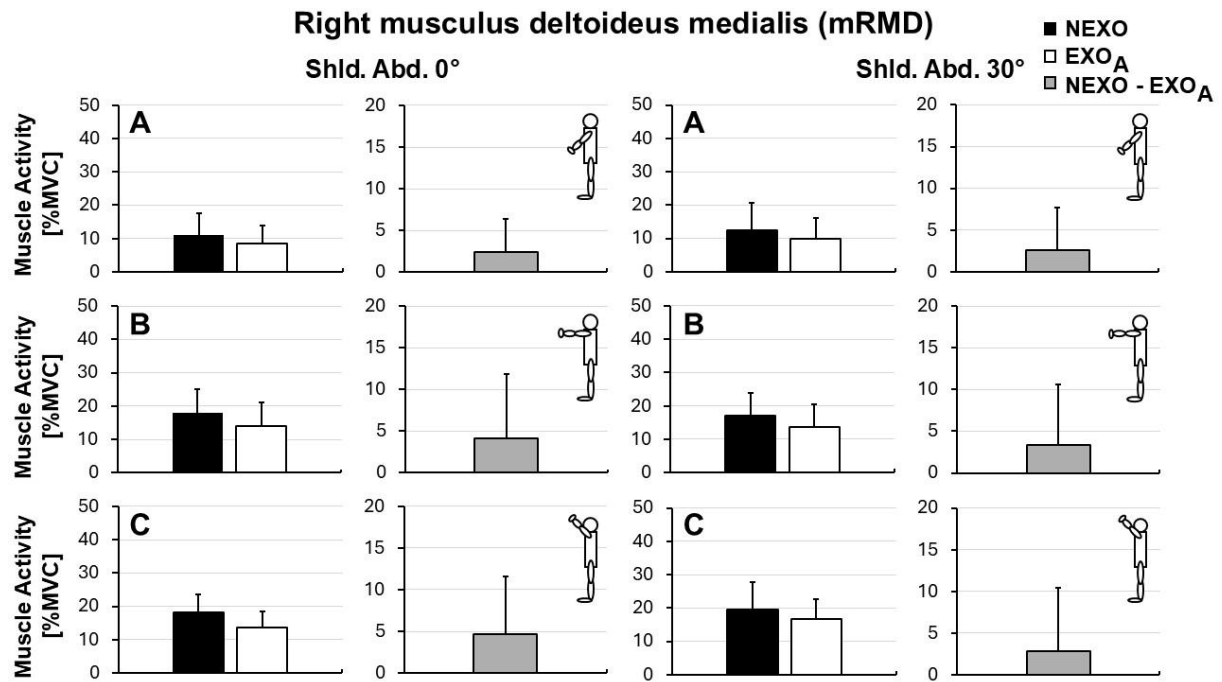


Figure 29: Mean bars of the activity for the right medial deltoid muscle during static arm elevations in 60° (A), 90° (B) and 120°(C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXO_A) and without (baseline, black, NEXO) a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference ($p < 0.05$) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXO_A conditions.

9.7. Discussion and limitations

The aim of this study was to conduct an exploratory analysis of the influence of an occupational shoulder exoskeleton (OSE) on mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand during static arm elevation across six arm postures, i.e. combinations of three different shoulder elevations angles (SEA) and two shoulder abduction angles (SAA). Therefore, the relative net shoulder joint moment for the right shoulder ($RNSM_{rel}$; variable: $mRNSM_{rel}$) and the muscle activity in the right musculus deltoideus pars clavicularis (right anterior deltoideus, RAD; variable: $mRAD$) and musculus deltoideus pars acromialis (right medial deltoideus, RMD; variable: $mRMD$) were measured with and without the OSE. They served as indicators for the changes in the mechanical load at the GHJ and shoulder muscle demand, respectively. The exoskeleton used in this study was the Levitate Airframe.

Across all six arm postures, the $RNSM_{rel}$ was reduced by an average of 23.5 - 34.9% while wearing the OSE. This can be attributed to a decrease in the absolute net shoulder joint moment for the right shoulder ($RNSM_{abs}$), i.e. the moment required to raise the arm to the desired position, due to the support provided by the OSE. The $RNSM_{abs}$ reduction ranged from -2.6 to -6.4 Nm, corresponding to a reduction of 14.5% to 39.6% when compared to the condition without the OSE. As the net shoulder joint moment decreased, it can be assumed that the mechanical load at the GHJ also decreased as contact forces, particularly the compressive forces, in the GHJ should be reduced (Praagman et al., 2000; van der Have et al., 2022). In addition, wearing the OSE also resulted in a reduction in right anterior deltoideus muscle activity ranging from 23.6% to 39.9% on average, independent of the arm posture, i.e. shoulder elevation and abduction angle. This is accordance with the results of a previous study, as the decrease in the right net shoulder joint moment should further result in lower RAD muscle activity (Anton et al., 2001).

Thus, the greatest reduction in the right net shoulder joint moment (both relative and absolute) was found when the arm was at 90° elevation (approximately 13 %MVC for both abduction angles). This result was expected, as this angle of elevation was close to the angle at which the maximum support moment of the OSE was determined in the previous study under the defined support setting (here: 94.0°, see chapter 8).

Similarly, the muscle activity in the right anterior deltoid muscle also showed the greatest reduction when wearing the OSE in the same arm posture (approximately 9 %MVC, independent of the shoulder abduction angle). Larger differences in the arm elevation angle from 90° (either higher or lower) were expected to result in less reduction in both the relative net shoulder net joint moment and shoulder muscle activity as the support provided by the exoskeleton used was assumed to be less (de Vries et al., 2019). This assumption is supported by the observed reductions in $mRNSM_{rel}$ and $mRAD$ which were both lower when the arm was elevated at 60° or 120° (between 9-12 %MVC for $mRNSM_{rel}$ and 5-7 %MVC for $mRAD$). However, for all arm postures, the mean muscle activity in RAD still exceeds the lower threshold for mean (or median) muscle demand of 10 %MVC proposed by Jonsson (1982) even with the support provided by the OSE. In addition, when the arm was elevated in 120° with the OSE, RAD muscle activity even remains above the upper threshold of 14% (Jonsson, 1982). Therefore, the arm elevation in those postures under the applied external load condition should not be maintained for more than an hour (Jonsson, 1982) even when working with the OSE.

Similar effects on net shoulder joint moment and partially on AD muscle activity have been shown in previous studies for either commercially available or prototype OSEs during static arm elevation (Brunner et al., 2023; Desbrosses et al., 2021; de Vries et al., 2019; Pacifico et al., 2020) and simulated industrial tasks (e.g. Pinho & Forner-Cordero, 2022; Sylla et al., 2014; van der Have et al., 2022). De Vries et al. (2019) reported a reduction in the absolute net shoulder joint moment and anterior deltoideus muscle activity while wearing an OSE during identical arm postures (fully extended arm with shoulder elevation angle between 30° - 150°) with an estimated exoskeleton support of up to 6.1 Nm. They also observed the greatest reduction in net shoulder joint moment with the OSE at 90° shoulder elevation, further suggesting OSE support is highest around this angle. Pinho and Forner-Cordero (2022) found larger reductive effects on AD muscle activity when using the OSE SuitX during a simulated screwing task in ≈90° shoulder flexion compared to ≈45° and ≈115° (moderate effects). This was expected from the authors as they determined the maximal support moment for the SuitX to occur between ≈70° and ≈100° shoulder elevation (Pinho & Forner-Cordero, 2022). Therefore, this suggestion for the support moment and conversely the resulting effect on AD muscle activity are in accordance with this study and de Vries et al. (2019).

Desbrosses et al. (2021) and Brunner et al. (2023) reported reduced shoulder muscle activity during static arm elevations for different arm postures using the Skelex and the Delta Suit exoskeletons, respectively. However, Desbrosses et al. (2021) found a slightly higher reduction in AD muscle activity of 13.4 %MVC (compared to 8-10 %MVC in this study) during 90° shoulder elevation (90° elbow flexion, 2 kg additional load). This may be due to the slightly higher exoskeleton support of 6.5 Nm provided by the Skelex compared to the OSE support provided in this study (mean support moment for 90° shoulder elevation ~ 6.3 Nm).

In contrast to this study, Brunner et al. (2023) measured a reduction in medial deltoideus muscle activity when performing static arm elevations at 90° and 120° shoulder elevation (arm fully extended) with the DeltaSuit exoskeleton. However, as the elevations in their study were performed without additional weight, it can be assumed that the net shoulder joint moment required to maintain the arm in the specific postures was lower. This may have led to a greater compensatory effect of the exoskeleton on muscle demand and, consequently, muscle activity, as indicated by the higher reduction in AD muscle activity of 11.5 %MVC at 120° shoulder elevation (extracted from the supplementary material of Brunner et al., 2023). In addition, as revealed in the study by Pinho and Forner-Cordero (2022), an OSE can have moderate to larger reductive effects on AD muscle activity when utilized during tasks primarily performed in the sagittal plane, but simultaneously only small to moderate effects (and partially no effect depending on the shoulder elevation angle) on MD. Finally, the OSE's effect on the medial deltoideus may increase if the arm elevation is performed entirely in the frontal plane (90° shoulder abduction; Tetteh et al., 2022). This would increase its activity, since this muscle is primarily required for arm elevation in the frontal plane (Echtermeyer & Sangmeister, 1996; Lee & An, 2002). Consequently, the OSE's effect may shift from the anterior deltoideus to the medial deltoideus. However, small reductions (although not significant) were found for the medial deltoideus in this study which may still be ergonomically relevant in reducing muscle fatigue when working with the exoskeleton for longer periods of time (van der Have et al., 2022). Especially when the arm was elevated at 60°, the muscle activity in MD was reduced below 10 %MVC (threshold suggested by Jonsson, 1982).

When considering the influence of shoulder elevation and abduction angle on the OSE effect, the results showed that the overall reduction in relative net shoulder joint moment and anterior deltoideus muscle activity was independent of both elevation and abduction angle. However, it can be suggested that the magnitude of the reduction in relative net shoulder joint moment is more influenced by the elevation angle (mean difference between the three angles up to 4.0 %MVC) rather than the abduction angle (mean difference between two angles up to 2.4 %MVC). This was expected as the exoskeleton support is dependent on the rotational angle in the exoskeleton joint, which is directly related to the shoulder elevation angle. Furthermore, these results are consistent with the findings by de Vries et al. (2019), who also reported that the effect of the exoskeleton in reducing the absolute net shoulder joint moment was more dependent on the shoulder elevation angle than on the abduction angle.

Several limitations of this study must be considered when interpreting the results. First, the arm postures were chosen to exploratively analyze the influence of the OSE on mechanical load at the GHJ and shoulder muscle demand during static arm elevation. However, these postures do not necessarily represent common working arm postures. In order to minimize shoulder joint moment and muscle demand, workers may adopt more ergonomic postures when performing their work tasks, e.g., working with the arms closer to the body (Anton et al., 2001; Chopp et al., 2010a; Maciukiewicz et al., 2016). Therefore, the effect of the OSE may vary and could even be nullified in certain postures, such as when working with the arms below shoulder height (Pinho & Forner-Cordero, 2022). In addition, to minimize inter-individual differences in exoskeleton support moment and increase internal validity, arm postures were standardized across participants and the hand heights were scaled based on their anthropometrics. Although this procedure is recommended to reduce muscle demand when working with the arm overhead (Chopp et al., 2010a), it is often not practical in the workplace. Therefore, it can be expected that the variation in the effect of the OSE will increase when implemented in a non-height-adjustable workstation due to the adaptation of individual arm postures during work.

Second, the support setting, i.e. the moment-angle curve, of the OSE was constant between the participants. Therefore, only small variations in the support moment were expected across all participants. However, if the support is not chosen according to the demands of the task, participants' effort in performing this task may increase (Sylla et al., 2014).

Adapting the exoskeleton to the required task demands (here: participant's arm weight), as was done in the study by Pacifico et al. (2020), could therefore increase the exoskeleton's effect in reducing muscle activity in the arm elevators and thus shoulder muscle demand.

Finally, the study focused solely on the influence of the OSE on the shoulder muscles primarily involved in arm elevation, namely the anterior and medial parts of the deltoideus muscle (Gohlke, 2002). As a consequence, it was not possible to determine the effect of wearing the OSE in the analyzed static arm posture on the muscles responsible for shoulder girdle elevation and stability, such as the upper trapezius muscle (Gohlke, 2002; Peat, 1986), or on glenohumeral joint stability, such as the rotator cuff muscles, e.g. the infraspinatus muscle (Peat, 1986). Although, many studies have already found beneficial effects of different OSEs in reducing upper trapezius muscle demand (e.g. de Vries et al., 2019; van Engelhoven et al., 2019), ambiguous results have been reported for the rotator cuff muscles, especially the infraspinatus (Ojelade et al., 2023; van Engelhoven et al., 2019).

9.8. Conclusion

In conclusion, the occupational shoulder exoskeleton decreases mechanical load at the glenohumeral joint and, to some extent, decreases muscle demand in the arm elevator muscles during static arm elevation by reducing the net shoulder joint moment and muscle forces, respectively. The magnitude of this effect was primarily influenced by the shoulder elevation angle and was independent of the shoulder abduction angle. However, it is unclear, if the relieve of the muscle through the exoskeleton support sufficiently reduces muscle demand to prevent muscle fatigue as the remaining muscle activity could be still too high, especially when working for a longer time period, e.g. one hour or more (Jonsson, 1982). In addition, the reported effect may not fully transfer to field situations where the OSE is used during an actual work task (de Bock et al., 2021; Pacifico et al., 2023). Therefore, the results of this study should be interpreted with caution, and further research is needed to evaluate the OSEs effectiveness, particularly in real working scenarios.

10. Influence of a shoulder exoskeleton on mechanical load and muscle tissue oxygenation during static arm elevation

The aim of this study was to conduct an exploratory analysis of the influence of an occupational shoulder exoskeleton (OSE) on mechanical load at the glenohumeral joint (GHJ) and muscle tissue oxygenation (MTO) in the shoulder during static arm elevation with varying shoulder joint load. Therefore, the net shoulder joint moment for the right shoulder (RNSM) and the tissue oxygenation index (TOI) in the right musculus deltoideus pars clavicularis (right anterior deltoideus, RAD) were measured. They were selected to serve as indicators for the changes in the mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand, respectively. The exoskeleton used in this study was the Paexo Shoulder.

10.1. Participants

Twelve young, healthy participants (seven males, five females, mean $\pm$ standard deviation; 22.8 years $\pm$ 1.4, 1.8 m $\pm$ 0.1, 71.9 kg $\pm$ 12.3) participated voluntarily in this study which was approved by the ethics review board of the University of Applied Sciences Koblenz. Prior to the measurements, study participants gave informed consent and filled out a questionnaire on health status. All participants reported no musculoskeletal or cardiovascular diseases or the intake of any medication before the measurement.

10.2. Experimental design and task description

A three-factorial repeated measures design with the independent variables “exoskeleton (EXO_P)”, “arm posture (AP)” and “external shoulder joint load (SJ_{Load})” was applied. The participants performed the elevation in the sagittal plane with (EXO_P) and without (NEXO) the OSE for three different arm postures (Figure 30):

- 1) shoulder flexion (SF) of 90°, elbow flexion (EF) of 0° ($AP_{EF0^\circ}^{SF90^\circ}$);
- 2) shoulder flexion (SF) of 90°, elbow flexion (EF) of 90° ($AP_{EF90^\circ}^{SF90^\circ}$); and
- 3) shoulder flexion (SF) of 135°, elbow flexion (EF) of 0° ($AP_{EF0^\circ}^{SF135^\circ}$).

During the elevation, the arm was internally rotated with the palms pointing in medial direction. To simulate different external loads, such as varying tool weights, the third independent variable SJ_{Load} was defined as 20% and 30% of the maximum shoulder joint moment ($\tau_{max}^{shoulder}$) in the sagittal plane which was obtained during maximal voluntary contraction (MVC) trials.

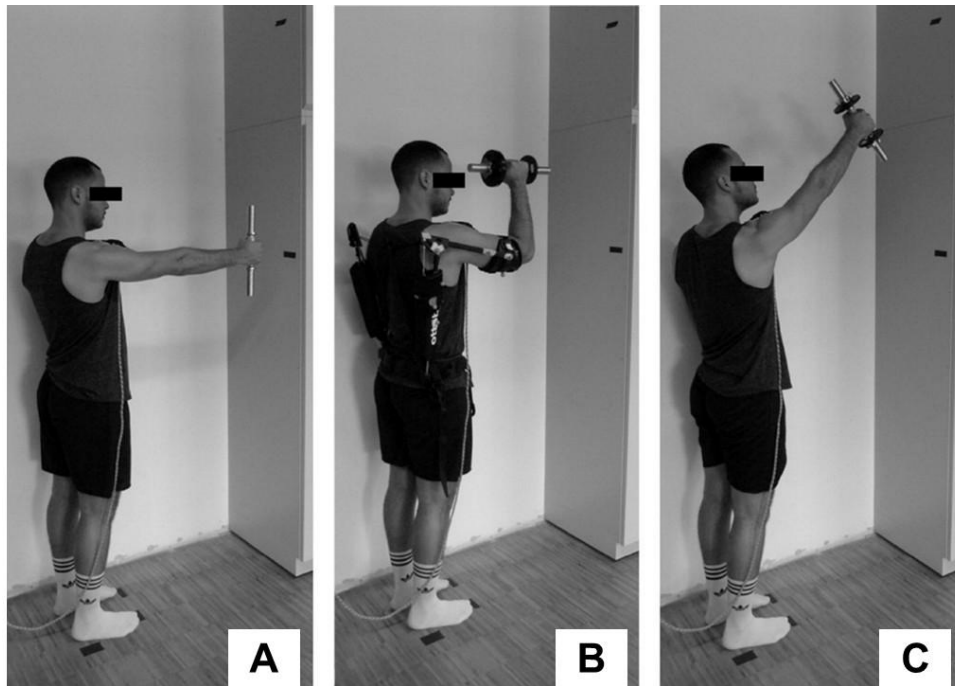


Figure 30: Participant performing an arm elevation in the sagittal plane with (B) and without (A,C) an shoulder exoskeleton in three arm postures (values for SF and EF specify the respective shoulder flexion and elbow flexion angle): (A) SF = 90°, EF = 0°, (B) SF = 90°, EF = 90°; and (C) SF = 135°, EF = 0°. Black markers were placed on the wall in front of the participant and aligned to their anthropometric measurements to help maintain the desired shoulder elevation angle.

The two load levels were based on findings from Barcroft and Millen (1939), who reported an impairment of muscle blood flow following contractions at 20 %MVC and 30 %MVC. To achieve the desired shoulder joint moment, specific weights were placed in the participant's hands increasing the external hand load (Figure 30). Additionally, for a given SJ_{Load} the shoulder joint moment was kept constant across all arm postures to isolate joint angle-dependent effects on MTO (Hisaeda et al., 2001).

10.3. Measurement instrumentation and data acquisition

First, segment lengths and circumferences of the participant's right arm (upper arm, forearm and fist) and body weight were obtained following the instructions from Hanavan (1964), to calculate inertial properties, similar to the previous study. Subsequently, participants performed three MVC trials in the $AP_{EF0^\circ}^{SF90^\circ}$ posture. Therefore, they vertically pulled on a digital scale (DIGIFLEX Hanging Digital Scale, Digiflex Ltd, Hockley, UK) fixed to the floor with the scale positioned at the height of the GHJ center in $AP_{EF0^\circ}^{SF90^\circ}$ to ensure a shoulder flexion of 90° .

Before the measurement, participants were instructed to adopt a comfortable, hip wide stance, maintaining a straight back and keeping the shoulders aligned parallel to the hips (i.e., avoiding anterior-posterior displacement of the GHJ). The scale was then placed in their right hand, and they were asked to lift the arm to reach the required posture. To avoid injuries, participants were instructed to gradually build up the vertical force over three seconds, hold the maximum contraction for another three seconds, and then slowly release the force before lowering the arm back into neutral position. This procedure was repeated three times, and the maximum vertical force was calculated by averaging the maximum values obtained from each repetition. During MVC trials, participants received verbal non-threatening encouragement and were instructed to primarily engage the shoulder muscles avoiding to "lift from the back" or "raise the shoulder (GHJ center)", thereby, minimizing the involvement of back- or neck muscles. Three-minute rest was provided between each repetition.

Similar to the previous study, $\tau_{max}^{shoulder}$ was estimated as the sum of the gravitational shoulder joint moment ($\tau_{grav}^{shoulder}$) and the maximum elevation moment (τ_{max}^{elev}):

$$\tau_{max}^{shoulder} = \tau_{grav}^{shoulder} + \tau_{max}^{elev} \quad (10.1)$$

The shoulder gravitational joint moment ($\tau_{grav}^{shoulder}$) was calculated as:

$$\tau_{grav}^{shoulder} = F_{grav}^{arm} * r_{CoG}^{arm} \quad (10.2)$$

where is F_{grav}^{arm} the weight force of the arm including the dumbbell and r_{CoG}^{arm} is the distance between the GHJ and the center of gravity (CoG) of the arm.

The maximum shoulder elevation moment ($\tau_{\max}^{\text{elev}}$) was defined as:

$$\tau_{\max}^{\text{elev}} = F_{\max}^{\text{elev}} * l^{\text{arm}} \quad (10.3)$$

with the maximum elevation force ($F_{\max}^{\text{elev}}$) obtained from the MVC trials prior to the measurements and l^{arm} as the arm length (as the angle between the arm and the force direction should be 90°).

Afterwards, the OSE was adjusted to the participant's anthropometry with all participants reporting a perfect fit. The OSE's support moment was adjusted based on participant's subjective feedback. Participants were asked to report whether they clearly perceived support from the OSE while holding the external load for the lowest SJ_{Load} condition in posture $AP_{EF0^\circ}^{SF90^\circ}$. However, since full compensation of the arm (and tool) weight is not recommended by the manufacturer, the arm should still be able to drop due to gravity.

To estimate the MTO in the RAD, near-infrared spectroscopy (NIRS) was applied as it is a valid non-invasive method to measure MTO in the capillaries (de Ruiter et al., 2007; Elcadi et al., 2011; Perrey et al., 2010a; Perrey et al., 2010b). A light source (LED or laser), placed over the investigated muscle, emits infrared light through the skin layers into deeper tissues of the human body, which travels along a banana-shaped curve (Figure 31). This light is absorbed and scattered by the deeper tissues and the reflected intensity is measured by a detector (a dual-segment photodiode chip), with a penetration (measurement) depth of approximately equal to half the distance between the light source and the detector (Figure 31; de Ruiter et al., 2007). The light source and the detector are therefore combined (optode) with a fixed distance (Figure 31). The NIRS method has already been used in ergonomic studies, involving upper extremity movement as reliable and sensitive measures for MTO (Ferguson et al., 2013; Mehta et al., 2014; Perrey et al., 2010b).

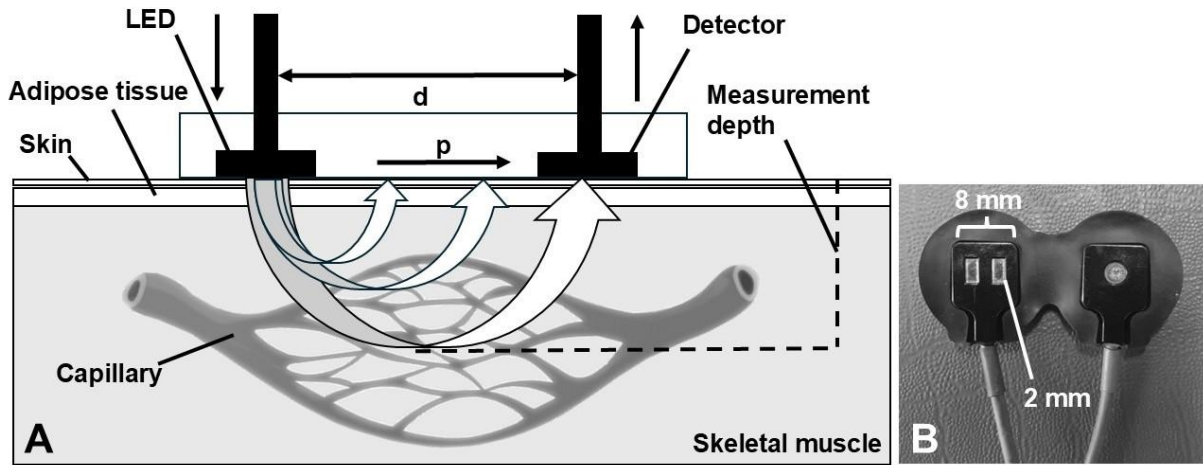


Figure 31: Near-infrared spectroscopy (NIRS) procedure (A). Infrared light is emitted (LED) into deeper muscle tissues (capillary level) and the reflected intensity along the path p is measured via a detector. The measurement depth is approximately half the distance between LED and detector (d) which are combined in a probe (optode; B). The graphic of the capillary was modified from CERT (n.d.), licensed under CC BY-SA 3.0, via Wikimedia Commons. Changes were made by the author.

Measurement preparations were finished by placing the NIRS-optode in a probe with a fixed diode-detector distance of 30 mm (measurement depth about 15 mm) on RAD. The NIRS data was recorded with the NIRO-200NX (5 Hz; continuous-wave measurement, Hamamatsu Photonics, Shizuoka Prefecture; Japan). The tissue oxygenation index (TOI) as a measure of MTO was recorded and analyzed as this parameter is closely linked to oxygen saturation of hemoglobin. TOI is defined as the percentage ratio between the concentration (c) of oxygenated (O_2Hb) and total hemoglobin (THb) and is calculated for a time point t as

$$TOI(t) = 100 * \frac{c(O_2Hb)}{c(THb)} \quad (10.4)$$

Thereby, these concentrations are calculated with the spatially resolved spectroscopy method by measuring the changes in the light reduction along the path p (Figure 31; Hamamatsu, 2012). To ensure a stable and standardized measurement of the NIRS-system, the optodes placement was determined using anatomical landmarks (Demura et al., 2007) following guidelines for EMG electrodes from Zipp (1982).

Prior to fixing the probe with light-impervious adhesive tape, the skin was shaved and the adipose tissue thickness (ATT; skin-fat-layer) was measured using ultrasound as a high ATT could influence the measurements (Lin et al., 2000; Yang et al., 2005). Since ATT values for AD were lower than 5 mm for all participants (mean $\pm$ standard deviation; 3.8 mm $\pm$ 0.6), a low subject-to-subject variation in the fat optical coefficients can be expected, and therefore, no corrective measures had to be applied (Yang et al., 2005).

Afterwards, the participants performed static arm elevation tasks under twelve experimental conditions (all combinations of “EXO_P”, “AP” and “ SJ_{Load} ”). A three-step randomization protocol was applied in order to minimize potential sequential effects caused by muscle fatigue:

- 1) First, the factor “EXO_P” was counterbalanced by having six subjects start with the exoskeleton and six without it. To maintain the NIRS-sensor fixated and thereby minimize the influence of different sensor locations, each participant completed all combinations of the remaining two factors with the OSE before repeating the procedure without the OSE (and vice versa).
- 2) Second, the factor SJ_{Load} was counterbalanced.
- 3) Third, the starting arm posture changed in ascending order for the participants, whereby every fourth participant started with $AP_{EF0^\circ}^{SF90^\circ}$. Then, the arm postures were continued in enumerative order.

In summary, twelve sequences of experimental conditions were constructed and assigned to individual participants. Before each measurement block (EXO_P/NEXO), the participants rested on a chair for five minutes. Then, they performed the static arm elevation task according to their assigned sequence. Each measurement sequence followed the same structure: 20 seconds before arm elevation, the required weight was placed in the participant’s dominant hand. Participants then performed the elevation in the designated posture for 60 seconds (Figure 32). Between each arm elevation, three-minute breaks were provided during which the participants rested on a chair and were instructed to move “as little as possible”.

10.4. Data processing and dependent variables

Data analysis was performed using Python (Software Spyder, Version 3.7.6, Python Software Foundation, Delaware, USA). First, the data were filtered with a moving average filter (window size 5 s, no overlay). Two specific time points were defined for each measurement (Figure 32): immediately before executing an arm elevation (t_0) and five seconds before completing the arm elevation (t_1). For the statistical analysis, the change in TOI over time was obtained from RAD and the dependent variable ΔTOI was calculated as the difference between the two time points ($t_1 - t_0$) in order to minimize the influence of increasing muscle fatigue (Perrey et al. 2010b; Figure 32).

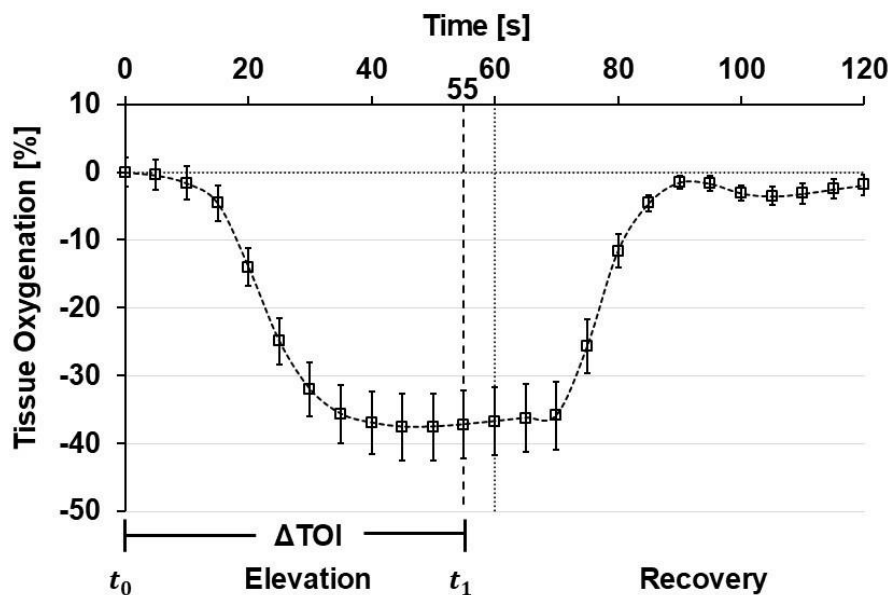


Figure 32: Mean changes in tissue oxygenation index (TOI) over time during an arm elevation (duration 60 s) in 90° shoulder flexion (elbow flexion 0°) and subsequent recovery phase (section of 120 s) with the arm lowered in neutral position. Time points t_0 mark the start point of the arm elevation. Change in TOI (ΔTOI) was calculated as the difference between the respective TOI values measured at the two time points t_0 and t_1 (55 s before the end of an arm elevation). Error bars indicate $\pm$ standard error of mean.

To calculate the support moment provided by the OSE during arm elevation for the configuration chosen by the participant, the corresponding polynomial equation obtained for the Paexo (presented in the chapter 8) was applied with the shoulder flexion angle (90° or 135°) as input variable.

Thus, the net shoulder joint moment (RNSM) when using the exoskeleton was defined as the difference of the $\tau_{grav}^{shoulder}$ and the support moment provided by the exoskeleton (τ^{exo}):

$$RNSM = \tau_{grav}^{shoulder} - \tau^{exo} \quad (10.5)$$

For all NEXO trials, RNSM was equal to $\tau_{grav}^{shoulder}$.

10.5. Statistics

Statistical analysis was performed using R (version 1.2.5042, RStudio, Inc., Boston, MA). The dependent variable included in the analysis was ΔTOI serving as an indicator of the change in MTO. A three-way repeated-measures ANOVA was conducted for the dependent variable to assess main and interaction effects for the three independent variables: EXO_P , AP and SJ_{Load} (2 x 3 x 2 repeated-measures design). The normal distribution of the residuals was visually inspected using Q–Q (quantile-quantile) plots. Boxplots were used to detect outliers and sphericity was evaluated with the Mauchly test applying a Greenhouse–Geisser correction, if necessary. The significance level was set at 5% and post-hoc pairwise comparisons were conducted using Student's t-test with Bonferroni-Holm correction.

10.6. Results

Due to measurement errors, the data of one participant were excluded and all data analyses were conducted for the remaining eleven participants.

For the right net shoulder joint moment (RNSM), a reduction was found when wearing the exoskeleton in both external shoulder joint load conditions (SJ_{Load}) for all three arm postures (AP, Table 6).

Table 6: Mean right net shoulder joint moment (RNSM) $\pm$ standard deviation during static arm elevation in 90° and 135° shoulder flexion (SF) with (EXO_P) and without (NEXO) a shoulder exoskeleton for an external sagittal shoulder load (SJ_{Load}) of 20% and 30% derived from maximum voluntary contraction (MVC).

<i>RNSM</i> (Nm)		<i>SJ_{Load}</i> 20 %MVC		<i>SJ_{Load}</i> 30 %MVC	
		NEXO	EXO _P	NEXO	EXO _P
	SF90°	14.9 (5.6)	10.7 (4.1)	22.3 (8.4)	18.1 (6.9)
	SF135°		13.1 (4.9)		20.5 (7.7)

TOI decreased in all measurements regardless of the three factors (EXO_P, AP and SJ_{Load} , exemplary in Figure 33).

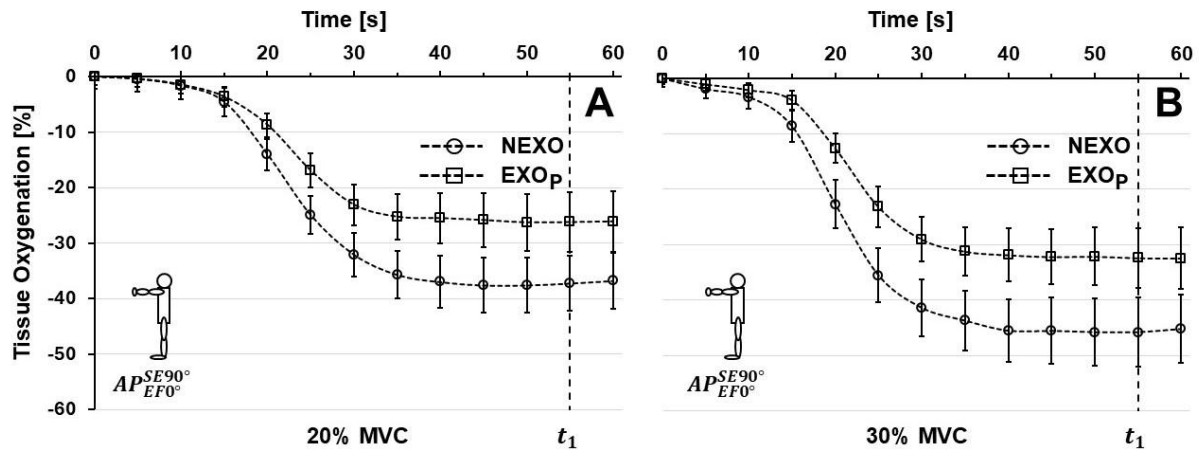


Figure 33: Mean changes in tissue oxygenation index (TOI) over time during an arm elevation (duration 60 s) in 90° shoulder flexion (SF) and 0° elbow flexion with (EXO_P) and without (NEXO) a shoulder exoskeleton for an external sagittal shoulder joint load of 20% (A) and 30% (B) derived from maximum voluntary contraction (MVC). Error bars indicate $\pm$ standard error of mean.

For ΔTOI , repeated-measures ANOVA showed significant main effects for all three factors ($EXO_P \times AP \times SJ_{Load}$; all p-values < 0.05). Subsequent pairwise comparison revealed significantly higher ΔTOI values when wearing the OSE, independent of posture or SJ_{Load} (Figure 34) with ΔTOI improving from -53.2% to -43.5%.

Regarding the main effect of arm posture, the $AP_{EF0^\circ}^{SF90^\circ}$ condition revealed the significantly highest ΔTOI values in all EXO_P and SJ_{Load} conditions (mean -35.4%) compared to $AP_{EF0^\circ}^{SF135^\circ}$ (mean -50.1%) and $AP_{EF90^\circ}^{SF90^\circ}$ (mean -48.2%; both p – values < 0.05). Furthermore, ΔTOI was significantly lower for the higher SJ_{Load} condition with -48.3% compared to the lower SJ_{Load} condition with -40.8% (p – value < 0.05). No significant differences were found between $AP_{EF90^\circ}^{SF90^\circ}$ and $AP_{EF0^\circ}^{SF135^\circ}$ and no interaction effects between the factors were observed for ΔTOI (all p – values > 0.05).

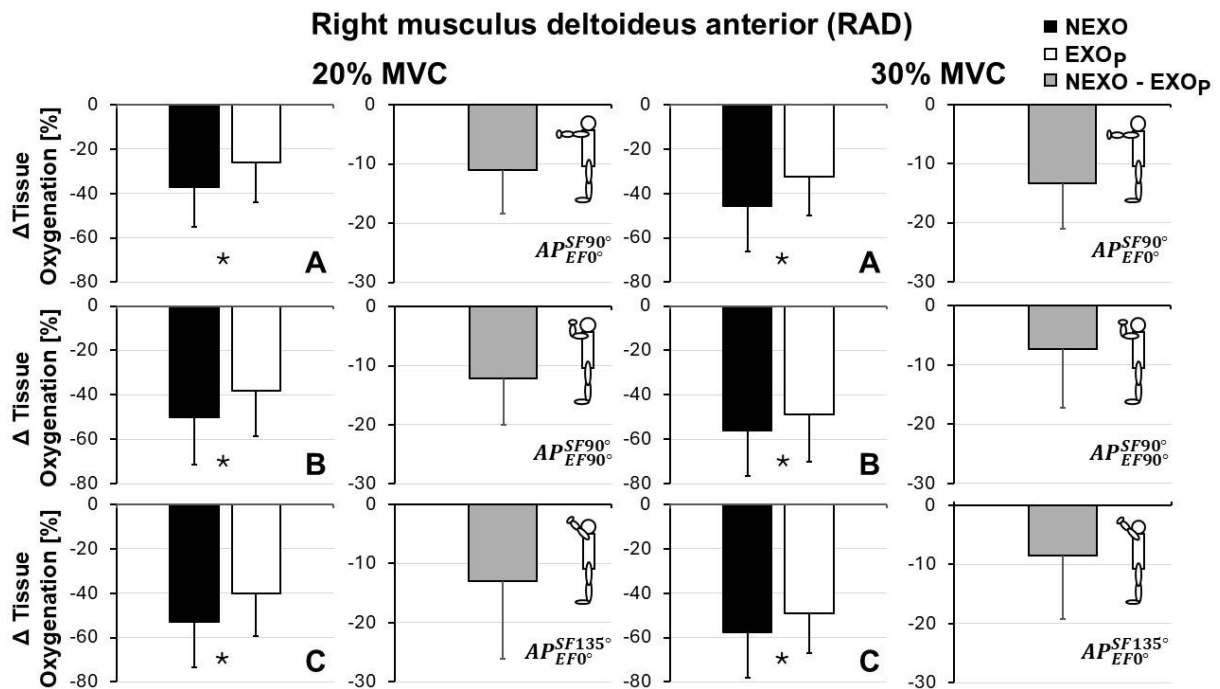


Figure 34: Mean bar plots for the changes in tissue oxygenation index (ΔTOI) during an arm elevation with (white, EXO_P) and without (black, NEXO) an shoulder exoskeleton in three static arm postures (AP; values for SF and EF specify the respective shoulder flexion and elbow flexion angle) for an external sagittal shoulder joint load of 20% and 30% derived from maximum voluntary contraction (MVC). Error bars indicate - standard deviation and stars (*) over the bars a significant difference (p < 0.05) between baseline and exoskeleton condition. Grey bars show the mean increase between EXO_P conditions (NEXO - EXO_P).

10.7. Discussion and limitations

The aim of this study was to conduct an exploratory analysis of the influence of an occupational shoulder exoskeleton (OSE) on mechanical load at the glenohumeral joint (GHJ) and muscle tissue oxygenation (MTO) in the shoulder for three arm postures (AP) in the sagittal plane under two external shoulder joint load (SJ_{Load}). Therefore, the net shoulder joint moment for the right shoulder (RNSM) and the tissue oxygenation index (TOI) in the right musculus deltoideus pars clavicularis (right anterior deltoideus, RAD) were measured. They were selected to serve as indicators for the changes in the mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand, respectively. The exoskeleton used in this study was the Paexo Shoulder.

When wearing the OSE, the RNSM was reduced on average by 8.1% to 28.8% across all arm postures for both external shoulder joint load conditions, therefore suggesting a decrease in mechanical at the GHJ load through decreased compression joint forces (Praagman et al., 2000). During arm elevation, TOI generally decreases in RAD independent of the experimental conditions (i.e., AP, SJ_{Load} or EXO_P). This suggest that MTO is reduced due to increased oxygen consumption caused by muscle contraction, potentially leading to ischemia and promoting the development of SMF (Ferguson et al., 2013). For both shoulder joint load conditions, significantly higher ΔTOI values in the anterior deltoideus were measured when wearing the OSE with an average increase between 7.2% and 13.4% across all arm postures corresponding an average increase of 12.9% to 29.7% compared to the baseline condition (without the OSE). This increase implies a higher MTO which may be attributed to reduced metabolic demands due to lower muscles forces required to maintain the static postures (Perrey et al., 2010a).

Weston et al. (2022) also revealed higher values for ΔTOI while performing repetitive simulated overhead work (here: exerting a vertical force) for one hour with three different OSE compared to a control condition. However, compared to this study, the differences in ΔTOI between the OSE and the baseline condition reported by Weston et al. (2022) were not large enough to reach the significance threshold for most conditions. The authors therefore suspected that their task was not sufficiently fatiguing due to the short exertion time of approximately 2 seconds per repetition.

For longer exertion phases, differences in ΔTOI between control and exoskeleton conditions may have been higher, since ΔTOI values tend to further decrease during sustained arm elevation until reaching a steady state (up to 30 seconds, as shown in this study and by Elcadi et al., 2011). This assumption is supported by the observation that differences in ΔTOI between EXO_P conditions may not be significantly different at the beginning of an arm elevation as indicated by overlapping standard errors during the first 5 seconds (Figure 33).

For the higher shoulder load joint condition, ΔTOI significantly decreased on average from -40.8% to -48.3%, independent of the EXO_P condition or arm posture. This decrease was expected as the higher shoulder joint load condition should have increased the muscle forces required to generate the higher shoulder moment (from 15.2 Nm to 22.9 Nm). This likely led to elevated metabolic demands and, consequently, greater muscle oxygen consumption. Similar findings were reported by Elcadi et al. (2011), who observed a decrease in oxygen saturation for the trapezius muscle with increasing isometric muscle contraction. Van Engelhoven et al. (2019) also reported higher shoulder agonist muscle activity with increasing tool load for a sustained overhead working task while receiving support from an OSE. Hence, for this study, it was assumed that an increase in external shoulder joint load may have compensated the effect of the OSE in reducing muscle forces during arm elevation resulting in similar muscle oxygen consumption and a smaller effect on ΔTOI . Similar assumptions can be drawn from Kim et al. (2018), who found a reduction in left AD median muscle activity when using an OSE during shoulder/overhead assembly tasks, which was only significant for the task with the expected lowest external load (here: wiring without tool compared to drilling with light/heavy tool). However, Desbrosses et al. (2021) reported an increased effect of the OSE on the reduction of muscle activity under a higher load condition (2 kg vs. 8 kg) during 90° static shoulder elevation with flexed elbows. Assuming no adjustment in exoskeleton support between load conditions, the resulting moment in all three studies (Desbrosses et al., 2021; Kim et al., 2018; van Engelhoven et al., 2019) should have increased with external load due to a higher shoulder elevation moment. Analogous to the lower SJ_{Load} condition, significantly higher values of ΔTOI in RAD when wearing the OSE were observed in all arm postures. These results imply that even with increased external shoulder joint load, the effect of the exoskeleton in reducing the gravitational shoulder joint moment, and consequently the required muscle forces, was still sufficient to beneficially influence MTO.

In the arm posture $AP_{EF\ 0^\circ}^{SF\ 90^\circ}$, the difference in ΔTOI in EXO_P conditions remained relatively stable across load conditions with 11.0% for 20 %MVC vs. 13.4% for 30 %MVC corresponding to 29.7% and 29.3% when compared to the baseline condition (without OSE). Thus, it can be suggested that the effect of the OSE in reducing muscle forces, and therefore conversely resulting in higher MTO, is similar between the load conditions for this arm posture. However, the two other arm postures ($AP_{EF\ 90^\circ}^{SF\ 90^\circ}$ and $AP_{EF\ 0^\circ}^{SF\ 135^\circ}$) showed a decrease in the effect with increasing load from 12.1% to 7.2% for $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$ (resp. from 24.1 % to 12.9% compared to the baseline) and from 13.0% to 8.5% for $AP_{EF\ 0^\circ}^{SF\ 135^\circ}$ (resp. from 24.5 % to 14.8% compared to the baseline). This finding was unexpected, since the external shoulder joint load was identical across the three arm postures in both load conditions. In addition, as the shoulder flexion angle was the same for the arm posture $AP_{EF\ 0^\circ}^{SF\ 90^\circ}$ and $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$, it can be assumed that the OSE provided the same support moment. Furthermore, the preconditions to generate the shoulder joint moment in $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$ (e.g., force-length relationship resulting from the same shoulder flexion angle) should be identical to $AP_{EF\ 0^\circ}^{SF\ 90^\circ}$. These results may suggest that even when the external load condition is identical between arm postures, MTO was lower in the postures $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$ and $AP_{EF\ 0^\circ}^{SF\ 135^\circ}$ compared to $AP_{EF\ 0^\circ}^{SF\ 90^\circ}$. This interpretation is further supported by the finding that both arm postures showed significantly lower ΔTOI values (mean ΔTOI -48.2% and -50.1%) compared to $AP_{EF\ 0^\circ}^{SF\ 90^\circ}$ (mean ΔTOI -35.4%) but not compared to each other, independent of the EXO_P and load conditions. For $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$, this difference may have resulted from higher internal muscle forces in RAD due to an additional stabilization to keep the lower arm in position. The center of gravity of the dumbbell might not have been in a horizontal line with the GHJ which would generate an external/internal rotational moment in the sagittal plane. To maintain the arm in a neutral position, the shoulder muscles, such as the deltoideus muscles, may have had to counterrotate the arm against this external/internal rotation moment, resulting in higher muscle forces. This hypothesis is supported by participant's feedback, as some of them described $AP_{EF\ 90^\circ}^{SF\ 90^\circ}$ as the "most physically strenuous" condition. Consequently, since the OSE primary supports the arm during elevation, it may have reduced the muscle forces, and thus the oxygen consumption, required to maintain the arm elevation, but not the additional forces required to stabilize the hand position.

For $AP_{EF 0^\circ}^{SF 135^\circ}$, the variation in ΔTOI – and consequently in MTO- compared to $AP_{EF 0^\circ}^{SF 90^\circ}$ may be explained by the differences in the shoulder flexion angle (Hisaeda et al., 2001), since the increased flexion of the arm could impair local blood flow in RAD. In this posture, the blood flow may already be restricted due to compression of blood vessels against surrounding muscle tissue or bone structures (e.g., the ribs; Rohter & Hyman, 1962), higher internal muscle forces (Hisaeda et al., 2001) or gravitational effects pushing the blood out of the arm. Thereby, similar to the assumptions made for $AP_{EF 90^\circ}^{SF 90^\circ}$, the OSE may have primarily contributed to reducing muscle forces required for arm elevation, but may not have been able to cause a reduction of MTO due to posture-dependent restrictions of the blood flow.

There are a few limitations for this study that should be considered when interpreting the results. First, the study population consisted of young, healthy, lean (low skin-fat-layer over the muscle) participants, who were able to perform static arm elevations without physiological constraints. Thus, it remains to be investigated whether similar results would be found for an older study population.

Second, the effect of an OSE on MTO was analyzed in static arm postures under standardized load conditions for one minute. Hence, it is unclear whether the findings can be transferred to different working conditions with other arm postures or even longer task duration. In addition, with prolonged muscle metabolic demands physiological adaptations such as improved blood supply to the working muscle are likely to occur.

Third, the support moment of the OSE was not adjusted during the measurements even when the load condition was increased. Thereby, an increase in the support moment, especially for the higher load condition, may have amplified the effects on MTO (Brunner et al., 2023).

Finally, it was not possible to simultaneously measure muscle activity for RAD. Such data could have contributed to a deeper understanding of the relationship between OSE and MTO under different arm postures due to an expected linear relationship between muscle activity and MTO for static muscle contraction (Elcadi et al., 2011).

10.8. Conclusion

In conclusion, across all experimental conditions, the support provided by the OSE led to a decrease in the mechanical load at the GHJ and higher muscle tissue oxygenation (MTO) in RAD during arm elevation. This increase in MTO may result from a decrease in muscle forces, therefore reducing muscle and metabolic demands. However, the findings suggest that the effect of the OSE on MTO interacts with both joint load and arm posture. Similar interactions were observed from the results of Gillette et al. (2020), who revealed different effects of another OSE in reducing muscle activity between the same static arm postures with increasing tool weight. The observed increase in MTO when using the OSE could contribute to a reduction in muscle fatigue in RAD during arm elevation, a conclusion also supported by previous studies reporting a reduction in muscle activity (Gillette et al., 2020; Schmalz et al., 2019; van Engelhoven et al., 2019). In addition, based on these results, it can be suggested that the benefits of an OSE on MTO could be transferred to other agonist and stabilizing muscles (e.g., trapezius and the rotator cuff), given the expected linear relationship between muscle activity and MTO during static muscle contraction (Elcadi et al., 2011). However, the OSE may also lead to an increased activity in certain agonistic muscles, such as the triceps brachii (de Vries & de Looze, 2019; McFarland & Fischer, 2019), potentially resulting in a decrease in MTO and a faster development of local fatigue for these muscles (Grieve & Dickerson, 2008; Hagberg, 1984). Therefore, the results of this study should be interpreted with caution, and further research is needed to evaluate the OSEs effectiveness for longer task durations and other muscle groups, such as shoulder stabilizing muscles (e.g. trapezius and infraspinatus).

11. Influence of shoulder exoskeletons on mechanical load and muscle demand during aircraft manufacturing tasks

The aim of this study was to conduct an exploratory analysis of the influence of three occupational shoulder exoskeletons (Airframe, Paexo and Skelex) on the mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand during three different (non-standardized) aircraft manufacturing tasks: drilling, riveting and screwing. Therefore, the left and right net shoulder joint moment (LNSM and RNSM with the variables mLNSM and mRNSM, respectively) and muscle activity (bilaterally) in four muscles were obtained:

- 1) the left and right musculus deltoideus pars clavicularis (anterior deltoideus, LAD and RAD; variables: mLAD and mRAD, respectively);
- 2) the left and right musculus deltoideus pars acromialis (medial deltoideus, LMD and RMD; variables: mLMD and mRMD, respectively);
- 3) the left and right musculus trapezius pars descendens (upper trapezius, LUT and RUT; ; variables: mLUT and mRUT, respectively); and
- 4) the left and right musculus infraspinatus (LINF and RINF; variables: mLINF and mRINF, respectively).

They were selected to serve as indicators for the changes in the mechanical load at the GHJ and shoulder muscle demand.

11.1. Participants

19 physically healthy employees (11 male, 8 female, 33.5 ± 11.1 years; 84.9 ± 19.3 kg; 176.8 ± 10.0 cm) of an aircraft manufacturing company voluntarily participated in the study which was approved by the ethics review board of the University of Applied Sciences Koblenz. The employees were recruited from two different work areas and divided into two groups based on their daily work activities. The first group (group A) consisted of 12 employees (8 male, 4 female) whose main work task, i.e. at the aircraft manufacturing company, is the assembly of mechanical structures, such as mounting brackets, inside and outside the aircraft fuselage. Those participants used tools of different weights for drilling and riveting. The other seven employees (5 male, 2 female) of the second group (Group B) are responsible for installing electrical cables and hydraulic circuits inside the aircraft fuselage at their workplace.

11.2. Workplace replica and task description

Participants were instructed to perform their real work tasks (mostly at shoulder or overhead level) in a simulated work environment without and with three different occupational shoulder exoskeletons (OSEs). For this purpose, two aircraft panels were mounted on a mobile platform at a fixed height and angle to simulate the real working conditions (Figure 35). The workplace conditions (e.g. height and angle of the panels) were maintained for all measurements and were not adapted to the anthropometry of the individual participants (non-standardized).



Figure 35: Mobile platform with two aircraft panels mounted at a fixed height and angle.

On the left panel, the participants of the first group (group A) were asked to complete an assembly task by removing and reattaching two metal brackets to two horizontal stringers (lower and upper stringer; Figure 36 A). The metal bracket was first removed by drilling into the two rivets that connect the bracket to the strut (drilling task; Figure 37 A). After the drilling task had been completed, the rivets and bracket were removed and the drill holes were deburred. The bracket was then reattached to the strut with two rivets using a riveting tool (riveting task; Figure 37 B).

The assembly task for the group B on the right panel consisted of installing (and removing) twelve mounting clips (screwing task; Figure 37 C) on four horizontal stringers (with three clips on each stringer), evenly spaced across each stringer (Figure 36 B & C).

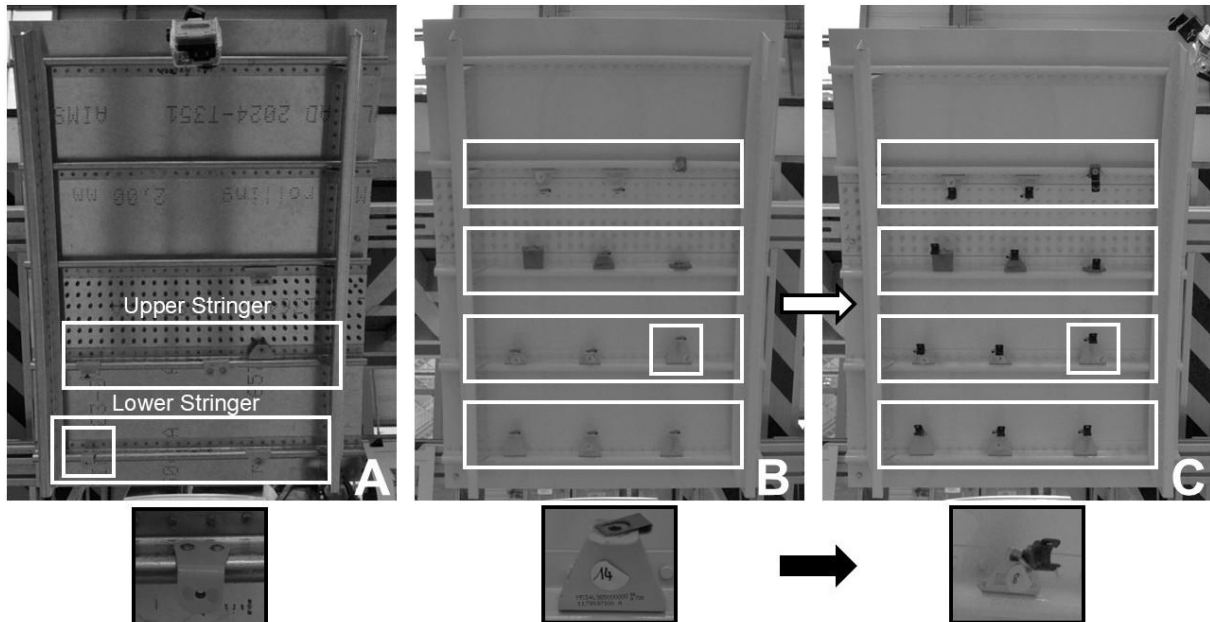


Figure 36: Two aircraft panels mounted on the mobile platform to perform a drilling and riveting task at two metal brackets each at a different stringer height (A) and screwing twelve mounting clips at four different stringer heights (B & C).

Naturally, at the real workplace inside the aircraft fuselage, the drilling is performed on an empty strut and the metal brackets are attached using the two drilled holes. Therefore, the brackets are usually only removed if assembly errors occur. However, in order to reduce material wear on the panel, this task was slightly modified compared to the real task in the aircraft fuselage. This modification was made in consultation with the aircraft manufacturer and in cooperation with ergonomic experts. Thus, the task performance and the task demands, i.e. the musculoskeletal load at the shoulder, were assured to be comparable to the real task. This task was completed three times on the two lowest horizontal struts of the panel: 1) once on the lowest strut (approximately at shoulder level) and 2) twice on the second lowest strut (approximately at eye level/overhead). This repetitive procedure was designed to increase the working time during the shoulder/overhead tasks.

Prior to the study, it was decided that the drilling and riveting tasks would only be performed on the two lowest of the four horizontal stringers. The two highest stringers were excluded because the majority of workers would use a stepladder to reach the specific heights of these two stringers. Therefore, similar shoulder elevation angles were expected when performing the task at the two highest stringers compared to working at the two lowest stringers, resulting in the same musculoskeletal shoulder load. For the screwing task, participants were instructed to unscrew the mounting clips after installation was finished. This procedure was utilized to ensure that the participant always started subsequent trials (with/without an OSE) from an identical initial situation. In addition, the data obtained when unscrewing the mounting clips could be further included in the analysis, as the task performance, i.e. arm movements, was similar when screwing the clips.

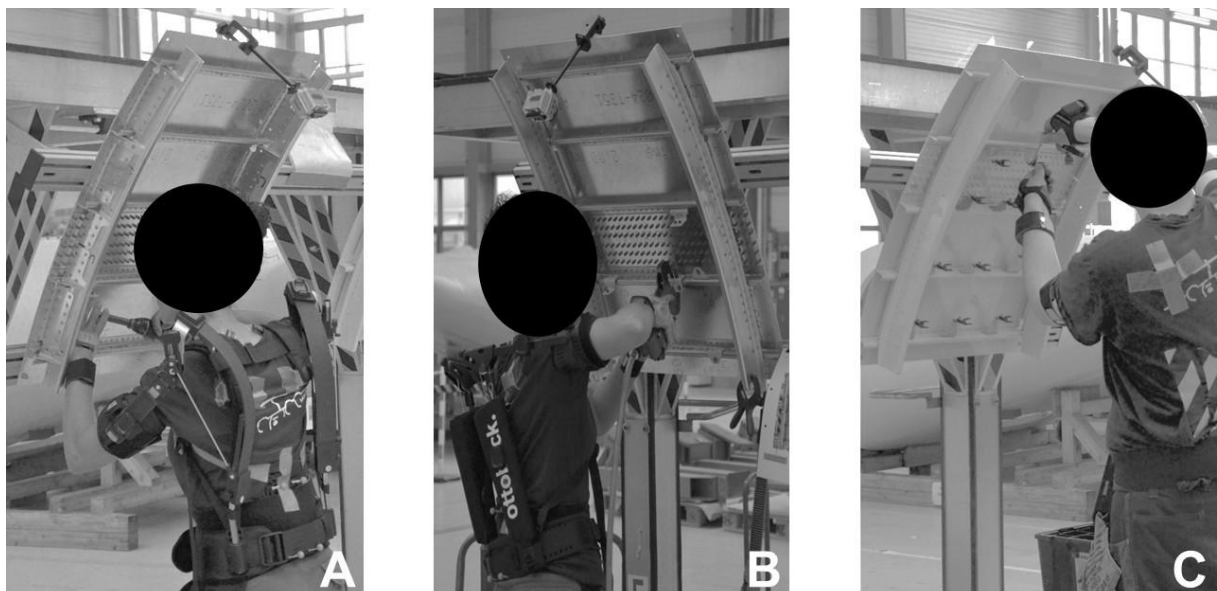


Figure 37: Two exemplary participants performing the drilling (A), riveting (B) and screwing task (C) with (A & B) and without (C) an occupational shoulder exoskeleton.

11.3. Measurement instrumentation and data acquisition

First, the respective assembly task was described to the participants. Then, they were positioned in front of their respective panel at the workstation replica. The position of the platform was then moved until the participants confirmed that they could work on the defined stringers of the panel without leaving the measurement area. Furthermore, they were instructed to perform their respective assembly task with the same level of accuracy and precision as when working on the aircraft fuselage. They were given no further restrictions on arm movements or task order. The participants were then instructed to set up their workstation in front of their panel by placing the tools required to complete the task on the provided workbench.

Once these preparations were completed, the height and weight of the participants were obtained according to the instructions provided by XSens (Movella Holdings Inc., Henderson, Nevada). Then, the three OSE (Paexo Shoulder, Skelex 360°, and Levitate Airframe) were adjusted to the participant's anthropometry according to the manufacturer's instructions. The support moments for the Skelex and Paexo were set based on their subjective feedback. Therefore, the participants were asked to report whether they felt a clear level of support from the exoskeleton while ensuring that their arms still naturally fell due to gravity. This was done to prevent full compensation of the shoulder joint moment. The same procedure was applied for the Airframe, but participants could only choose between support levels 3 or 4.

To measure upper body kinematics during the assembly tasks, participants were equipped with the XSens AWINDA system (60 Hz, Upper Body Set-Up, Movella Holdings Inc., Henderson, Nevada) consisting of eleven inertial measurement units (IMU) placed bilaterally on the hand, forearm, upper arm and scapula, and unilaterally on the head, sternum and pelvis (Figure 38). To quantify the effectiveness of the OSE in reducing shoulder muscle demand, electromyographic activity was recorded during the work tasks using the Cometa Wave Plus EMG system (2000 Hz, Cometa Systems Inc., Bareggio, Italy). For this purpose, electrodes (Kendall H124SG EKG, adhesive electrodes, 30 x 24 mm, Cardinal Health Germany 507 GmbH, Norderstedt, Germany) were placed bilaterally (left L; right R) over the following muscles (Figure 38) according to a standardized procedure (Boettcher et al., 2008): musculus deltoideus pars clavicularis (anterior deltoideus, AD), musculus deltoideus pars acromialis (medial deltoideus, MD), musculus trapezius pars descendens (upper trapezius, UT) and musculus infraspinatus (INF).

Before applying the electrodes, the relevant parts of the skin were shaved, pretreated with abrasives and cleaned with alcohol. Data collection was performed using the software XSens MVN Record (version 2022.0.2, Movella Holdings Inc., Henderson, Nevada) to ensure synchronization between the two systems.

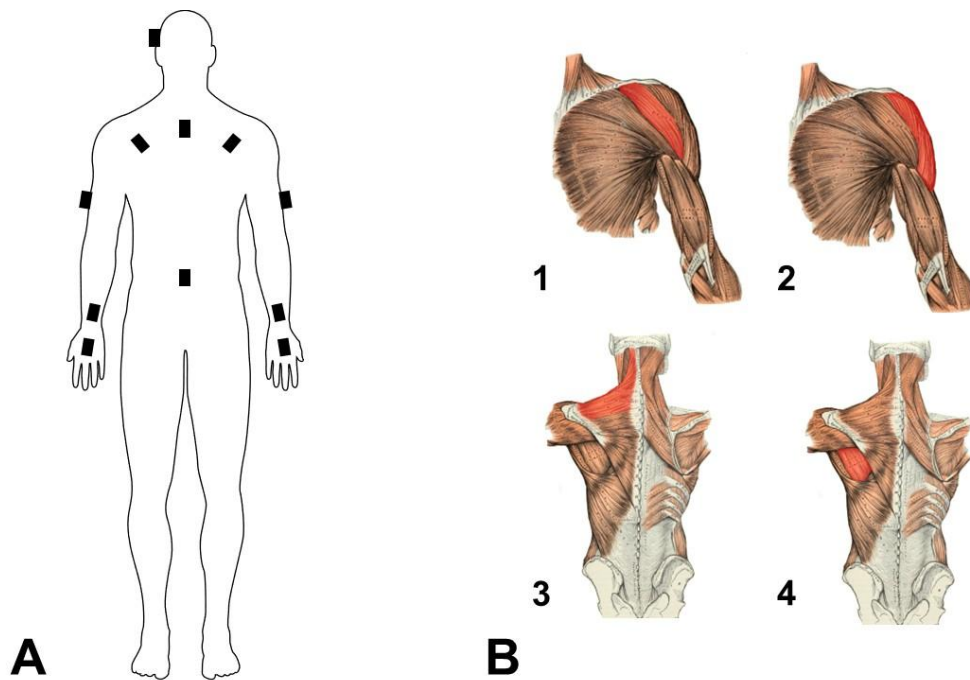


Figure 38: Illustration of the sensor placement of the XSens motion capture suit on a participant (black squares; A) and the muscle analyzed in this study (B): the anterior deltoideus (1), the medial deltoideus (2), the upper trapezius (3), and the infraspinatus (4). All graphics in B were extracted and modified from Gray and Lewis (1918; p. 433 & p. 436). The graphic of the human body silhouette (male; A) was modified from Openclipart (n.d.). Public domain. Changes were made by the author.

After completing the measurement preparations, the participants conducted maximal voluntary contraction (MVC) tests to obtain comprehensive reference values for each analyzed muscle. The number of MVC tests were kept to a minimum in order to reduce muscle fatigue prior to the work tasks. In addition, as the working height and therefore the shoulder angles were not standardized (as in the previous studies), an MVC protocol consisting of two isometric tests was derived according to Boettcher et al. (2008). They proposed a four-test protocol to calculate MVC values for twelve neck and shoulder muscles, including the four muscles analyzed in this study.

Therefore, two of the four MVC tests were included in the normalization protocol: the "empty can" test, to obtain reference values for UT, AD and MD simultaneously, and the "125° flexion" test, to obtain reference values for INF. The "empty can" test was performed with the shoulder abducted at 90° in the scapular plane, the arm internally rotated (palm facing backwards) and the elbow fully extended (Figure 39 A). To perform the MVC test in this posture, participants were instructed to "raise the arm" (increase abduction) while resistance was applied to the wrist by the experimenter. In addition, participants were told to avoid "lifting the shoulder" (i.e. the center of the GHJ) during the test, as well as "leaning to the other side", in order to minimize the influence of the lateral trunk muscles. For the second MVC test ("125° flexion"), the participant's arm was positioned in a shoulder flexion of approximately 125° and the participants were instructed to "rotate the shoulder outwards" (lateral rotation of the scapula) while the experimenter applied resistance above the elbow and at the inferior angle of the scapula (Figure 39 B).

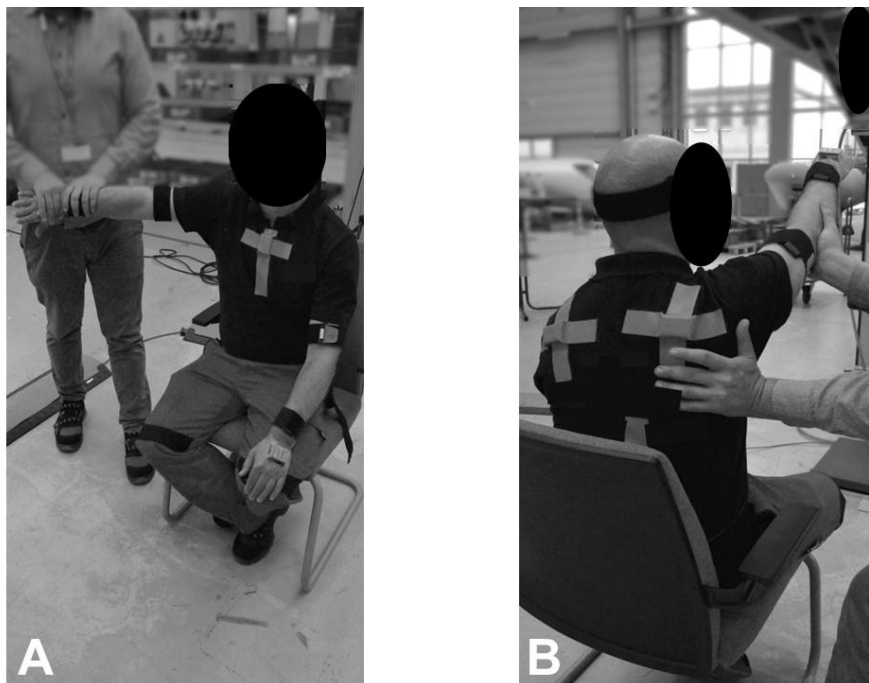


Figure 39: Exemplary participant performing maximal voluntary contraction (MVC) trials for the right arm in the two specific test postures "empty can" (A) and "125° flexion" (B).

Before performing an MVC test, participants were seated in a chair and asked to find a comfortable, upright and straight position without leaning against the back of the chair (upright position without back support).

Each MVC test began with a ten-second interval in which their arms were resting at their sides to collect zero values for muscle activity. Then they had to raise the arm to the specified height while the experimenter checked the correct arm posture. Participants were then instructed to slowly build up the force against the pressure applied by the experimenter to a maximum time of three seconds, and then to hold this contraction for a further three seconds. Finally, the force was slowly released and the arm was lowered back. This procedure was repeated three times to measure three maximum contractions in each of the two arm posture ("empty can" or "125° flexion") for each arm, separately. The reference value for each muscle was then determined as the maximum activity obtained from the three contractions. Between each repetition, the arm was rested on the participant's thigh for 30 seconds. The same MVC test was then performed on the contralateral arm, with the order of the MVC tests counterbalanced. Thus, half of the participants performed both "empty can" tests first before proceeding to the "flexion 125°" trials (and vice versa). Three minutes of rest were provided between each test, and participants received non-threatening verbal encouragement during the MVC.

After completing the MVC tests, participants performed their respective assembly task without or with one of the OSEs (factor EXO). The order for the EXO factor was randomized using a 4x4 Latin square prior to the study with an estimated study population of 20 participants, i.e. every fourth participant would perform their assembly task without an OSE. However, one employee was unable to take part in the study due to illness on the day of the measurement. Due to the short and limited study period, no replacement participant could be found. Each measurement trial consisted of the following steps: the participants remained in front of their respective panel for five seconds (to collect zero values for the EMG data and an initial pose for the XSens system) and then continued with their respective assembly task as described above. Participants were allowed at least ten minutes of rest before repeating the assembly task for the next EXO condition. Between each trial, the XSens system was recalibrated to ensure that the next measurement was not influenced by any sensor movements that might have occurred during the donning or doffing of an OSE. To identify specific tasks (e.g. drilling or screwing) in the data during a trial, two video cameras filmed the measurement (from the side and from the back).

11.4. Data processing and dependent variables

The XSens data were HD-processed for each measurement in the software “XSens MVN Record”. The XSens and EMG data were then synchronized with the corresponding video recordings in the “WIDAAN” software (version 2023.4, Institute for Occupational Safety and Health of the German Social Accident Insurance, Sankt Augustin, Germany; Jäger, 2018). The tasks performed were then identified and each measurement was divided into different time intervals, each containing the data for the specific task (Figure 40). For further biomechanical data analysis, the focus was on three specific tasks: drilling and riveting at the two stringer heights for group A (lower and upper stringer height; Figure 40) and screwing the mounting clips at the four stringer heights for group B.

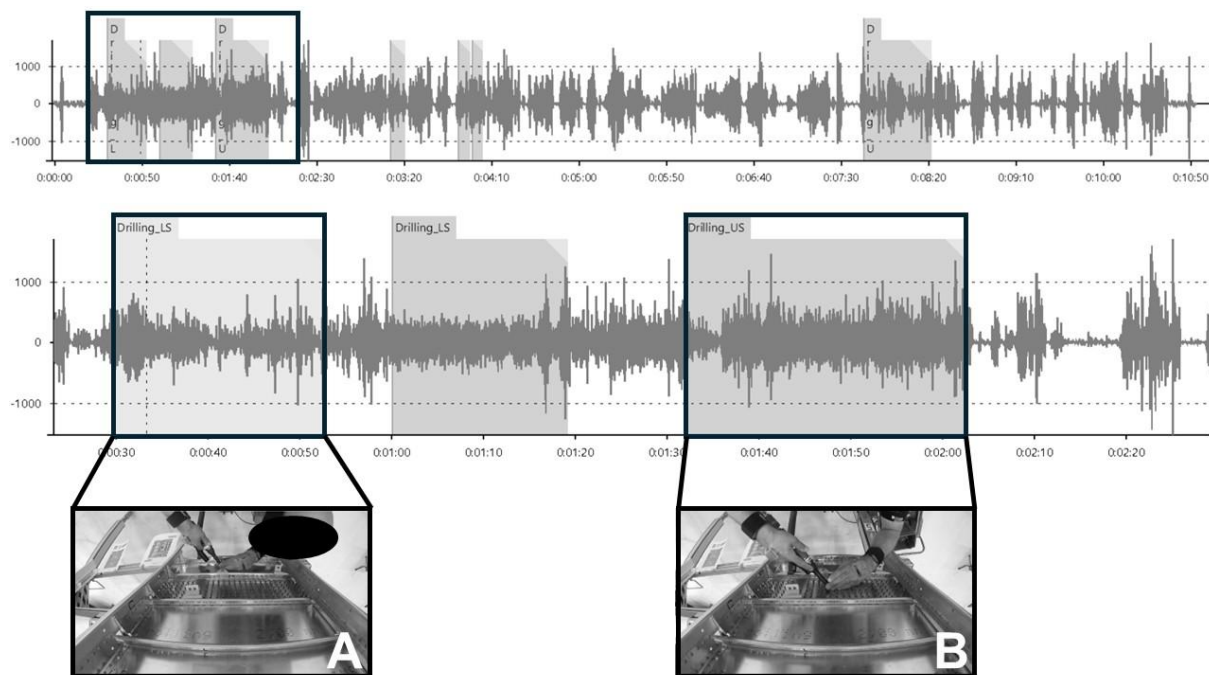


Figure 40: Muscle activity of an exemplary participant during a measurement. Two-time sections when performing the drilling task at the lower (A, Drilling_LS) and at the upper stringer height (B, Drilling_US), respectively, were highlighted (black squares) and the respective snippet from the synchronized video was included.

The shoulder joint moment was calculated for both arms from the kinematic data in WIDAAN based on the implemented biomechanical model "Der Dortmunder" (Jäger, 2018).

For the two tasks drilling and riveting additional weights were added to the participants' hands, each representing the weight of the specific working tool, in order to correctly determine the shoulder joint moment with the external loads. As all participants worked with the drill in their right hand, the weight of the drill (0.71 kg) was added to the right hand. However, the weight of the riveting machine (2.45 kg) was divided equally between the two hands, because this task was performed with both hands. As the weight of the tools used for screwing the mounting clips (group B) was significantly lower (<0.2 kg), these tool weights were not further included in the analysis.

Further data processing was performed in Python (version 3.12., Python Software Foundation, Wilmington, USA) using the "Kinetics Toolkit" (Chénier, 2021). The EMG data was offset corrected by subtracting the mean value calculated over the first five seconds of a measurement (while the participant remained in the neutral position). The EMG signals were then rectified and the root mean square value (window length 100 milliseconds) was computed. The EMG data for each muscle was then normalized with respect to the corresponding maximum reference value obtained during the MVC tests.

In order to estimate the support moment provided by each of the OSEs, the polynomial equations calculated in the previous section were applied separately for each participant, based on their recorded level of support, with the shoulder elevation angle (calculated from WIDAAN) as the input variable. In addition, the support moment for the Airframe was set to zero when an elevation angle greater than 128° was reached, as this was close to the maximum support angle (McFarland et al., 2022). Further increases in shoulder elevation would cause the arm to leave the arm cuff because they were not attached to each other (compared to the other exoskeletons). However, the equations for the support moments (derived from the laboratory study experiment) are not mathematically constrained with respect to the angle of elevation as an input variable and would therefore predict an incorrect support moment. The net shoulder joint moment (NSM) while wearing an OSE was then obtained by subtracting the support moment provided by the exoskeleton τ^{exo} from the shoulder joint moment ($\tau^{shoulder}$) calculated from the biomechanical model in WIDAAN as follows:

$$NSM = \tau^{shoulder} - \tau^{exo} \quad (11.1)$$

For the trials without OSE, the NSM was equal to $\tau^{shoulder}$.

For further statistical analysis of a trial, the mean value over all intervals (exemplary in 11.2 for the RAD to obtain mRAD) was calculated separately for the three working tasks (drilling, riveting and screwing) at the specific stringer heights (e.g. drilling at the lower stringer *DrillingLS*):

$$\text{mRAD}^{\text{DrillingLS}} = \frac{\sum_0^{n_0} \text{RAD}_0^{\text{rms}} + \dots + \sum_0^{n_k} \text{RAD}_0^{\text{rms}}}{n_0 + \dots + n_k} \quad (11.2)$$

with $n_0, \dots, n_k$ being the number of data points in the respective interval and k the number of intervals. This procedure was performed for both arms (L/R) on the processed EMG data of UT, AD, MD and INF and net shoulder joint moment (NSM) to obtain the respective variables: mLNSM, mRNSM, mLAD, mRAD, mLMD, mRMD, mLUT, mRUT, mLINF and mRINF.

11.5. Statistics

Statistical analysis was performed using SPSS (version 29.0.2.0 (20), IBM Corporation, Armonk, NY). The dependent variables included in the analysis were the obtained mean values for NSM, AD, MD, UT, and INF for both sides (L/R): mLNSM, mRNSM, mLAD, mRAD, mLMD, mRMD, mLUT, mRUT, mLINF and mRINF. They served as indicators of changes in the net shoulder joint moment and the muscle activity for the anterior and medial deltoideus, upper trapezius, and infraspinatus, respectively. The mean values obtained for the different working heights of a task for one participant were treated as separate measurements to increase statistical power. Friedman tests were then performed separately on the dependent variables for each of the three work tasks to determine the main effects of the independent variable EXO. The statistical significance level was set to 5% and post-hoc pairwise comparisons were performed with Benjamini-Hochberg correction (false discovery rate 5%). In addition, when significant differences were found using the Friedman test, further pairwise comparisons were only performed between the baseline condition (without OSE) and the three OSE conditions to limit the amount of post-hoc tests.

11.6. Results

Due to measurement errors, the data of two participants from group A were excluded. In addition, two participants (one from each group) were identified as outliers and were not included in the statistical analysis.

In group A, the Friedman test showed significant differences between conditions for the drilling task bilateral for AD, MD, UT and NSM (all p-values < 0.05; Figure 41). Pairwise comparisons revealed a significant reduction in mLAD (left) and mRAD (right) in all three exoskeleton conditions compared to the baseline condition with 3.4 %MVC (left) and 4.4 %MVC (right) for the Paexo, 2.7 %MVC (left) and 2.8 %MVC (right) for the Skelex and 4.4 %MVC (left) and 4.5 %MVC (right) for the Airframe (all p-values < 0.05; Figure 41). Additionally, mLNSM and mRNSM showed a significant reduction for the Paexo, Skelex and Airframe compared to the baseline with reductions of 3.3 Nm, 4.0 Nm and 3.6 Nm for the left side, and 3.4 Nm, 3.9 Nm and 3.8 Nm for the right side, respectively (all p-values < 0.05; Figure 41). A significant reduction compared to the baseline condition was found for mLMD and mRMD, but only for the two OSEs Paexo (left: 0.1 %MVC, right: 1.7 %MVC) and Airframe (left: 0.6 %MVC, right: 3.6 %MVC; all p-values < 0.05; Figure 41). Furthermore, mLUT and mRUT showed a significant reduction between the baseline and the Airframe with 3.4 %MVC and 2.6 %MVC, respectively (all p-values < 0.05; Figure 41). No significant differences between conditions were found for mLINF and mRINF in the drilling task as revealed by the Friedman test (both p-values > 0.05).

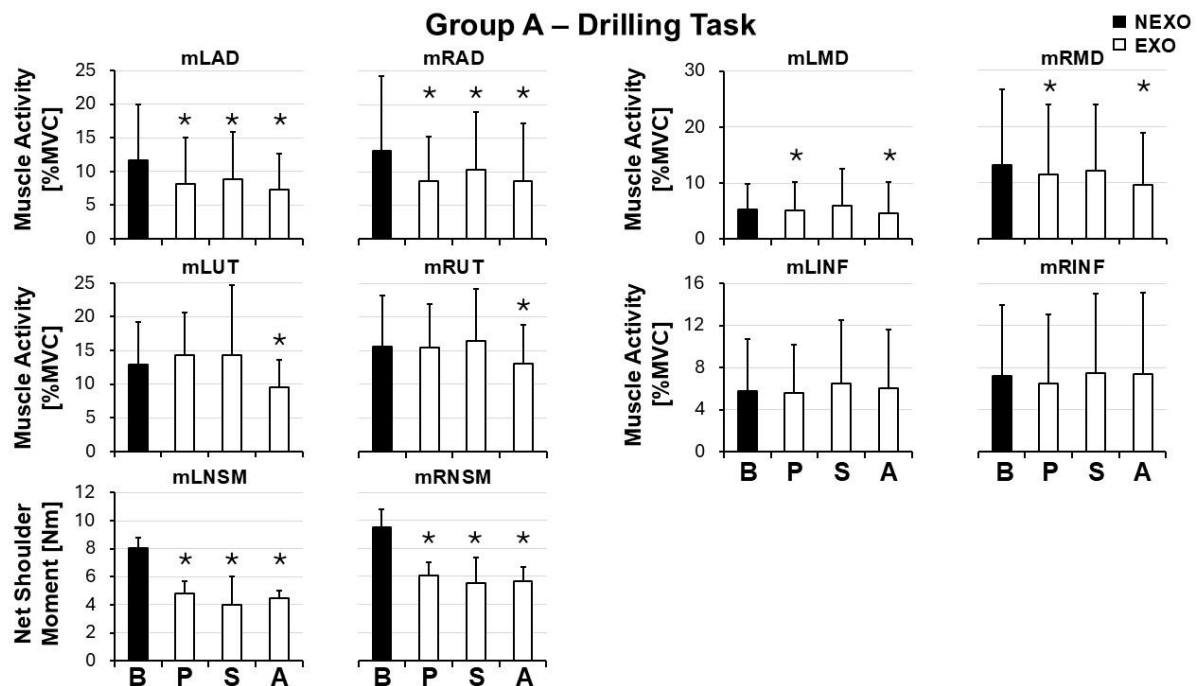


Figure 41: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex; A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton.

In group A, the Friedman test showed significant differences between conditions for the riveting task bilateral for AD, MD, UT and NSM (all p-values < 0.05; Figure 42). Pairwise comparisons revealed a significant reduction in mRAD and mRMD in all three OSE conditions compared to the baseline condition with 3.4 %MVC (mRAD) and 2.3 %MVC (mRMD) for the Paexo, 2.2 %MVC (mRAD) and 2.6 %MVC (mRMD) for the Skelex and 4.0 %MVC (mRAD) and 4.5 %MVC (mRMD) for the Airframe (all p-values < 0.05; Figure 42). Additionally, mLNSM and mRNSM showed a significant reduction for the Paexo, Skelex and Airframe to the baseline with 2.9 Nm, 2.9 Nm and 3.0 Nm for the left side and 3.4 Nm, 3.7 Nm and 3.7 Nm for the right side, respectively (all p-values < 0.05; Figure 42). A significant reduction compared to the baseline condition was found for mLAD, but only for the two OSEs Paexo (3.3 %MVC) and Airframe (5.7 %MVC; both p-values < 0.05; Figure 42). Furthermore, mLUT, mRUT and mLMD showed a significant reduction between the baseline and the Airframe with 2.9 %MVC, 3.3 %MVC and 1.3 %MVC, respectively (all p-values < 0.05; Figure 42). No significant differences between conditions were found for mLINF and mRINF in the riveting task as revealed by the Friedman test (both p-values > 0.05).

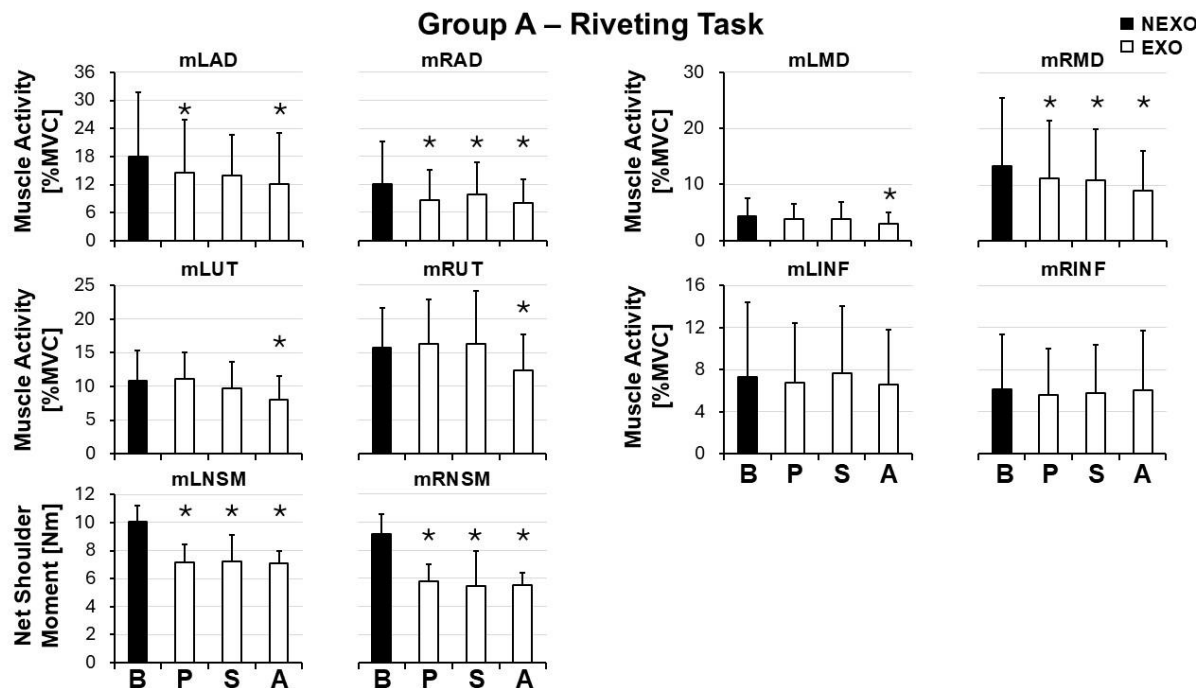


Figure 42: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex; A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton.

In group B, the Friedman test showed significant differences between conditions for the screwing task bilateral in AD, MD, UT, INF and NSM (all p-values < 0.05; Figure 43). Pairwise comparisons revealed a significant reduction in mLAD (left) and mRAD (right) in all three OSE conditions compared to the baseline condition with 2.1 %MVC (left) and 4.0 %MVC (right), 2.7 %MVC (left) and 4.0 %MVC (right) and 3.3 %MVC (left) and 3.6 %MVC (right) for the Paexo, the Skelex and the Airframe, respectively (all p-values < 0.05; Figure 43). Additionally, mLNSM and mRNSM showed a significant reduction for the Paexo, Skelex and Airframe compared to the baseline with 3.3 Nm, 6.8 Nm and 4.7 Nm for the left side and 3.4 Nm, 7.1 Nm and 5.0 Nm for the right side, respectively (all p-values < 0.05; Figure 43). mLMD revealed a significant reduction for all three OSEs compared to the baseline with 1.3 %MVC for the Paexo, 1.7 %MVC for the Skelex and 1.4 %MVC for the Airframe (all p-values < 0.05; Figure 43). Furthermore, a significant reduction to the baseline condition was found for mLUT, mRUT, mRMD and mLINF, but only for the two OSE conditions Airframe and Skelex, with 1.1 %MVC and 2.2 %MVC for the mLUT, 2.4 %MVC and 4.3 %MVC for the mRUT, 3.3 %MVC and 2.6 %MVC for the mRMD and 0.8 %MVC and 0.7 %MVC for the mLINF (all p-values < 0.05; Figure 43). mRINF showed a significant reduction for the Paexo compared to the baseline with 0.6 %MVC (p-value = 0.04; Figure 43).

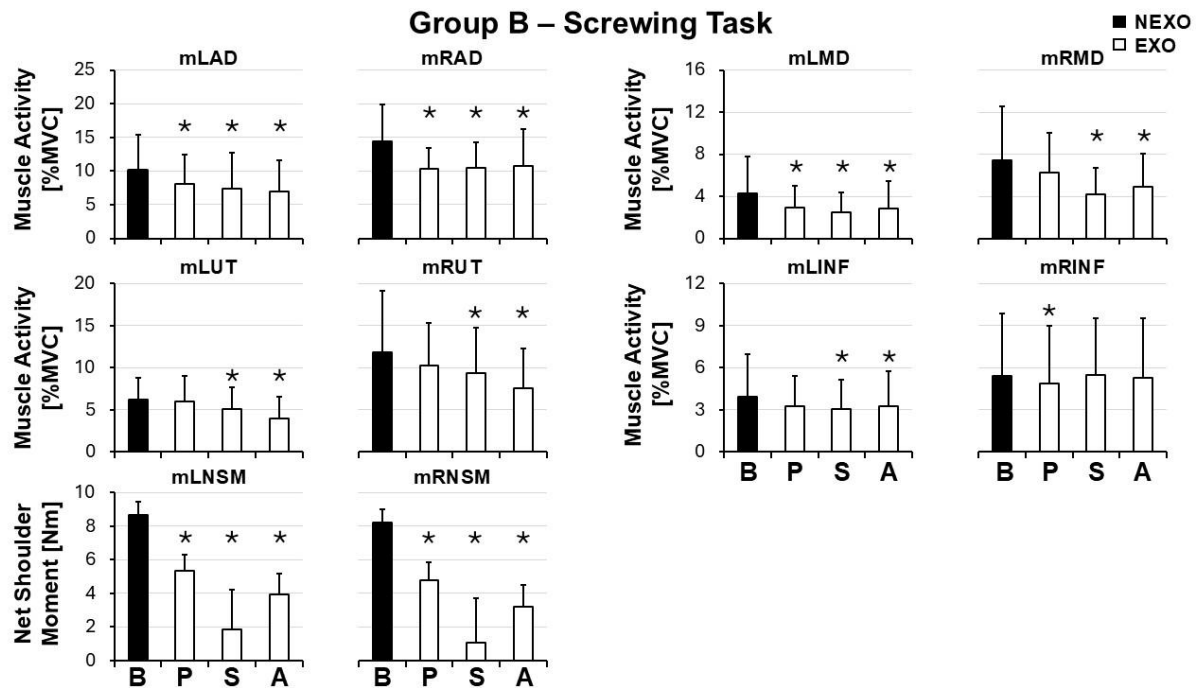


Figure 43: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex; A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton.

11.7. Discussion and limitations

The aim of this study was to provide an exploratory analysis of the influence of three occupational shoulder exoskeletons (Airframe, Paexo and Skelex) on the mechanical load at the glenohumeral joint (GHJ) and shoulder muscle demand during three different (non-standardized) aircraft manufacturing tasks: drilling, riveting and screwing. Therefore, the left and right net shoulder joint moment (LNSM and RNSM) and the muscle activity for the left and right anterior deltoideus (LAD/RAD), medial deltoideus (LMD/RMD), upper trapezius (LUT/RUT) and infraspinatus (LINF/RINF) were measured with and without the OSEs as indicators for the changes in the mechanical load at the GHJ and shoulder muscle demand.

While wearing any of the OSEs, the net shoulder joint moment was reduced for both arms across all three manufacturing tasks due to the provided support, with reductions ranging between 28.4% and 86.7%. This effect indicates a decrease in the joint compression forces and, therefore, a reduction of the mechanical load at the GHJ, assuming a linear relationship between net shoulder joint moment and compression forces (Praagman et al., 2000). These findings are consistent with those of van der Have et al. (2022) who reported reduced shoulder joint moments and shoulder contact forces during an overhead wiring task with a prototype passive shoulder exoskeleton indicating reduced musculoskeletal shoulder demand (van der Have et al., 2022). This assumption is further supported by the fact that for most exoskeleton conditions, a bilateral reduction in muscle activity during the tasks was shown for the anterior and medial deltoideus being the main agonists for arm elevation. In addition, activity in the upper trapezius muscle, which acts as an elevator and stabilizer of the shoulder girdle (Gohlke, 2002; Peat, 1986), partially decreased bilaterally when wearing an OSE, although these results were not consistent for all exoskeletons. The infraspinatus, which is part of the rotator cuff and responsible for stabilizing the humeral head in the shoulder joint during arm elevation (Peat, 1986), showed only a small (and debatable) reduction.

Similar findings have been reported in previous studies analyzing the effects of OSEs in laboratory studies during simulated shoulder and/or overhead industrial tasks such as riveting (Jorgensen et al., 2022a), sealing (Jorgensen et al., 2022b), drilling (Alabdulkarim & Nussbaum, 2019a; Alabdulkarim et al., 2019b; de Bock et al., 2022; Kim et al., 2018; Kim & Nussbaum, 2019; Kong et al., 2023; Schmalz et al., 2019), screwing and tracing with a drill (van Engelhoven et al., 2019),

screwing with heavier tools (Brunner et al., 2023; Pinho & Forner-Cordero, 2022), screwing by hands or light tools (Brunner et al., 2023; Musso et al. 2024, Schmalz et al., 2019), wiring (de Bock et al., 2022; Kim et al., 2018; Kim & Nussbaum, 2019; van der Have et al., 2022) and mounting with a nutrunner (Ojelade et al., 2023). Furthermore, studies conducted either in the workplace (field studies) or under real working conditions have also demonstrated the beneficial effects of OSEs across various sectors. These include the automotive industry (Gillette et al., 2022; Iranzo et al., 2020), a gas company (Pentenga et al., 2024), multiple construction sites (Mänttari et al., 2024), a slaughterhouse (Dalbøge et al., 2024) and different tasks such as plastering (de Vries et al., 2021), cleaning surgical instruments (Arnoux et al., 2023), manual material handling (Schrøder Jakobsen et al., 2023) or wool textile processing (Coccia et al., 2024).

11.7.1. Influence of the OSEs during the drilling task

For the drilling task, wearing an OSE resulted in a bilateral average reduction in net shoulder joint moment of 45.1% (left) and 39.3% (right) as well as a reduction in anterior deltoideus muscle activity of 30.1% (left) and 29.6% (right). For the upper trapezius (UT), a bilateral reduction in muscle activity was shown when wearing the Airframe with 26.4% and 16.7% for the left and right side, respectively. Based on these results, it can be assumed that the mechanical load at the GHJ and the muscle demand in the anterior deltoideus (as a shoulder elevator) and partially the upper trapezius (as a shoulder girdle stabilizer; Peat, 1986) decreased in both shoulders. These findings for anterior deltoid and/or upper trapezius muscle activity have also been reported in previous laboratory studies for passive OSEs while performing comparable drilling or screwing tasks overhead or at shoulder height (Alabdulkarim & Nussbaum, 2019a; Alabdulkarim et al., 2019b; Brunner et al., 2023; de Bock et al., 2022; Kim et al., 2018; Kim & Nussbaum, 2019; Schmalz et al., 2019; van Engelhoven et al., 2019). Van Engelhoven et al. (2019) also found an overall bilateral reduction in AD and UT activity when working overhead with a drill with the ShoulderX exoskeleton at different support levels. However, they reported an even greater reduction of up to 18 %MVC for AD (81% compared to the unassisted condition) and 10 %MVC for UT (46% compared to the unassisted condition) exceeding the differences in %MVC found in this study (van Engelhoven et al., 2019).

Similar, Brunner et al. (2023) found a higher reduction in the activity for both muscles (unilateral, dominant side) when screwing with a power tool (1.8 kg) while wearing the OSE DeltaSuit by 13.2 %MVC (AD) and 9.7 %MVC (UT). Additional studies also showed higher levels of reduction in AD and/or UT muscle activity when drilling or screwing with a heavier tool either above or at shoulder height, e.g. bilateral for AD between 26%–59% (Kim & Nussbaum, 2019) or unilateral around 30% and 40% for UT and AD, respectively (derived from the figures by Schmalz et al., 2019). Though, in contrast to these studies, de Bock et al. (2022) reported similar bilateral reductions in anterior deltoideus muscle activity with 2.8 %MVC during an overhead drilling task with an OSE, but no influence on upper trapezius activity. One possible assumption is that the support moment provided by the exoskeletons was higher in the studies by Brunner et al. (2023), van Engelhoven et al. (2019), Kim and Nussbaum (2019) and Schmalz et al. (2019). This may have contributed to a greater reduction in muscle activity and, consequently, a more pronounced decrease of deltoideus and trapezius muscle demand compared to the results of de Bock et al. (2022) and the present study. This assumption is supported by the fact that Brunner et al. (2023; 6.6 Nm peak moment), van Engelhoven et al. (2019; support moment ranging between 5 - 15 Nm) and Schmalz et al. (2019; ~70% of shoulder moment) all reported higher support from the exoskeletons, whereas the maximum support moment for de Bock et al. (2022) was between ~2 - 3.5 Nm and therefore similar to the values reported in this study (no information on the support moment was provided by Kim & Nussbaum, 2019).

In addition, the direction of the force applied to perform the specific task may also have influenced the effect of the OSEs (Ojelade et al., 2023). Ojelade et al. (2023) found differences in the magnitude of the effect of three OSEs in reducing AD and UT muscle activity while performing a simulated mounting task with a nutrunner between two different force directions; however, these differences were mostly not significant. Hence, for an upward force direction a lower reduction in muscle activity was observed (~2-17% for AD, ~10-27% for UT) compared to when a forward force was applied with the nutrunner (~24–46% in AD, ~26-53% in UT; Ojelade et al., 2023). Based on their results, Ojelade et al. (2023) concluded that the found differences occurred since the application of an upward force requires higher activity from the larger upper extremity muscle, e.g. the trapezius (Ojelade et al., 2023).

Contradictory, the muscle activity in the upper extremities should not vary between when a forward force is applied (Ojelade et al., 2023) compared to a downward force (Chopp et al., 2010a); therefore a similar influence of the OSE could be expected. However, since the drill head in this study had to be pushed downwards, opposite to the direction of the support moment, this may also have contributed to the reduced effect of the exoskeleton (Figure 37 A).

For the medial deltoideus (MD), the muscle activity in the drilling task was reduced by 6.3% on the average for the LMD and by 20.0% for RMD, but only for the Paexo and the Airframe. These findings imply a larger effect of these two exoskeletons for the right MD in decreasing its muscle demand. Additionally, it is debatable whether the small effects on LMD are even considered functionally relevant. Similar results have been reported for both exoskeletons in previous studies analyzing their effects on MD and UT for the dominant/right arm during drilling (Kong et al., 2023; Schmalz et al., 2019). Both Schmalz et al. (2019) and Kong et al. (2023) found reduced muscle activity in MD and UT while drilling at shoulder and/or overhead height when using the Paexo or the Airframe, respectively. Although, the reduction shown by Schmalz et al. (2019) was higher (up to 40% in both muscles), Kong et al. (2023) observed a similar decrease of 34.5% in MD and 21.9% in UT. Since the participants in the study of Kong et al. (2023) performed this drilling task at a fixed working height of 160 cm rather than a standardized height (as for Schmalz et al., 2019), the results of this study and those of Kong et al. (2023) are more comparable. However, both studies analyzed the effects of the OSE only unilaterally (for one arm) and not bilaterally while drilling with both hands.

In the study of Alabdulkarim and Nussbaum (2019a), bilateral peak muscle activity in the medial deltoideus (as well as unilateral in anterior) was found to be significantly lower when drilling with an OSE (ShoulderX) at overhead height. The findings of Alabdulkarim and Nussbaum (2019a) contrast with the results for MD in the present study, which showed conflicting effects of an OSE during the drilling task when comparing the left and right sides. The bilateral differences in the effect of the OSEs could be attributed to the task specific arm postures during the drilling task. All participants performed the task with the drill in the right hand and the drilling head pushed downward onto the rivet (Figure 37 A).

Since the angle between the drill and drill head was $\sim 90^\circ$ (in contrast to the study from Alabdulkarim & Nussbaum, 2019a), they had to rotate the forearm and elevate the arm into a position requiring both shoulder flexion and abduction performed by the anterior and medial deltoideus, respectively. Thus, the left hand was placed above the drill head to stabilize the drill and to apply the downward force. Hence, the left arm was elevated mainly in the sagittal plane, i.e. the arm elevation was primarily conducted by the anterior deltoideus. This assumption is supported by the fact that medial deltoideus activity in the baseline condition (i.e. without an OSE) was twice as high on the right side compared to the left. These findings are comparable to those of Kim et al. (2018), who found a reduction in shoulder muscle activity, respectively the anterior and medial deltoideus, only for the right side when performing a simulated drilling task at shoulder height with both hands while wearing an OSE (Ekso Vest).

Further, there was no effect when drilling with an OSE on infraspinatus muscle activity. In addition, some exoskeletons even showed an increase in infraspinatus activity (though not significant). This could indicate a detrimental effect of exoskeleton support on infraspinatus muscle demand, which could lead to impairment of glenohumeral joint stability (Ojelade et al., 2023). To maintain the humeral head on the glenoid fossa during arm elevation, the muscle forces of the humeral flexor and abductor muscles interact with the forces of the infraspinatus muscle (Peat, 1986). Therefore, a balanced reduction in shoulder muscle activity (both agonist and antagonist muscles) by the OSE may be required to reduce infraspinatus activity. Subsequently, it could be suggested that for the drilling task, the support provided by the exoskeletons may not have been sufficiently adapted to the demands required to reduce INF muscle activity for this specific task (van Engelhoven et al., 2019). This suggestion is further supported by the fact that participants identified their preferred support moment prior to the measurements and without any external load, such as the weight of the tool. Furthermore, the support moment of each exoskeleton was not adjusted between or during the manufacturing tasks. This is consistent with the findings of van Engelhoven et al. (2019), who measured a reduction in infraspinatus activity in only one of the three support settings of the shoulder exoskeleton (medium support 10 Nm), but not in the other two conditions (5 or 15 Nm). In addition, van Engelhoven et al. (2019) reported that this setting was the most preferred by the participants to perform the specific tasks.

Therefore, to achieve a beneficial effect of the OSE in reducing muscle activity of the infraspinatus muscle during drilling, the support moment of the shoulder exoskeleton should be adjusted according to the specific task requirements.

11.7.2. Influence of the OSEs during the riveting task

For the riveting task, wearing any of the OSE lead to a bilateral average reduction in the net shoulder joint moment by 28.8% and 39.3%, therefore decreasing the mechanical load at the GHJ (Praagman et al., 2000). However, in contrast to the right shoulder, the effect for the left shoulder was lower compared to the drilling task (45%) which can be explained by the increased external load from the riveting tool weight (~1.2 kg per arm). As for the drilling task, the reduction in mechanical load at the GHJ in the riveting task can be attributed to an expected reduction in shoulder muscle forces caused by the exoskeleton support. For the right shoulder, this assumption is supported since an average reduction in the muscle activity of RAD, RMD and RUT (the latter only in the Airframe) by 26.7%, 23.0% and 21.1% respectively, was found when performing the riveting task with an OSE.

A previous study by Jorgensen et al. (2022a) found similar effects for an identical horizontal aircraft riveting task in their laboratory study for three OSE (Paexo, Skelex and EVO), but only for both deltoideus muscles (anterior and medial) and not for the upper trapezius, when working at a higher stringer inside a similar aircraft panel. If the riveting task was performed at a lower stringer, i.e. at a working height comparable to this study, no significant reduction in the activity in all three muscles was found for any of the OSEs (Jorgensen et al., 2022a). These contradictory findings may be caused by different approaches regarding the adjustments of the support moments provided by the OSEs. In the study by Jorgensen et al. (2022a), the evaluation of the exoskeleton support moment prior to the measurements was performed with a 90° shoulder abduction and the elbow flexed at 90°. Therefore, the shoulder joint moment required to keep the arm elevated in this position should be lower compared to this study where the arm was fully extended (elbow flexion 0°) in the reference position. Thereby, the center of gravity (CoG) of the arm is shifted in distal direction along the arm increasing the distance between CoG and the GHJ. Due to the increased shoulder joint moment, the participants in this study might have preferred a higher support moment in contrast to the participants from Jorgensen et al. (2022a).

In addition, the riveting tool utilized by Jorgensen et al. (2022a) was slightly heavier (approximately 0.2 kg) contributing to a higher external load which would further decreasing the exoskeleton effect. Furthermore, an average reduction in muscle activity of 25.2% and 26.6% during the riveting task was found for LAD (not for the Skelex) and LUT (only for the Airframe), respectively.

In contrast to the RMD, a reduced activity in LMD was measured primarily for the Airframe with 30.6%, while the reduction for Paexo and Skelex did not reach the significant threshold (both differences below 1 %MVC). It can be concluded that the reduction in muscle demand through the OSEs during the riveting task may be more pronounced for the right medial deltoideus compared to the left. As previously indicated for the drilling task and consistent with the findings of Jorgensen et al. (2022a), this variation in the OSE effect might be attributed to differences in arm posture when working with the riveting tool (Jorgensen et al., 2022a). The participants lifted the head of the riveting tool with their left hand up to the designated working height while positioning the tool with their right hand above the rivet (identical to Jorgensen et al., 2022a; Figure 37 B). Since the left arm has to support the weight of the tool, as indicated by the higher muscle activity in LAD (17.9 %MVC) compared to RAD (12.0 %MVC) when working without an OSE, participants might have kept the left arm close to the body to reduce the shoulder joint moment in the frontal plane. Subsequently, the arm elevation on the left side was conducted by the anterior deltoideus as muscle activity was higher in LAD (17.9 %MVC) in relation to LMD (4.3 %MVC). Therefore, a possible conclusion is that the support moment of the OSEs has a main effect on LAD muscle forces. However, for the right side, it can be assumed that the muscle forces for the anterior and medial deltoideus evenly contributed to the arm elevation, since the mean levels of muscle activity during the measurement without an exoskeleton were nearly identical (RAD 12.0 %MVC vs. RMD 13.4 %MVC). As a result, the exoskeleton should have reduced the muscle forces both for RAD and RMD. Similar to the drilling task, there was no effect of the OSEs on the activity of the infraspinatus muscle during the riveting task, and therefore on its muscle demand, which could also indicate that the support moment of the OSEs not sufficiently adapted to the task-specific requirements.

11.7.3. Influence of the OSEs during the screwing task

During the screwing task, the net shoulder joint moment was bilaterally reduced by the support of the OSEs by 57.0% and 63.0% on the average for the left and right shoulder joint, respectively. Furthermore, muscle activity for both sides was lower on average when working with one of the exoskeletons in the anterior deltoideus by 26.5% (left) and 27.9% (right), the medial deltoideus by 35.1% (left) and 39.2% (right, except for the Paexo) and the upper trapezius (except for the Paexo) by 26.3% (left) and 28.5% (right). Due to the small differences between left and right side in the reduction of muscle activities and shoulder joint moments, a symmetrical effect of the OSEs in reducing mechanical load at the GHJ and shoulder muscle demand for the screwing task can be assumed. This symmetrical effect might have resulted from an even bilateral support moment provided by the exoskeletons.

These results are consistent with the findings from previous studies by Kim and colleagues who reported a reduction in bilateral muscle activity for anterior and medial deltoideus and upper trapezius during a wiring task at shoulder and overhead when working with the EkSo Vest shoulder exoskeleton (Kim et al., 2018; Kim & Nussbaum, 2019). Jorgensen et al. (2022b) showed similar results in the effect for AD and MD muscle activity for three shoulder exoskeletons (Skelex, Paexo, EVO) for a horizontal sealing task (with low tool weight of 0.024 kg) at two different working heights. In contrast to the presented reduction in muscle activity for the screwing task, all these three studies found a higher reduction when working with the respective exoskeleton, partially up to 59.5% for LAD (Kim et al., 2018) or 11.5 %MVC and 5.5 %MVC for anterior and medial deltoideus, respectively (Jorgensen et al., 2022b). Schmalz et al. (2019) even reported a reduction of over 40% in RAD and RMD muscle activity and over 30% for RUT during an overhead screwing task for the Paexo. Similar, Brunner et al. (2023) found for all three muscles higher reductions in their activity for the dominant side when screwing with the DeltaSuit by 15.6 %MVC (AD), 9.8% (MD) and 10.1% (UT). In contrast, the bilateral reductions reported in the dominant (D) and non-dominant (ND) arm for the anterior (3.4 %MVC in D and 6.7 %MVC in ND) and medial deltoideus (2.1 %MVC in D and 1.8 %MVC in ND) by Musso et al. (2024) for the Skelex during an overhead screwing task were in accordance with the findings in this study. Analogous to the previous assumptions for the drilling task (chapter 11.7.1.), the larger effect on muscle activity from Brunner et al. (2023) and Schmalz et al. (2019) could have resulted from a higher support provided by the Paexo and the DeltaSuit.

However, for the study by Musso et al. (2024), a similar support moment for the Skelex can be assumed, as they applied the same procedure to estimate the individual support settings for a participant. Furthermore, this might indicate that the support moment setting applied in this study for the Paexo was too low to significantly reduce RMD or upper trapezius muscle activity during the screwing task. The Paexo showed the lowest reduction in shoulder joint moment was shown for this measurement (mean reduction 3.4 Nm) compared to the Airframe (mean reduction 5.0 Nm) and the Skelex (mean reduction 7.1 Nm). In addition, other studies measured no influence on upper trapezius activity in their overhead wiring tasks when a similar support moment (max. 3.5 Nm) from the exoskeleton was provided (de Bock et al., 2022; van der Have et al., 2022). However, in contrast to Schmalz et al. (2019) the support moment provided by the exoskeletons analyzed by Jorgensen et al. (2022b) should be lower compared to the exoskeletons utilized in the present study. The reason is that both studies from Jorgensen and colleagues selected the support moment based on the same procedure (i.e. in the same reference posture). Consequently, the reduction in the muscle activity for UT found in this study compared to Jorgensen et al. (2022b) could have resulted from the higher exoskeleton support. In contrast, the higher reduction in deltoideus muscle activity reported by Jorgensen et al. (2022b) for their sealing task through a higher exoskeleton support is rather unlikely (no information on the selection of the support moment was provided in both studies from Kim and colleagues). Subsequently, variations in the magnitude of the exoskeleton's effect on reducing deltoideus muscle activity may have resulted from the different approaches in defining the working heights in the respective studies. The working height in the present study was fixed and not adjusted to the individual anthropometry of the participants (non-standardized) in order to replicate the real workplace conditions more realistically. This approach is different to the previous laboratory studies where the working height was standardized between the participants (Jorgensen et al., 2022b; Kim et al., 2018; Kim & Nussbaum, 2019). Consequently, the difference in the working height increased the variations in arm postures (especially shoulder elevation angles) and likewise increased the variability in the support provided by the OSEs. This might cause a larger deviation in the individual effect of the OSEs for each participant thereby minimizing the overall difference in relation to working without the OSE.

In contrast to the results from the mentioned laboratory studies, Pentenga et al. (2024) found in their field test similar reductions in muscle activity of 2.0 %MVC for anterior deltoideus (mean reduction in this study of 3.3 %MVC) and of 2.1 %MVC for upper trapezius (mean reduction in this study of 2.5 %MVC) during the utilization of a shoulder exoskeleton when tightening nuts with the hands at a real workplace in a gas company. Therefore, the findings from Pentenga et al. (2024) are more comparable to the result of this study, additionally since both working heights were fixed and not standardized to the anthropometrics of the participants.

For the infraspinatus, a reduction in muscle activity was found during the screwing task of 21.6% (0.8 %MVC) and 17.1% (0.7 %MVC) for the left shoulder when working with the Skelex and Airframe, respectively, and of 10.2% (0.6 %MVC) in the right shoulder for the Paexo. Although the reductions found are relatively small, they could be beneficial in reducing the muscle demand on the infraspinatus. In the literature, previous studies have reported mixed results on the effect of shoulder exoskeletons on infraspinatus activity while performing industrial tasks above or at shoulder height, both beneficial (Kim & Nussbaum, 2019; van Engelhoven et al., 2019) and detrimental (Ojelade et al., 2023). Kim and Nussbaum (2019) found a beneficial effect of an OSE in bilaterally reducing INF muscle activity during their wiring task when working at overhead height. However, when working at shoulder height, this effect only occurred in the left infraspinatus, suggesting that its bilateral occurrence depends on the working height, i.e. arm elevation. This suggestion could further support why the reduction in INF muscle activity shown in this study for the Airframe and Skelex during the screwing task was found only for the left shoulder. Regarding the effects for the Paexo, it could be suggested that the support moment was too low to (significantly) reduce muscle activity in the left infraspinatus (similar to the upper trapezius). In contrast, the results for the right infraspinatus suggest that the support provided by the Paexo may be sufficient to reduce its muscle demand (van Engelhoven et al., 2019). Consequently, the support moment from the Airframe and Skelex may have been too high during the screwing task. Therefore, to increase the effect during the screwing task, the support moment of the OSE should be adjusted according to the specific task requirements.

11.7.4. Differences in the influence between the OSEs

While the aim of this study was to provide an overall exploratory analysis of the influence of occupational shoulder exoskeletons on the mechanical load at the GHJ and shoulder muscle demand during specific non-standardized aircraft manufacturing tasks, the statistical analysis focused on the differences between working with and without an OSE. Therefore, pairwise comparisons between the three exoskeletons were not performed in order to minimize the number of statistical tests. However, some conclusions can still be drawn about their individual effects on the mechanical load at the GHJ and shoulder muscle demand when comparing the individual results for each exoskeleton.

The Airframe was the only OSE that reduced muscle activity, i.e. muscle demand, in the upper trapezius muscle across all three manufacturing tasks. In contrast, the Paexo showed no effect on upper trapezius activity, while for the Skelex a reduction was found only during the screwing task. As previously described, the differences for the screwing task may be due to the variation in the exoskeleton support, as the reduction in net shoulder joint moment was lowest for the Paexo. However, the difference in the estimated reduction in net shoulder joint moment through the support provided by the Airframe compared to the other two OSEs was smaller for the drilling and riveting task (difference in mean net shoulder joint moment < 0.5 Nm) than for the screwing task (difference in mean net shoulder joint moment > 1 Nm). In the drilling task, an even greater reduction in net shoulder joint moment was measured for the Skelex compared to the Airframe. Consequently, the reduction in upper trapezius activity found for the Airframe in the drilling and riveting task cannot be fully contributed to the exoskeleton support moment, but may also have resulted from differences in the OSE design as well as different task demands (Ojelade et al., 2023). As indicated by the higher muscle activity without an OSE, the drilling (mean UT activity 14.3 %MVC) and riveting (mean UT activity 13.3 %MVC) task placed a higher muscle demand on the upper trapezius than the screwing task (mean UT activity 9.0 %MVC) due to the increased external shoulder load (from the higher tool weight) and higher precision requirements (Sporrong et al., 1998). To reduce the musculoskeletal load in the shoulder, the OSE redistributes the external load (from the weight of the arm with or without a tool) from the shoulder to the hips. Therefore, a proper fit with the respect to the hip coupling is required in order to efficiently transfer the gravitational forces caused by the external load via the exoskeleton structures to the hips (van Engelhoven et al., 2019).

If the hip coupling is too low, the load could be partially transferred via the shoulder straps to the neck and upper back, which could increase upper trapezius activity, as noted by van Engelhoven et al. (2019). As the hip belt of the Airframe is more rigid compared to the other two OSEs, deformations within the belt caused by the reactive forces should be reduced, further improving the connection of the exoskeleton to the hip. This may explain why the reduction in upper trapezius activity was only found when working with the Airframe during the drilling or riveting task.

No difference in the muscle activity was found bilaterally for the medial deltoideus when drilling with the Skelex, while at the same time the greatest reduction in net shoulder joint moment was estimated for this OSE. Alabdulkarim et al. (2019b) reported a reduction in deltoideus muscle activity when drilling at shoulder height with an OSE, but also an increase in drilling errors with high precision requirements. Consequently, the support provided by the Skelex may have been too high and therefore, due to the high precision requirements of the drilling task analyzed in this study, participants may have had to increase their muscle activity in the medial deltoideus to stabilize the arm position.

11.7.5. Summary

In summary, for all three manufacturing tasks, the OSEs bilaterally reduced the net shoulder joint moment by decreasing shoulder muscle forces, as evidenced by a reduction in arm elevator muscle activity. In addition, the OSEs showed potentially beneficial effects by reducing the activity of the shoulder stabilizer muscles. Thus, as concluded, all these effects are dependent on exoskeleton design, support moment, and task demands. For the drilling and riveting task, differences in posture between the left and right arms combined with increased external load and higher precision requirements resulted in bilateral differences in the exoskeleton's effects on the activity of the anterior and medial deltoideus muscles. In contrast, similar bilateral effects of the exoskeletons were found for the screwing task, evenly distributed between the anterior and medial deltoideus muscles.

These findings, particularly the reduction in muscle activity, are consistent with those reported in previous laboratory and field studies. However, the magnitude of the effects observed in the laboratory studies tended to be higher (e.g. up to 18 %MVC resp. 81% to the unassisted condition, van Engelhoven et al., 2019) than those found in this study (<1 - 5.7 %MVC resp. <10 – 43.9%).

While these higher reductions in muscle activity may be partly due to a potentially higher support moment provided by the respective exoskeleton (as concluded in the previous chapters), differences in study protocols such as variations of the workplace settings (e.g. working height) or the study population may also have contributed to the observed variations in the effects of the exoskeletons. In almost all of the laboratory studies, participants performed the specific tasks at an individual work height that was adjusted based on their anthropometry, resulting in standardized arm postures. In contrast, a fixed work height (as in this study and in Kong et al., 2023) allows participants to adopt individual arm postures. In addition, in most laboratory studies, participants were “untrained” workers with little or no experience with the work task (depending on whether familiarization with the task was provided). Increased work experience gained over time may lead to the adoption of a more ergonomically favorable work posture to reduce musculoskeletal load (Kadefors et al., 1976). Kadefors et al. (1976) found increased muscle fatigue in the anterior deltoideus and upper trapezius when performing overhead welding, but only in a group of inexperienced welders compared to experienced welders. Consequently, the described adjustments in arm posture may increase the variability of the individual effects of the exoskeletons, further minimizing the respective overall effect (Pacifico et al., 2022). As the manufacturing tasks analyzed in this study were performed at a fixed working height by professional aircraft manufacturers, similar influences of the variation in the arm postures on the exoskeleton's effects can be assumed. In addition, the study protocol was more comparable to a field study as the tasks were performed under real working conditions. Therefore, the results may be more consistent with the findings of those studies that have evaluated the influence of OSE on musculoskeletal load during daily work activities. Similar to the results of this study, Iranzo et al. (2020) and Gillette et al. (2022) both measured a comparable reduction in muscle activity for the anterior deltoideus of 34% and 3.2 %MVC, respectively, during overhead assembly tasks using the Airframe. Furthermore, reduced activity in the anterior and medial deltoideus (32.5% and 46.2%, respectively) was reported for various construction work groups, such as roofers or builders, when working with the Skelex (Mänttari et al., 2024). Overall, these findings are more in line with the results reported in this study regarding the magnitude of the exoskeleton's effects in reducing shoulder muscle demand.

11.7.6. Limitations

Several limitations of this study must be considered when interpreting the results. Firstly, the study aimed to provide a general analysis of the influence of the OSEs on the mechanical load at the GHJ and shoulder muscle demand for the three manufacturing tasks. Although beneficial effects were found for all three tasks, it remains unclear whether these effects vary between different working heights, particularly for the screwing task. However, previous studies have shown that the magnitude of the effects of OSEs in reducing muscle activity is highly dependent on working height, and is greater at overhead than at shoulder height (Jorgensen et al., 2022a; Kim & Nussbaum, 2019; Kong et al., 2023). When working below shoulder height, exoskeletons may not offer any beneficial effects (Pinho & Forner-Cordero, 2022).

Secondly, to quantify the effect of the OSEs in reducing shoulder joint moment, the net shoulder joint moment was determined by subtracting the support moment of each exoskeleton from the corresponding current shoulder joint moment. However, this approach is limited because it assumes that the GHJ axis of elevation and the exoskeleton axis of rotation are collinear during the tasks. Therefore, a misalignment between the two axes would lead to deviations in the estimated net shoulder joint moment due to variations in the exoskeleton support moment. To correct for these deviations, the position of both the GHJ centre and the exoskeleton joint centre must be estimated. To the best of the author's knowledge, no method currently exists to adequately measure the position of the GHJ centre when evaluating OSEs in the field at real workstations. Another approach to correct deviations in the net shoulder joint moment is to estimate the support provided by the OSE by measuring the contact forces between the upper arm and the arm cuff (as performed in chapter 8). However, to date and the best of the author's knowledge, no commercially available passive shoulder exoskeleton features integrated force sensors in the arm cuff. Furthermore, the installation of a force sensor in or on the arm cuff of an existing commercial exoskeleton, as performed by de Vries et al. (2019), has not yet been tested in the field. Therefore, the authors' approach was considered sufficiently robust to estimate the effect of an OSE on shoulder joint moment under real working conditions.

Thirdly, the support moment of each OSE was defined in a reference arm posture for all participants prior to the measurements and was not adjusted between or during the measurements. Also, the participants were not able to practice with the OSEs beforehand. To increase the benefits of the OSEs (and reduce potential adverse effects), the support moment should be optimized according to the specific task demands (e.g. bilateral differences in external load especially when tools are used single-handed) and the use should be practiced during a familiarization process (van Engelhoven et al., 2019).

Fourthly, the focus of the study was on muscles involved in either arm elevation (agonists) or shoulder stabilization. Therefore, the effect of the OSE on antagonists (such as the posterior deltoideus or triceps) or on trunk and back muscles (such as the erector spinae or latissimus dorsi) was not measured. Although, it is possible that their muscle demands may not have been negatively affected when working with the OSEs (Coccia et al., 2024; Gillette et al., 2022; Schröder Jakobsen et al., 2023; Jorgensen et al., 2022a; Jorgensen et al., 2022b). In addition to the beneficial effects on agonist muscle activity, OSEs may also have the potential to reduce antagonist or trunk and back muscles activity when performing industrial tasks (Arnoux et al., 2023; de Bock et al., 2022; Jorgensen et al., 2022a; Jorgensen et al., 2022b; Kim & Nussbaum, 2019; Kong et al., 2023; Ojelade et al., 2023; Schmalz et al., 2019). However, working with an OSE can also lead to an increase in these muscle groups (Dalbøge et al., 2024; van Engelhoven et al., 2019), especially if the support moment provided by the OSE is too high (van Engelhoven et al., 2019).

Fifthly, the weight of the riveting tool was divided equally between both arms, implying an even load distribution on both shoulder joints. This was a necessary constraint as the actual load distribution could not be measured. Although, the generally higher muscle activity for the left anterior deltoideus during the riveting task indicated that the left arm primarily carries the weight of the tool (Jorgensen et al., 2022a). Consequently, the shoulder joint moment, i.e. the mechanical load at the GHJ, should be higher for the left shoulder and lower for the right shoulder.

Finally, the work duration, defined as the time spent on working with and without OSE, was limited to between 5 and 20 minutes depending on the group. Therefore, the study provides only a brief overview of the influence of OSEs for these work tasks, with a beneficial effect on reducing mechanical load at the GHJ and partially shoulder muscle demand.

Even small (non-significant) reductions could further contribute to the reduction of muscle fatigue when the exoskeleton is used over a longer period of time. In addition, the focus was placed on three specific tasks, hence the influence of the OSE on other tasks (or side tasks) performed during aircraft manufacturing at the construction line is unclear.

11.8. Conclusion

In conclusion, the occupational shoulder exoskeletons have demonstrated to be an ergonomically viable intervention in order to reduce mechanical load at the glenohumeral joint and decrease muscle demand in the arm elevator and, to some extent, shoulder stabilizer muscles during several specific non-standardized aircraft manufacturing tasks performed at or around shoulder level by reducing the net shoulder joint moment and muscle activity, respectively. Furthermore, as the tasks analyzed in the work replica were almost identical to the real work tasks, similar beneficial effects can be expected when exoskeletons are used for these tasks at the aircraft fuselage. However, it is still unclear whether these short-term beneficial effects can lead to a reduction in shoulder musculoskeletal disorders at the real workplace. Therefore, the use of OSEs needs to be analyzed over a longer period of time (weeks to months), including working with the OSEs during full-time shifts. In addition, the exoskeleton support should be adapted to the specific task demands in order to increase its effects on the analyzed muscles. Finally, the analysis should also focus on the influence of the exoskeletons on muscle demand in antagonist and postural muscles, and also measure subjective feedback regarding discomfort and perceived exertion.

12. Discussion and limitations of the thesis

The overall aim of this thesis was to conduct exploratory analyze of the biomechanical effectiveness of occupational shoulder exoskeletons (OSEs) in reducing musculoskeletal load (MSL) in the shoulder during tasks performed at or above shoulder level. To this end, the analysis was structured into four separate studies focusing on the effect of OSEs on the mechanical load at the glenohumeral joint (GHJ), shoulder muscle demand and muscle tissue oxygenation (MTO), as these parameters are strong indicators of how MSL can be affected by OSE use.

The first study aimed to quantify the support moment generated by three different OSEs across their respective support range and for all available support levels (SL) during repetitive dynamic elevations under a standardized measurement protocol. The results showed that, for all three OSEs, increasing the support level resulted in a higher support moment achieved by an increase in the support force at the arm cuff (identical to Musso et al., 2022), while the external lever arm remained constant. Furthermore, all three OSEs demonstrated a clear support effect (greater than 1 Nm) within an elevation range of approximately 40° to 120°, extending up to 140° for certain support level configurations (Figure 44). The collected data was then used to model the support moment as a function of the exoskeleton angle using polynomial regression for all support levels. This enabled further estimation of the support moment provided by the OSE to the user in both laboratory and field studies, as well as the determination of the maximum peak moment and its corresponding angle. It was found that, depending on the OSE, peak support moments ranged from 2.5 Nm up to 12.7 Nm occurring at exoskeleton angles between 74.0° and 94.0° (Figure 44). These results for peak support moments and angles were generally consistent with findings in the literature (de Vries et al., 2019; Hartmann et al., 2021; Musso et al., 2022; Pinho & Forner-Cordero, 2022; Watterworth et al., 2023) with slight variations attributed to differences in the study design and exoskeleton age.

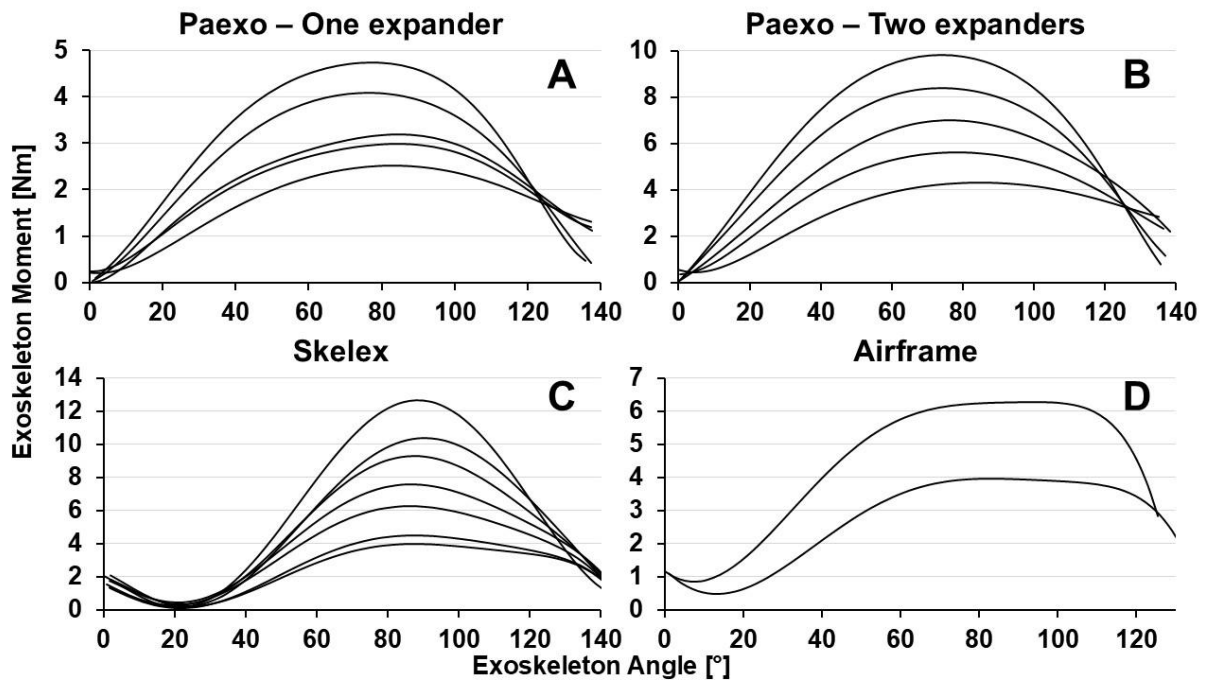


Figure 44: Exoskeleton support moments (values calculated from the polynomial regression equations) as a function of the exoskeleton angle for all available support levels of the Paexo in the configuration with one (A) and two expanders (B, each configuration with five support levels), the Skelex (C, seven support levels) and the Airframe (D, two support levels). The bottom curve in each graph represents the lowest support level while the top curve represents the highest support level (see chapter 8.4).

In two laboratory studies, the influence of two of the OSEs (Paexo and Airframe) on mechanical load at the GHJ and shoulder muscle demand during various static arm elevations was analyzed (Figure 45). When using the Airframe during static arm elevations, a reduction in mechanical load at the GHJ (average reduced net shoulder joint moment of 5.4 Nm) and shoulder muscle demand (average reduced anterior deltoideus [AD] muscle activity of 7.2 %MVC) was found independent of the arm posture (Figure 45). The greatest reduction in the shoulder joint moment and AD muscle activity was reported for an arm elevation of 90°. Similar beneficial effects were found in the second laboratory study for the Paexo when used during static arm elevations, which also led to a reduction in mechanical load at the GHJ (average reduced net shoulder joint moment of 3.1 Nm) and higher MTO in the anterior deltoideus (as indicated by an average increase in the tissue oxygenation index of 10.9%), even when the external load was increased (Figure 45).

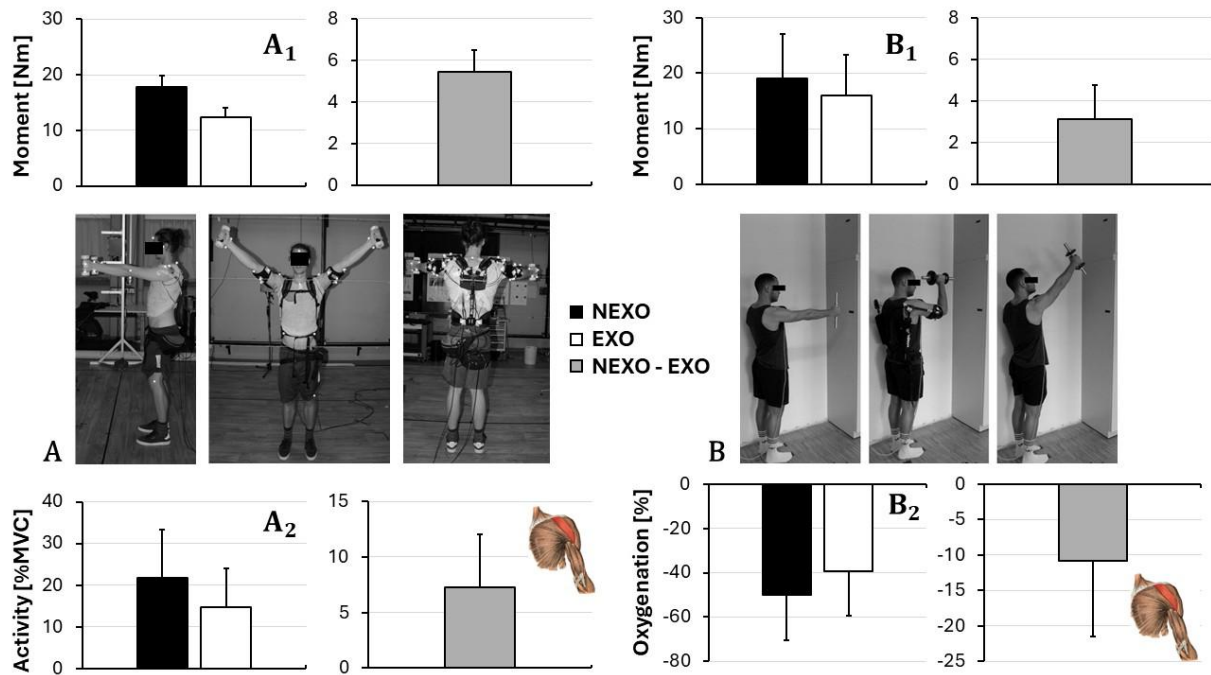


Figure 45: Summary of the results of the two laboratory studies (A & B) for the shoulder joint moment (A_1 & B_1) and muscle activity (A_2) and changes in tissue oxygenation index (B_2) of the anterior deltoideus in different arm postures with (white, EXO) and without (black, NEXO) an shoulder exoskeleton (Airframe A and Paexo B). Mean bar plots summarize the values across all experimental conditions of the two studies (six in each study). Error bars indicate + or - standard deviation and grey bars the mean reduction (A_1 , B_1 & A_2) or increase (B_2) between EXO conditions.

Consistent with the findings of the previous study, the greatest effect on both mechanical load at the GHJ and MTO was reported when the arm was in 90° elevation. These results were expected, as the peak support moments for both OSE were estimated to occur between approximately 74° and 94°, based on the moment-angle curves established in the first study. Conversely, the effects on muscle activity and muscle tissue oxygenation found in the two laboratory studies at 90° shoulder elevation further support the estimated angular positions for the peak support moment. This can be assumed as both parameters are directly linked to the muscle response in relation to the support provided by the respective OSE. For higher (120° and 135°) and lower (60°) elevation angles, smaller effects were reported in both studies, regardless of the parameter analyzed.

While these two studies provide a valuable insight into the biomechanical effectiveness of OSEs in reducing musculoskeletal load in the shoulder, the extent to which these findings can be generalized to real working scenarios is limited and may not align with laboratory results (de Bock et al., 2021; Pacifico et al., 2022). Therefore, a fourth study was conducted with aircraft manufacturers to analyze the influence of the two OSEs (plus the Skelex) in reducing musculoskeletal shoulder load during real work activities (here: drilling, riveting and screwing). In all three tasks (Figure 46), wearing one of the OSEs resulted in a reduction of mechanical load (average reduced shoulder joint moment 4.0 Nm across all tasks and OSE) and, in most cases, to a reduction of muscle demand of the shoulder elevators and partially the shoulder stabilators (average reduced shoulder muscle activity 2.7 %MVC across all tasks and OSEs).

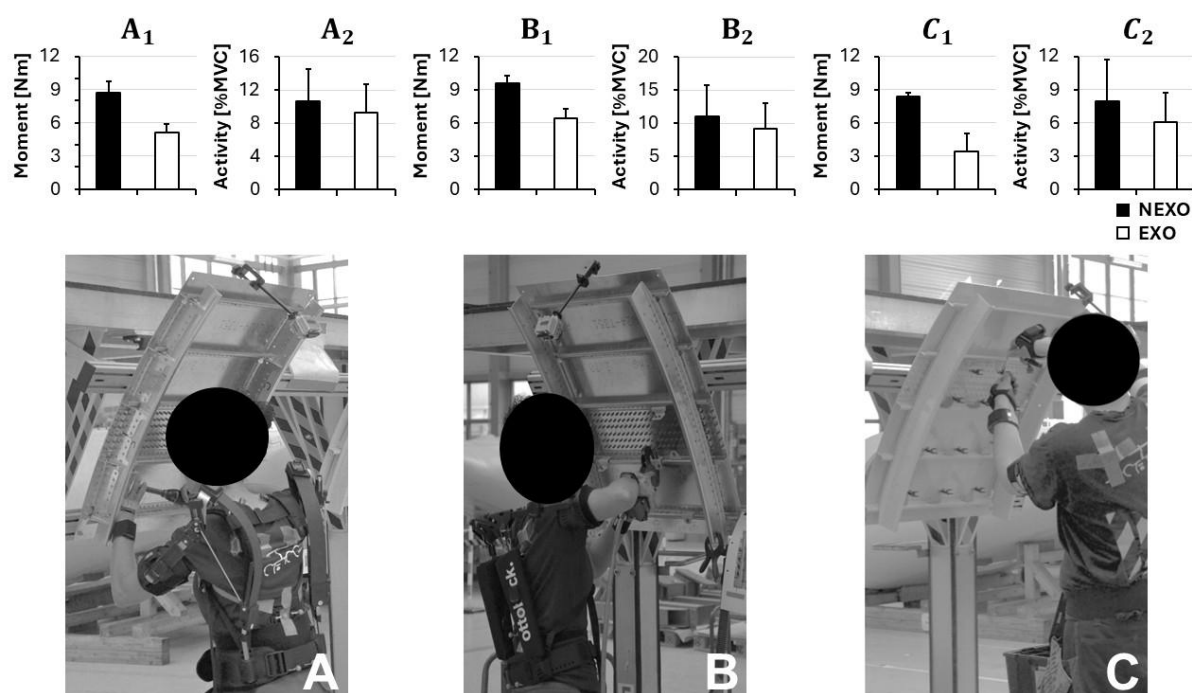


Figure 46: Summary of the results of the three aircraft manufacturing tasks (drilling A, riveting B and screwing C) for the shoulder joint moment (A₁, B₁, C₁) and muscle activity (A₂, B₂, C₂, averaged over all analyzed muscles) with (white, EXO) and without (black, NEXO) the shoulder exoskeletons (averaged over all three EXO conditions). Error bars indicate + standard deviation.

In summary, as observed in both the laboratory and field studies, the support provided by an OSE resulted in a reduction in the shoulder joint moment. Furthermore, a decrease in muscle activity in the anterior and medial part of the deltoideus was partially observed both during static arm elevation and the aircraft manufacturing tasks while wearing an OSE indicating a decrease in muscle demand. In addition, when wearing the Paexo during static arm elevations higher muscle tissue oxygenation (MTO) in RAD was found which also implies a reduction in muscle activity due to an expected linear relationship between muscle activity and MTO for static muscle contraction (Elcadi et al., 2011). Thus, it can be concluded that the support moment provided by an OSE decreases musculoskeletal load in the shoulder both in a laboratory setting and under real working conditions by reducing mechanical load at the GHJ and shoulder muscle demand. Generally, the decrease in muscle activity when wearing an OSE indicates a lower number of activated muscle fibers (Robertson et al., 2014). This could further suggest a reduction in the muscle forces required to maintain arm elevation, e.g. in the deltoideus muscles, as the tasks performed in these thesis were either static or quasistatic (i.e. muscle fiber length is constant, contraction velocity is approximately zero; Robertson et al., 2014). Therefore, a reduction in muscle forces could further result in a corresponding decrease in the joint compression- and contact forces in the glenohumeral joint and consequently in a reduced mechanical load at the joint (Praagman et al., 2000). This interpretation can be supported by the reported reduction in the net shoulder joint moment when wearing an OSE as a linear relationship between the net shoulder joint moment (as indicator of the mechanical load at the GHJ) and the joint compression forces when the arm is maintained in a static or quasi-static elevation can be assumed (Praagman et al. 2000). The decrease in muscle forces could also result in reduced metabolic demands and oxygen consumption (Perrey et al., 2010a) and lower intramuscular pressure (IMP) of the affected muscle, the latter contributing to a less impaired muscle blood flow (Järvholm et al., 1988b; Järvholm et al., 1991b). Therefore, it can be hypothesized that OSE use may delay (or even prevent) the development of muscle fatigue, thereby potentially preserving glenohumeral joint kinematics (Ebaugh et al., 2006), and superior translation of the humeral head (Chopp et al., 2010b). This could further reduce the risk of increased impairment of the subacromial space during the arm elevation and consequently help in preventing the development of impingement syndrome.

When comparing the results of studies conducted in a laboratory setting with those of the aircraft manufacturing tasks (performed under real working conditions), it was found that the effects of the OSE were partially more pronounced under laboratory conditions. For instance, wearing the Airframe during the static arm elevations in the laboratory reduced the shoulder joint moment and anterior deltoideus muscle activity by an average of 5.4 Nm and 7.2 %MVC, respectively. Under real working conditions, the reductions for both parameters were smaller: 3.9 Nm for the shoulder joint moment and 4.2 %MVC for AD muscle activity when working with the Airframe across all three manufacturing tasks. For the Paexo, a higher average reduction (4.4 Nm) was found for the shoulder joint moment during static arm elevations in 90° shoulder flexion in the laboratory compared to the drilling and riveting task (3.3 Nm and 3.1 Nm, respectively, both tasks performed approximately between shoulder and eye level). As mentioned in the previous chapter, this difference in the effects of an OSE when comparing laboratory and field results has been discussed in previous studies (de Bock et al., 2021; Pacifico et al., 2022). In their study, de Bock et al. (2021) reported a discrepancy in the effectiveness of two passive OSEs when tested under controlled laboratory conditions during isolated work activities compared to their use in a real work scenario in the field. Although they also found positive effects of the OSEs during isolated activities in the laboratory, the transferability of these effects to real work scenarios was limited. While Pacifico et al. (2022) reported a reduction in total shoulder muscle activity during similar work activities performed both in the field and in a simulated environment, this reduction was lower in the field compared to the simulated environment (Pacifico et al., 2022). In most of the laboratory studies, the tasks were standardized according to the anthropometrics of the participants, either by adjusting the working height for the specific task (e.g. Kim et al., 2018; Pinho & Forner-Cordero, 2022; Ojelade et al., 2023; Schmalz et al., 2019; van Engelhoven et al., 2019) or by directly assigning specific arm postures (e.g. Brunner et al., 2023; Desbrosses et al., 2021; de Vries et al., 2019). Therefore, as previously mentioned, a similar effect of the OSE on the mechanical load at the GHJ and muscle demand can be expected as the exoskeleton angle is typically kept constant across participants which can result in a reduction in muscle activity of up to 18 %MVC (van Engelhoven et al., 2019). In contrast, in the field, the working height often cannot (or only partially) be adjusted to match individual anthropometrics.

This leads to the adaptation of varying arm postures, i.e. working techniques, among the workers, especially when there are large differences in the worker's anthropometrics. Thus, differences in arm postures also lead to variations in the exoskeleton angle and, conversely, in the support moment. This increases the variability in the OSE's impact on the mechanical load at the GHJ and muscle demand.

According to the findings of Chopp et al. (2010a), it is preferable to adapt the workstation to the worker's individual anthropometry when performing an overhead task, as this can reduce muscle shoulder demand compared to a fixed working height (Chopp et al., 2010a). In addition, it is recommended that the target point for the hand during task execution be positioned closer to the body, either by adjusting the workstation configuration or by using tools such as a step ladder (Anton et al., 2001; Chopp et al., 2010a; Maciukiewicz et al., 2016). Positioning the hand lower and closer to the GHJ, thereby reducing the shoulder elevation angle, can decrease the shoulder joint moment and the corresponding shoulder muscle activity (Anton et al., 2001; Chopp et al., 2010a; Maciukiewicz et al., 2016). This, in turn, leads to a reduction of mechanical shoulder load and muscle demand (Anton et al., 2001; Chopp et al., 2010a; Maciukiewicz et al., 2016). This may further reduce the risk of developing a shoulder impingement syndrome, especially when the arm is lowered below 60° elevation (Chen et al., 1999; Flatow et al., 1994).

In case where workstation modifications are not (or only partially) feasible, an OSE is considered a viable intervention to mitigate work-related MSL in the shoulder. However, its implementation in the workplace should still be considered with caution and should not be the first solution in order to reduce work-related MSL (Moeller et al., 2022; Steinhilber et al., 2020). In line with the STOP principle, it is suggested to prioritize 1) substitutional measures, e.g. full automation of the work processes (de Looze et al., 2016); 2) technical measures, e.g. modification of the workstation (Bär et al., 2021), ideally to match the individual anthropometry of the workers (Chopp et al., 2010a); or 3) organizational measures, e.g. reducing the duration and frequency of working in constrained postures (Steinhilber et al., 2020). Only after these options have been considered should the use of an OSE be evaluated, since as a wearable device, it is more similar to a personal protective equipment (PPE; Lowe et al., 2019) or measure (PPM; Steinhilber et al., 2020).

In addition, as suggested in the first laboratory study (chapter 9) and for the aircraft manufacturing tasks, the residual muscle demand (indicated by the muscle activity) may still be unacceptable (above 10 %MVC; Jonsson, 1982) when the task is performed over a longer period of time (one hour or more; Jonsson, 1982). Additionally, Järvholm et al. (1991b) pointed out, that even when arm support is provided, residual muscle forces may still be sufficient to impair blood flow in shoulder muscles and thereby contribute to the development of shoulder muscle fatigue. Therefore, the implementation of an OSE alone may not be sufficient to prevent the development of shoulder muscle fatigue during overhead work even if the musculoskeletal load is reduced (Gillette et al., 2022).

Gillette et al. (2022) reported a reduction in anterior deltoideus activity and therefore muscle demand when performing various overhead automotive assembly tasks using the Airframe. However, conflicting results were found regarding the effect of the OSE on muscle fatigue: in four processes, muscle activity was not sufficiently reduced below a calculated threshold level (relative to the duration of muscle activation during the process) leading to the prediction that these processes would still lead to the development of muscle fatigue. A possible way to increase the effectiveness of OSEs in reducing shoulder muscle demand and in further preventing the development of muscle fatigue could be to adapt the support moment to the specific task demand, particularly, over extended periods of use. However, the support moment has to be carefully adjusted, as an increase could also lead to a higher muscle activity of the shoulder stabilizing muscles (such as the trapezius and infraspinatus; Ojelade et al., 2023; van Engelhoven et al., 2019).

Subsequently, a higher support moment would also increase the forces acting on the arm cuff (both the vertical force and the shear forces) as shown by the findings of the COBOT study (chapter 8) and noted by Musso et al. (2022), especially for the Skelex. This increase may result in greater user discomfort, even if the impact on physiological and biomechanical parameters is considered positive (Weston et al., 2022). Furthermore, it is still unclear how these acting forces (vertical and horizontal) affect the position of the humeral head in relation to the glenoid cavity when the support moment is increased. Seiferheld et al. (2022) concluded, based on their simulation of OSE use (ShoulderX) during an overhead manual material handling task, that the optimal moment-angle profile must be selected according to the specific task demands.

For the maximum support moment, they estimated an even higher musculoskeletal load, especially at lower elevation angles, represented by an increase in infraspinatus muscle activity to stabilize the humeral head against anterior dislocation (Seiferheld et al., 2022). These findings are consistent with the results for the aircraft manufacturing tasks and of van Engelhoven et al. (2019), and further support the hypothesis that the support moment of an OSE needs to be carefully adjusted to reduce muscle demand in the rotator cuff muscles. Finally, another possible way to effectively reduce MSL in the shoulder during work at or above shoulder level might be to adapt the workstation to the worker's anthropometry in order to promote ergonomically acceptable postures with the additional application of an OSE adapted to the remaining task demands. This can further reduce shoulder MSL to a level that potentially prevents the development of shoulder muscle fatigue.

As outlined in the previous chapters, each of the studies presented had its own specific limitations that must be taken into account when interpreting the results. In addition to these study specific aspects, there are overarching limitations that apply to thesis as whole. First, the sample sizes in all studies were small and kept to a minimum, especially for the laboratory studies. In the two laboratory studies, the small sample sizes were due to limited availability of participants, increased legal requirements and research institute-dependent guidelines, as both studies were conducted during the COVID-19 pandemic. In the fourth study, it was not possible to recruit more workers because the aircraft manufacturing company could not release more workers from their duties in order to participate in the study due to an intensified time schedule in their production line. It was therefore decided to focus on a limited set of parameters to minimize the number of performed statistical tests and the alpha error corrections required.

Second, in order to calculate the net shoulder joint moment when wearing an OSE, the support moment was subtracted from the shoulder joint moment during both the static arm elevations and the aircraft manufacturing tasks. While this approach is sufficiently robust to estimate the effect of the OSEs on the mechanical load at the GHJ and has been applied in previous studies (Fritzsche et al., 2021; Musso et al., 2022; Seiferheld et al., 2022; van der Have et al., 2022), it is still unclear whether the support forces acting at the arm cuff (especially the shear forces) could influence the humeral head position and consequently, the joint reaction forces.

Therefore, future studies should, where possible, directly measure these forces during the use of an OSE (e.g. from a force sensor inside the arm cuff) and integrate them into their models to improve the simulation results.

Third, only three OSEs were analyzed in this thesis, but there are more commercially available models on the market (e.g. ShoulderX, Ekso EVO, EXHAUSS Stronger). Therefore, while the results of this thesis still provide a valuable addition to the existing literature, they are limited to a certain extent by the OSEs presented. As shown during the aircraft manufacturing tasks and in the literature (Desbrosses et al., 2021; Jorgensen et al., 2022a; Jorgensen et al., 2022b; Ojelade et al., 2023), the effectiveness of the OSEs in reducing musculoskeletal shoulder load may vary, even though the basic mechanical functionality (initiation of a support force at the upper arm) is identical. Differences in their design could also lead to different effects even when performing the same task (Desbrosses et al., 2021; Jorgensen et al., 2022a; Jorgensen et al., 2022b; Ojelade et al., 2023). In addition, the moment-angle profile can also vary and may change if the support moment or range is adjusted, as shown by the COBOT study (chapter 8) and by Watterworth et al. (2023). Therefore, the author recommends to first evaluate the workstation and its specific task demands (e.g. body postures, tool weight, work duration). Then, if the use of an OSE is still considered a viable ergonomic intervention, it should be selected based on the findings of this evaluation in order to match the specifications of the OSE with the requirements of the workplace.

13. Conclusion of the thesis

In summary, this thesis exploratively demonstrated the biomechanical effectiveness of occupational shoulder exoskeletons (OSE) in reducing musculoskeletal load (MSL) in the shoulder during tasks performed at shoulder or overhead level. The positive effects of the OSE were found not only on shoulder muscle demand (as indicated by reduced muscle activity and increased muscle tissue oxygenation), but also on mechanical load at the glenohumeral joint (as indicated by reduced shoulder joint moment). To the best of the author's knowledge, this is the first scientific work that combines the estimation of the support moment of the OSE with the implementation of moment-angle profiles in both laboratory settings and under real working conditions to assess their effect on the mechanical load at the glenohumeral joint (GHJ). Furthermore, this thesis presents a novel standardized measurement setup for the estimation of the support moment of different shoulder exoskeletons including the quantification of the forces acting on the arm cuff. This setup allows for further analysis of how different arm cuff configurations (e.g. distance from the exoskeleton joint center) affect support forces at the arm cuff, even when the support moment remains constant. When applied to the users, it is expected that the cuff forces decrease if the arm cuff is positioned further away from the joint center (Musso et al., 2022), which may occur in participants with longer arms. Therefore, this measurement setup can be used to infer how variations in the arm configuration due to differences in a participant's anthropometrics influence the resulting support forces. These findings may contribute to further improving the estimation of the support moment and may also provide a better insight into how the support forces influence humeral head positioning and joint reaction forces, especially in cases where the integration of force sensors within the arm cuff is not applicable. In addition, they can aid in interpreting future study results on perceived discomfort and muscular demand when performing a work task with a shoulder exoskeleton, as it has been shown that the effect can vary between these two measures (Weston et al., 2022). This work also revealed differences in the magnitude of the effect of OSEs when evaluated in the laboratory under standardized conditions (i.e. similar arm postures and/or external load conditions) compared to when performing real work tasks with an OSE at non-standardized workstations (similar to previous studies; de Bock et al., 2021; Pacifico et al., 2022).

Finally, as stated by the author, although the OSEs showed beneficial effects on musculoskeletal shoulder load during work above or at shoulder level, it remains uncertain whether these effects are sufficient to prevent the development of shoulder muscle fatigue during prolonged work. Therefore, it is proposed to analyze the effects of the OSEs during a working day to further assess whether the muscles still develop fatigue. If a clear prevention of fatigue is estimated, the use of OSEs should be evaluated in long-term studies (months to years possibly compared to a control group) to analyze their potential in preventing the development of work-related musculoskeletal disorders (WMSD). It is also recommended that future analyses include subjective measures (questionnaires or the BORG scale) to assess perceived discomfort and/or exertion when working with the OSE, in order to complement biomechanical and physiological data. Furthermore, the worker's kinematics should be additionally recorded to evaluate if the OSE changes the body posture, which has already been reported in previous studies (McFarland et al., 2022; Schmalz et al., 2019). Finally, if feasible, long-term studies should be accompanied by regular medical examinations of participants to detect any orthopedic changes (particularly within the subacromial space) that may be attributed to OSE use, whether beneficial or adverse.

14. Zusammenfassung

Schulterexoskelette (SE) stellen eine vielversprechende ergonomische Maßnahme dar, um arbeitsbedingte Muskel-Skelett-Erkrankungen (AMSEs) der Schulter zu reduzieren. Solche Erkrankungen entstehen häufig durch eine erhöhte Muskel-Skelett-Belastung (MSB), d.h. eine erhöhte Beanspruchung der Muskulatur sowie der mechanische Belastung am Glenohumeralgelenk (GHG), bei Tätigkeiten auf oder über Schulterhöhe. SEs dieser Belastung entgegen, indem sie das Schultergelenksmoment (SGM), erforderlich für die Elevation des Armes, reduzieren und dadurch die Schultermuskelkräfte und somit die Beanspruchung der Muskulatur verringert werden. Während die Effekte von SEs auf die Muskelbeanspruchung bereits umfassend in Labor- und Feldstudien untersucht wurden, ist ihr Einfluss auf die mechanische Belastung am GHG und auf die lokale Muskelgewebeeröxygenierung (MGO) bislang unzureichend erforscht. Beide Parameter sind jedoch entscheidend für ein besseres Verständnis des Einflusses von SEs auf Schultermuskelkräfte und MSB. Ziel dieser Arbeit war daher, die explorative Analyse der biomechanischen Wirksamkeit von SEs in der Reduktion von Schulter-MSB bei Tätigkeiten auf oder über Schulterhöhe. Untersucht wurden insbesondere das SGM, die Muskelaktivität und die MGO. Um den Effekt auf das SGM zu bestimmen wurde das Unterstützungsmoment von drei SEs quantifiziert mit Höchstwerten zwischen 2.5 Nm und 12.7 Nm. Zwei SEs (Paexo Shoulder und Levitate Airframe), jeweils in separaten Laborstudien getestet, reduzierten das SGM bei statischen Armhebungen um durchschnittlich 4,3 Nm. Gleichzeitig wurden eine Abnahme der Muskelaktivität um 7,2 %MVC (engl. maximal voluntary contraction, d.h. in Relation zu einem maximalen Referenzwert; Studie 1) und ein Anstieg des Gewebeeröxygenierungsindex um 10,9 % (Studie 2) beobachtet. In drei praxisnahen Tätigkeiten aus dem Flugzeugbau (Bohren, Nieten und Schrauben) führte das Tragen eines SEs zu einer Reduktion des SGM um etwa 4,0 Nm sowie überwiegend zu einer Verringerung der Muskelaktivität um 2,7 %MVC, insbesondere in Teilen des M. Deltoideus. Diese Ergebnisse zeigen, dass SEs sowohl im Labor als auch im realen Arbeitsumfeld die MSB in der Schulter verringern, in dem sie die mechanische Belastung am GHG und die muskuläre Beanspruchung reduzieren. Die Effekte waren dabei unter realen Bedingungen etwas geringer als unter kontrollierten Laborbedingungen. Insgesamt legen die Befunde nahe, dass SEs somit ein präventives Potential zur Reduktion von AMSEs in der Schulter besitzen.

15. Abstract

Occupational shoulder exoskeletons (OSEs) represent a promising ergonomic intervention to reduce work-related musculoskeletal disorders (WMSDs) of the shoulder. These disorders often result from increased musculoskeletal load (MSL), i.e. an increased muscle demand and mechanical load at the glenohumeral joint (GHJ), during tasks performed at or above shoulder level. OSEs counteract this load by reducing the required shoulder joint moment (SJM) to elevate the arm, thereby reducing shoulder muscle forces and consequently decreasing muscle demand.

While the effects of OSEs on shoulder muscle demand have been extensively investigated in both laboratory and field studies, their influence on mechanical load at the GHJ (as indicated by the SJM) and on local muscle tissue oxygenation (MTO) has not yet been sufficiently explored. However, both parameters are crucial for a more comprehensive understanding of the effects of SEs on shoulder muscle forces and MSL.

The aim of this thesis was therefore to conduct an exploratory analysis of the biomechanical effectiveness of OSEs in reducing shoulder MSL during tasks performed at or above shoulder level. The analysis focused in particular on their influence on the SJM, muscle activity and MTO. To determine the effect on the SJM, the support moment provided by three OSEs was quantified revealing maximal values between 2.5 Nm and 12.7 Nm. Two of the OSEs (Paexo Shoulder and Levitate Airframe), each tested in separate laboratory studies, reduced the SJM during various static arm elevations by an average of 4.3 Nm. At the same time, a decrease in muscle activity of 7.2 %MVC (maximal voluntary contraction, i.e. in relation to a maximal reference value; Study 1) and an increase in tissue oxygenation index of 10.9% (Study 2) were observed. In three industry-representative tasks from aircraft manufacturing (drilling, riveting and screwing), wearing an OSE led to a reduction in the SJM by approximately 4.0 Nm and, in most cases, to a decrease in shoulder muscle activity by 2.7 %MVC, particularly in parts of the deltoideus muscle.

These results indicate that OSEs reduce MSL by decreasing mechanical load at the GHJ and muscle demand, both in laboratory setting and under real working conditions, with effects being somewhat smaller in the latter. Overall, the findings suggest that OSEs may have preventive potential for reducing shoulder WMSEs.

16. List of Figures

Figure 1: Days of incapacity to work by diagnostic groups in Germany for the years 2013 and 2023. The data were derived from BAuA, 2014 and BAuA, 2024.	1
Figure 2: Anterior view of the upper part of the sternum (A) and the shoulder complex (B). Both graphics were extracted and modified from Sobotta (1909; A; p. 120) and Gray and Lewis (1918; B; p. 316). Public domain.....	4
Figure 3: Glenohumeral joint in anterior (sliced view A) and sagittal (B) view. Both graphics (A & B) were extracted and modified from Sobotta (1909; p. 120). Public domain.....	5
Figure 4: Anterior (A) and posterior view (C) on various shoulder muscles in addition with a sliced posterior view of the rotator cuff muscles (B). All three graphics were extracted and modified from Gray and Lewis (1918; p. 436, p. 441 & p. 433). Public domain.....	7
Figure 5: Glenohumeral joint with the surrounding muscles in posterior (A) and anterior (B) view. Thin arrows indicate force direction during contraction for the rotator cuff muscles (black) and the deltoideus parts (white). Thick arrows indicate the direction the humeral head is moving caused by the respective muscle contraction. Based on Singleton (1966). Both graphics were extracted and modified from Sobotta (1909; p. 186). Public domain.	8
Figure 6: Calculation of the gravitational shoulder joint moment ($\tau_{gravshoulder}$; A) as a product of the gravitational force ($F_{gravarm}$), acting at the center of gravity (CoG), and the distance between the glenohumeral joint center and the CoG (r_{CoGarm}). In addition, $\tau_{gravshoulder}$ is presented for an exemplary participant performing an arm elevation in 90° shoulder flexion (arm fully extended, wrist in neutral position) for five seconds (B).	10
Figure 7: STOP – principle for the prevention or reduction of musculoskeletal load at the workplace (DGUV, 2023; Steinhilber et al., 2020).	17
Figure 8: Example of the functional principle of an occupation shoulder exoskeleton (OSE). The exoskeleton generates a force FE , which in turn generates a corresponding support moment over the internal lever arm of the exoskeleton. This support moment then transmits a corresponding force FA to the upper arm via an external lever arm. This causes a reduction in the shoulder joint moment, which is intended to reduce the	

demand on the shoulder muscles (modified photo: © Airbus Defence & Space GmbH, 2021).	21
Figure 9: Measurement setup of the collaborative robot (COBOT) with an attached force sensor (B; white square, right upper corner), retro-reflective markers placed the arm (A; white square) and shoulder segment (B, white square, left lower corner) and an exemplary exoskeleton.	29
Figure 10: Illustrations of the self-constructed top of the force sensor (visualized in Solidworks; version 2022, Dassault Systemes Deutschland GmbH, Stuttgart, Germany) in isometric view as full (A) and see through model (D) as well as in side (B) and top view (C).	30
Figure 11: Measurement setup with the collaborative robot (COBOT) for the three tested occupational shoulder exoskeletons (OSE); Levitate Airframe (A), Skelex 360 (B) and Paexo Shoulder (C); in front and back view. Each exoskeleton and the COBOT were equipped with retro-reflective markers (at least three per segment).	32
Figure 12: Measurement setup of the collaborative robot (COBOT) with an attached force sensor (white circle), an exemplary exoskeleton and the four angular positions (T_{int} , T_{min} , T_{90} and T_{max}) which were defined in each measurement session.	33
Figure 13: Exemplary illustration of the mathematical approach to reconstruct the global coordinates (X, Y and Z) of a point (PB) on a rigid body equipped with three reflective markers (indicated by circles) from its local coordinates (X_B, Y_B, Z_B). The local coordinates were estimated by the global coordinates (X, Y and Z) of the top of the pointer (PP). The global coordinates of the pointer top PP were calculated from its respective local coordinates (X_P, Y_P, Z_P) obtained during a calibration trial.	35
Figure 14: Mechanism to increase the support level (SL) for the exoskeletons Paexo (A) and Skelex (B) by increasing the distance between the point of force application and the rotational joint center of the respective exoskeleton ($JCExo$), i.e. the internal lever arm ($r_{internal}$). FE is the retraction force generated by the passive structures of the exoskeleton (white arrow indicates the force direction). For both exoskeletons, two exemplary support levels are marked (PSL11 and PSL15 for the Paexo and SxSL1 and SxSL4 for the Skelex).	37
Figure 15: Force sensor attached to the ring on the collaborative robot (COBOT) for the measurement setup of the exoskeleton Paexo and the coordinate system of the sensor (X, Y and Z).	39

Figure 16: Exoskeleton support moment, estimated from the highest support level for the exoskeleton Paexo, relative to the measurement time (A) and the exoskeleton angle (B) during ten dynamic elevations. Dotted lines (A) indicate the start and end time of each elevation. Arrows (B) indicate the upward (from left to right) and downward phase (from right to left) during the elevations.....	41
Figure 17: Mean support moment for the five support levels (PSL11-PSL15) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B)..	43
Figure 18: Mean support moment for the five support levels (PSL21-PSL25) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B)..	45
Figure 19: Mean support moment for the seven support levels (SxSL1-SL4) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B)..	47
Figure 20: Mean support moment for the two support levels (ASL3 and ASL4) relative to the mean exoskeleton angle with their respective standard error of means ($\pm$, A) and with the modeled values calculated from the polynomial regression equations (B)..	49
Figure 21: Measurement setup for the Paexo Shoulder in front (A) and back view (B). To attach the Paexo to the measurement table, the protective cover around the expanders (C) was removed. The exoskeleton angle was defined between the arm element (line 3 in C) and the shoulder element (line 2 in C) in this study. For Watterworth et al. (2023), it was assumed that the exoskeleton angle was defined between the flat side of the cover (line 1 in C) in relation to the arm element (line 3 in C).....	55
Figure 22: Participant performing arm elevations with (B,C) and without (A) an occupational shoulder exoskeleton in three arm postures (values for SEA and SAA specify the respective shoulder elevation and abduction angle): (A) SEA = 90°, SAA = 0°, (B) SEA= 120°, SAA = 30°; and (C) SEA = 90°, SAA = 30°.....	60
Figure 23: Illustration of the calculation of the shoulder elevation (α_{elev} , A) and shoulder abduction angle (α_{abd} , B) with the respective retro-reflective markers placed on the participants (black circles) and the muscle analyzed in this study: the musculus anterior deltoideus (C, red) and the musculus medial deltoideus (D, red). Graphics C and D were extracted and modified from Gray and Lewis (1918; p. 436).....	61

Figure 24: Estimation of the maximum net shoulder joint moment ($\tau_{maxshoulder}$, C) for 90° shoulder elevation as the sum of the gravitational shoulder joint moment ($\tau_{gravshoulder}$, A) and the maximum shoulder elevation moment ($\tau_{maxelev}$, B), the later obtained by the maximal voluntary contraction trial (MVC). In this posture, $\tau_{gravshoulder}$ was estimated as the product of the gravitational force ($F_{gravarm}$), acting at the center of gravity (CoG), and the distance between the glenohumeral joint center and the CoG (r_{CoGarm}). $\tau_{maxelev}$ was estimated as the product of the elevation force ($F_{maxelev}$) and the arm length (l_{arm}). 63

Figure 25: Estimation of the shoulder elevation angle (α_{elev}) by the dot product ($\cdot$) of the vectors (vec) between the three markers UPA, KNE and FIN, with $norm$ indicating the calculation of the norm (i.e. length) of the specific vector and $arcos$ indicating the calculation of the angle based on its corresponding arcus cosines value..... 65

Figure 26: Muscle activity of the right anterior deltoideus (RAD) and right relative net shoulder joint moment ($RNSM_{rel}$) of an exemplary participant for the static arm elevation performed in 90° shoulder flexion with 0° shoulder abduction. Vertical dot lines show the start time and end time for the static phase (five-second window) during an arm elevation. The graphic for the anterior deltoideus (right side) was extracted from Gray and Lewis (1918; p. 436) and modified. 68

Figure 27: Mean bars of the mean (m) right relative net shoulder joint moment during static arm elevations in 60° (A), 90° (B) and 120°(C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXOA) and without (baseline, black, NEXO) a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference ($p < 0.05$) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXOA conditions. 71

Figure 28: Mean bars of the mean (m) activity for the right anterior deltoid muscle during static arm elevations in 60° (A), 90° (B) and 120°(C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXOA) and without (baseline, black, NEXO) a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference ($p < 0.05$) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXOA conditions. 73

Figure 29: Mean bars of the activity for the right medial deltoid muscle during static arm elevations in 60° (A), 90° (B) and 120°(C) shoulder elevation with a shoulder abduction angle of 0° and 30° with (white, EXOA) and without (baseline, black, NEXO)

a shoulder exoskeleton. Error bars indicate + standard deviation and stars (*) over the bars a significant difference ($p < 0.05$) between baseline and exoskeleton condition. Grey bars show the mean reduction between EXOA conditions. 74

Figure 30: Participant performing an arm elevation in the sagittal plane with (B) and without (A,C) an shoulder exoskeleton in three arm postures (values for SF and EF specify the respective shoulder flexion and elbow flexion angle): (A) SF = 90°, EF = 0°, (B) SF = 90°, EF = 90°; and (C) SF = 135°, EF = 0°. Black markers were placed on the wall in front of the participant and aligned to their anthropometric measurements to help maintain the desired shoulder elevation angle. 81

Figure 31: Near-infrared spectroscopy (NIRS) procedure (A). Infrared light is emitted (LED) into deeper muscle tissues (capillary level) and the reflected intensity along the path p is measured via a detector. The measurement depth is approximately half the distance between LED and detector (d) which are combined in a probe (optode; B). The graphic of the capillary was modified from CERT (n.d.), licensed under CC BY-SA 3.0, via Wikimedia Commons. Changes were made by the author. 84

Figure 32: Mean changes in tissue oxygenation index (TOI) over time during an arm elevation (duration 60 s) in 90° shoulder flexion (elbow flexion 0°) and subsequent recovery phase (section of 120 s) with the arm lowered in neutral position. Time points t_0 mark the start point of the arm elevation. Change in TOI (ΔTOI) was calculated as the difference between the respective TOI values measured at the two time points t_0 and t_1 (55 s before the end of an arm elevation). Error bars indicate $\pm$ standard error of mean..... 86

Figure 33: Mean changes in tissue oxygenation index (TOI) over time during an arm elevation (duration 60 s) in 90° shoulder flexion (SF) and 0° elbow flexion with (EXOP) and without (NEXO) a shoulder exoskeleton for an external sagittal shoulder joint load of 20% (A) and 30% (B) derived from maximum voluntary contraction (MVC). Error bars indicate $\pm$ standard error of mean. 88

Figure 34: Mean bar plots for the changes in tissue oxygenation index (ΔTOI) during an arm elevation with (white, EXOP) and without (black, NEXO) an shoulder exoskeleton in three static arm postures (AP; values for SF and EF specify the respective shoulder flexion and elbow flexion angle) for an external sagittal shoulder joint load of 20% and 30% derived from maximum voluntary contraction (MVC). Error bars indicate - standard deviation and stars (*) over the bars a significant difference (p

< 0.05) between baseline and exoskeleton condition. Grey bars show the mean increase between EXOP conditions (NEXO - EXOP).	89
Figure 35: Mobile platform with two aircraft panels mounted at a fixed height and angle.	96
Figure 36: Two aircraft panels mounted on the mobile platform to perform a drilling and riveting task at two metal brackets each at a different stringer height (A) and screwing twelve mounting clips at four different stringer heights (B & C).	97
Figure 37: Two exemplary participants performing the drilling (A), riveting (B) and screwing task (C) with (A & B) and without (C) an occupational shoulder exoskeleton.	98
Figure 38: Illustration of the sensor placement of the XSens motion capture suit on a participant (black squares; A) and the muscle analyzed in this study (B): the anterior deltoideus (1), the medial deltoideus (2), the upper trapezius (3), and the infraspinatus (4). All graphics in B were extracted and modified from Gray and Lewis (1918; p. 433 & p. 436). The graphic of the human body silhouette (male; A) was modified from Openclipart (n.d.). Public domain. Changes were made by the author.	100
Figure 39: Exemplary participant performing maximal voluntary contraction (MVC) trials for the right arm in the two specific test postures “empty can” (A) and “125° flexion (B).	101
Figure 40: Muscle activity of an exemplary participant during a measurement. Two-time sections when performing the drilling task at the lower (A, Drilling_LS) and at the upper stringer height (B, Drilling_US), respectively, were highlighted (black squares) and the respective snippet from the synchronized video was included.	103
Figure 41: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex; A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton.	107
Figure 42: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex;	

A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton. 109

Figure 43: Mean bar of the mean (m) muscle activity for the anterior deltoideus (AD), medial deltoideus (MD), upper trapezius (UT) and infraspinatus (INF) as well as of the mean (m) net shoulder joint moment (NSM) for the left and right side (L/R) in the baseline (B, NEXO) and the three exoskeleton (EXO) conditions (P: Paexo; S: Skelex; A: Airframe). Error bars indicate + standard deviation and stars (*) a significant difference ($p < 0.05$) between baseline and the respective exoskeleton. 111

Figure 44: Exoskeleton support moments (values calculated from the polynomial regression equations) as a function of the exoskeleton angle for all available support levels of the Paexo in the configuration with one (A) and two expanders (B, each configuration with five support levels), the Skelex (C, seven support levels) and the Airframe (D, two support levels). The bottom curve in each graph represents the lowest support level while the top curve represents the highest support level (see chapter 8.4). 129

Figure 45: Summary of the results of the two laboratory studies (A & B) for the shoulder joint moment (A1 & B1) and muscle activity (A2) and changes in tissue oxygenation index (B2) of the anterior deltoideus in different arm postures with (white, EXO) and without (black, NEXO) an shoulder exoskeleton (Airframe A and Paexo B). Mean bar plots summarize the values across all experimental conditions of the two studies (six in each study). Error bars indicate + or - standard deviation and grey bars the mean reduction (A1, B1 & A2) or increase (B2) between EXO conditions..... 130

Figure 46: Summary of the results of the three aircraft manufacturing tasks (drilling A, riveting B and screwing C) for the shoulder joint moment (A1, B1, C1) and muscle activity (A2, B2, C2, averaged over all analyzed muscles) with (white, EXO) and without (black, NEXO) the shoulder exoskeletons (averaged over all three EXO conditions). Error bars indicate + standard deviation. 131

17. List of Tables

Table 1: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the five support levels (PSL11-15).....	44
Table 2: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the five support levels (PSL21-25).....	46
Table 3: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the seven support levels (SxSL1-4).....	48
Table 4: Maximum standard error of the mean (SEM) for the measured support moment and exoskeleton angle, the peak support moment and its corresponding exoskeleton angle estimated from the polynomial regression equations and the order of the applied polynomial equations with the adjusted R^2 -value for the two support levels (ASL3 and ASL4).....	50
Table 5: Mean value ($\pm$ standard deviation) for the right absolute net shoulder joint moment, obtained from trials with (EXOA) and without (NEXO) exoskeleton, in all combinations of shoulder elevation (SEA) and shoulder abduction (SAA) angle.....	70
Table 6: Mean right net shoulder joint moment (RNSM) $\pm$ standard deviation during static arm elevation in 90° and 135° shoulder flexion (SF) with (EXOP) and without (NEXO) a shoulder exoskeleton for an external sagittal shoulder load (<i>SJLoad</i>) of 20% and 30% derived from maximum voluntary contraction (MVC).....	88

18. List of Abbreviations

ACJ	Acromioclavicular joint
AD	Anterior deltoideus muscle (LAD, left; RAD, right)
AMSE	Arbeitsbedingte Muskel-Skelett- Erkrankungen
AP	Arm posture
ASL	Airframe support level
ATT	Adipose tissue thickness
BAuA	Bundesanstalt für Arbeitsschutz und Arbeitsmedizin (engl. Federal Institute for Occupational Safety and Health)
BMAS	Bundesministerium für Arbeit und Soziales (engl. Federal Ministry of Labour and Social Affairs)
COBOT	Collaborative robot
CoG	Center of gravity
CoM	Center of mass
DGUV	Deutsche Gesetzliche Unfallversicherung
EF	Elbow Flexion
EMG	Electromyography
EU-OSHA	European Agency for Safety and Health at Work
EXO	Exoskeleton
FIN	Finger
GHG	Glenohumeralgelenk
GHJ	Glenohumeral joint
IFA	Institut für Arbeitsschutz (engl. Institute for Occupational Safety and Health of the German Social Accident Insurance)
IMP	Intramuscular pressure
IMU	Inertial measurement unit

INF	Infraspinatus muscle (LINF, left; RINF, right)
KNE	Knee
LT	Lower trapezius muscle
MD	Medial deltoideus muscle (LMD, left; RMD, right)
MGO	Muskelgewebeoxygenierung
MSB	Muskel-Skelett-Belastung
MSD	Musculoskeletal disorders
MSL	Musculoskeletal load
MT	Middle trapezius muscle
MTO	Muscle tissue oxygenation
MVC	Maximal voluntary contraction
NEXO	No exoskeleton
NIRS	Near-infrared spectroscopy
NSM	Net shoulder joint moment (LNSM, left; RNSM, right)
OSE	Occupational shoulder exoskeleton
PD	Posterior deltoideus muscle
PPE	Personal protective equipment
PPM	Personal protective measure
PSL	Paexo support level
RMS	Root mean square
SAA	Shoulder abduction angle
SARA	Symmetrical axis of rotation approach
SCA	Subscapularis muscle
SCJ	Sternoclavicular joint
SCoRE	Symmetrical Centre of Rotation Estimation
SE	Schulterexoskelette
SEA	Shoulder elevation angle
SEM	Standard error of mean
SF	Shoulder flexion
SGM	Schultergelenksmoment

SJ	Shoulder joint
SJM	Shoulder joint moment
SL	Support level
SMF	Shoulder muscle fatigue
STJ	Scapulothoracic joint
SUP	Supraspinatus muscle
SxSL	Skelex support level
TM	Teres minor muscle
TOI	Tissue oxygenation index
UPA	Upper arm
UT	Upper trapezius muscle (LUT, left; RUT, right)
WMSD	Work-related musculoskeletal disorders

19. List of References

- Alabdulkarim, S., Kim, S., & Nussbaum, M. A. (2019b). Effects of exoskeleton design and precision requirements on physical demands and quality in a simulated overhead drilling task. *Applied Ergonomics*, 80, 136–145. <https://doi.org/10.1016/j.apergo.2019.05.014>
- Alabdulkarim, S., & Nussbaum, M. A. (2019a). Influences of different exoskeleton designs and tool mass on physical demands and performance in a simulated overhead drilling task. *Applied Ergonomics*, 74, 55–66. <https://doi.org/10.1016/j.apergo.2018.08.004>
- Anton, D., Shibley, L. D., Fethke, N. B., Hess, J., Cook, T. M., & Rosecrance, J. (2001). The effect of overhead drilling position on shoulder moment and electromyography. *Ergonomics*, 44(5), 489–501. <https://doi.org/10.1080/00140130120079>
- Aptel, M., Aublet-Cuvelier, A., & Cnockaert, J. C. (2002). Work-related musculoskeletal disorders of the upper limb. *Joint Bone Spine*, 69(6), 546–555. [https://doi.org/10.1016/s1297-319x\(02\)00450-5](https://doi.org/10.1016/s1297-319x(02)00450-5)
- Arnoux, B., Farr, A., Boccara, V., & Vignais, N. (2023). Evaluation of a Passive Upper Limb Exoskeleton in Healthcare Workers during a Surgical Instrument Cleaning Task. *International Journal of Environmental Research and Public Health*, 20(4). <https://doi.org/10.3390/ijerph20043153>
- Bär, M., Steinhilber, B., Rieger, M. A., & Luger, T. (2021). The influence of using exoskeletons during occupational tasks on acute physical stress and strain compared to no exoskeleton - A systematic review and meta-analysis. *Applied Ergonomics*, 94, 103385. <https://doi.org/10.1016/j.apergo.2021.103385>
- Barcroft, H., & Millen, J. L. E. (1939). The blood flow through muscle during sustained contraction. *The Journal of Physiology*, 97(1), 17–31. <https://doi.org/10.1113/jphysiol.1939.sp003789>
- Bodin, J., Ha, C., Chastang, J.-F., Descatha, A., Leclerc, A., Goldberg, M., Imbernon, E., & Roquelaure, Y. (2012a). Comparison of risk factors for shoulder pain and rotator cuff syndrome in the working population. *American Journal of Industrial Medicine*, 55(7), 605–615. <https://doi.org/10.1002/ajim.22002>
- Bodin, J., Ha, C., Le Petit Manac'h, A., Sérazin, C., Descatha, A., Leclerc, A., Goldberg, M., & Roquelaure, Y. (2012b). Risk factors for incidence of rotator cuff syndrome in a large working population. *Scandinavian Journal of Work, Environment & Health*, 38(5), 436–446. <https://doi.org/10.5271/sjweh.3285>
- Boettcher, C. E., Ginn, K. A., & Cathers, I. (2008). Standard maximum isometric voluntary contraction tests for normalizing shoulder muscle EMG. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society*, 26(12), 1591–1597. <https://doi.org/10.1002/jor.20675>
- Brunner, A., van Sluijs, R., Luder, T., Camichel, C., Kos, M., Bee, D., Bartenbach, V., & Lambercy, O. (2023). Effect of passive shoulder exoskeleton support during working with arms over shoulder level. *Wearable Technologies*, 4, e26. <https://doi.org/10.1017/wtc.2023.21>
- Bundesanstalt für Arbeitsschutz und Arbeitsmedizin (2014). Sicherheit und Gesundheit bei der Arbeit 2013: Unfallverhütungsbericht Arbeit.
- Bundesanstalt für Arbeitsschutz und Arbeitsmedizin (2024). Sicherheit und Gesundheit bei der Arbeit - Berichtsjahr 2023: Unfallverhütungsbericht Arbeit. Advance online publication. <https://doi.org/10.21934/BAUA:BERICHT20240912>
- Bundesministerium für Arbeit und Soziales (2021). Wissenschaftliche Begründung für die Berufskrankheit Nr. 2117 "Läsion der Rotatorenmanschette der Schulter

- durch eine langjährige und intensive Belastung durch Überschulterarbeit, repetitive Bewegungen im Schultergelenk, Kraftanwendungen im Schulterbereich durch Heben von Lasten oder Hand-Arm-Schwingungen".
- Burns, W. C., & Whipple, T. L. (1993). Anatomic relationships in the shoulder impingement syndrome. *Clinical Orthopaedics and Related Research*(294), 96–102.
- CERT. (n.d.). *Capillary system* *CERT.jpg*. https://en.wikipedia.org/wiki/File:Capillary_system_CERT.jpg. Retrieved May 05, 2025
- Cha, J. S., Athanasiadis, D. I., Asadi, H., Stefanidis, D., Nussbaum, M. A., & Yu, D. (2024). Evaluation of a passive arm-support exoskeleton for surgical team members: Results from live surgeries. *Journal of Safety Research*, 89, 322–330. <https://doi.org/10.1016/j.jsr.2024.02.003>
- Chen, S.-K., Simonian, P. T., Wickiewicz, T. L., Otis, J. C., & Warren, R. F. (1999). Radiographic evaluation of glenohumeral kinematics: A muscle fatigue model. *Journal of Shoulder and Elbow Surgery*, 8(1), 49–52. [https://doi.org/10.1016/s1058-2746\(99\)90055-1](https://doi.org/10.1016/s1058-2746(99)90055-1)
- Chénier, F. (2021). Kinetics Toolkit: An Open-Source Python Package to Facilitate Research in Biomechanics. *Journal of Open Source Software*, 6(66), 3714. <https://doi.org/10.21105/joss.03714>
- Chopp, J. N., Fischer, S. L., & Dickerson, C. R. (2010a). The impact of work configuration, target angle and hand force direction on upper extremity muscle activity during sub-maximal overhead work. *Ergonomics*, 53(1), 83–91. <https://doi.org/10.1080/00140130903323232>
- Chopp, J. N., O'Neill, J. M., Hurley, K., & Dickerson, C. R. (2010b). Superior humeral head migration occurs after a protocol designed to fatigue the rotator cuff: A radiographic analysis. *Journal of Shoulder and Elbow Surgery*, 19(8), 1137–1144. <https://doi.org/10.1016/j.jse.2010.03.017>
- Coccia, A., Capodaglio, E. M., Amitrano, F., Gabba, V., Panigazzi, M., Pagano, G., & D'Addio, G. (2024). Biomechanical Effects of Using a Passive Exoskeleton for the Upper Limb in Industrial Manufacturing Activities: A Pilot Study. *Sensors (Basel, Switzerland)*, 24(5). <https://doi.org/10.3390/s24051445>
- Culham, E., & Peat, M. (1993). Functional anatomy of the shoulder complex. *The Journal of Orthopaedic and Sports Physical Therapy*, 18(1), 342–350. <https://doi.org/10.2519/jospt.1993.18.1.342>
- Dalbøge, A., Frost, P., Andersen, J. H., & Svendsen, S. W. (2014). Cumulative occupational shoulder exposures and surgery for subacromial impingement syndrome: A nationwide Danish cohort study. *Occupational and Environmental Medicine*, 71(11), 750–756. <https://doi.org/10.1136/oemed-2014-102161>
- Dalbøge, A., Frost, J., Grytnes, R., Roy, J.-S., Samani, A., & Høyrup Christiansen, D. (2024). Effects of a passive shoulder exoskeleton on muscle activity among Danish slaughterhouse workers. *Applied Ergonomics*, 114, 104111. <https://doi.org/10.1016/j.apergo.2023.104111>
- de Bock, S., Ghillebert, J., Govaerts, R., Elprama, S. A., Marusic, U., Serrien, B., Jacobs, A., Geeroms, J., Meeusen, R., & de Pauw, K. (2021). Passive Shoulder Exoskeletons: More Effective in the Lab Than in the Field? *IEEE Transactions on Neural Systems and Rehabilitation Engineering : A Publication of the IEEE Engineering in Medicine and Biology Society*, 29, 173–183. <https://doi.org/10.1109/TNSRE.2020.3041906>
- de Bock, S., Rossini, M., Lefeber, D., Rodriguez-Guerrero, C., Geeroms, J., Meeusen, R., & de Pauw, K. (2022). An Occupational Shoulder Exoskeleton

- Reduces Muscle Activity and Fatigue During Overhead Work. *IEEE Transactions on Bio-Medical Engineering*, 69(10), 3008–3020. <https://doi.org/10.1109/TBME.2022.3159094>
- de Looze, M. P., Bosch, T., Krause, F., Stadler, K. S., & O'Sullivan, L. W. (2016). Exoskeletons for industrial application and their potential effects on physical work load. *Ergonomics*, 59(5), 671–681. <https://doi.org/10.1080/00140139.2015.1081988>
- de Ruiter, C. J., Goudsmit, J. F. A., van Tricht, J. A., & Haan, A. de (2007). The isometric torque at which knee-extensor muscle reoxygenation stops. *Medicine and Science in Sports and Exercise*, 39(3), 443–453. <https://doi.org/10.1249/mss.0b013e31802dd3cc>
- de Vries, A. W., & de Looze, M. P. (2019). The Effect of Arm Support Exoskeletons in Realistic Work Activities: A Review Study. *Journal of Ergonomics*. Advance online publication. <https://doi.org/10.35248/2165-7556.19.9.255>
- de Vries, A. W., Krause, F., & de Looze, M. P. (2021). The effectivity of a passive arm support exoskeleton in reducing muscle activation and perceived exertion during plastering activities. *Ergonomics*, 64(6), 712–721. <https://doi.org/10.1080/00140139.2020.1868581>
- de Vries, A. W., Murphy, M., Könemann, R., Kingma, I., & de Looze, M. P. (2019). The Amount of Support Provided by a Passive Arm Support Exoskeleton in a Range of Elevated Arm Postures. *IIE Transactions on Occupational Ergonomics and Human Factors*, 7(3-4), 311–321. <https://doi.org/10.1080/24725838.2019.1669736>
- Del Ferraro, S., Falcone, T., Ranavolo, A., & Molinaro, V. (2020). The Effects of Upper-Body Exoskeletons on Human Metabolic Cost and Thermal Response during Work Tasks-A Systematic Review. *International Journal of Environmental Research and Public Health*, 17(20). <https://doi.org/10.3390/ijerph17207374>
- Demura, S., Yamaji, S., & Yamada, T. (2007). Comparison of oxygenation kinetics measured by different placements of the NIRS probe during sustained isometric gripping. *Sport Sciences for Health*, 2(2), 71–75. <https://doi.org/10.1007/s11332-007-0042-2>
- Desbrosses, K., Schwartz, M., & Theurel, J. (2021). Evaluation of two upper-limb exoskeletons during overhead work: Influence of exoskeleton design and load on muscular adaptations and balance regulation. *European Journal of Applied Physiology*. Advance online publication. <https://doi.org/10.1007/s00421-021-04747-9>
- Deutsche Gesetzliche Unfallversicherung (2023). DGUV Information 208-033: Muskel-Skelett-Belastungen - erkennen und beurteilen.
- Ebaugh, D. D., McClure, P. W., & Karduna, A. R. (2006). Effects of shoulder muscle fatigue caused by repetitive overhead activities on scapulothoracic and glenohumeral kinematics. *Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology*, 16(3), 224–235. <https://doi.org/10.1016/j.jelekin.2005.06.015>
- Echtermeyer, V., & Sangmeister, M. (1996). *Praxisbuch Schulter: Erkrankungen und Verletzungen systematisch diagnostizieren, therapieren, begutachten ; 5 Tabellen*. Thieme.
- Ehrig, R. M., & Heller, M. O. (2019). On intrinsic equivalences of the finite helical axis, the instantaneous helical axis, and the SARA approach. A mathematical perspective. *Journal of Biomechanics*, 84, 4–10. <https://doi.org/10.1016/j.jbiomech.2018.12.034>

- Ehrig, R. M., Taylor, W. R., Duda, G. N., & Heller, M. O. (2006). A survey of formal methods for determining the centre of rotation of ball joints. *Journal of Biomechanics*, 39(15), 2798–2809. <https://doi.org/10.1016/j.jbiomech.2005.10.002>
- Ehrig, R. M., Taylor, W. R., Duda, G. N., & Heller, M. O. (2007). A survey of formal methods for determining functional joint axes. *Journal of Biomechanics*, 40(10), 2150–2157. <https://doi.org/10.1016/j.jbiomech.2006.10.026>
- Elcadi, G. H., Forsman, M., & Crenshaw, A. G. (2011). The relationship between oxygenation and myoelectric activity in the forearm and shoulder muscles of males and females. *European Journal of Applied Physiology*, 111(4), 647–658. <https://doi.org/10.1007/s00421-010-1688-z>
- European Agency for Safety and Health at Work (2019). Work-related musculoskeletal disorders: prevalence, costs and demographics in the EU.
- Ferguson, S. A., Allread, W. G., Le, P., Rose, J., & Marras, W. S. (2013). Shoulder muscle fatigue during repetitive tasks as measured by electromyography and near-infrared spectroscopy. *Human Factors*, 55(6), 1077–1087. <https://doi.org/10.1177/0018720813482328>
- Flatow, E. L., Soslowsky, L. J., Ticker, J. B., Pawluk, R. J., Hepler, M., Ark, J., Mow, V. C., & Bigliani, L. U. (1994). Excursion of the rotator cuff under the acromion. Patterns of subacromial contact. *The American Journal of Sports Medicine*, 22(6), 779–788. <https://doi.org/10.1177/036354659402200609>
- Forde, M. S., Punnett, L., & Wegman, D. H. (2002). Pathomechanisms of work-related musculoskeletal disorders: Conceptual issues. *Ergonomics*, 45(9), 619–630. <https://doi.org/10.1080/00140130210153487>
- Fritzsche, L., Galibarov, P. E., Gärtner, C., Bornmann, J., Damsgaard, M., Wall, R., Schirrmester, B., Gonzalez-Vargas, J., Pucci, D., Maurice, P., Ivaldi, S., & Babič, J. (2021). Assessing the efficiency of exoskeletons in physical strain reduction by biomechanical simulation with AnyBody Modeling System. *Wearable Technologies*, 2, e6. <https://doi.org/10.1017/wtc.2021.5>
- Gillette, J. C., Saadat, S., & Butler, T. (2020). Exoskeleton Usage, Arm Posture, and Tool Weight Changes Shoulder Fatigue Risk. *American Society of Biomechanics, 44th Annual Meeting*.
- Gillette, J. C., Saadat, S., & Butler, T. (2022). Electromyography-based fatigue assessment of an upper body exoskeleton during automotive assembly. *Wearable Technologies*, 3, e23. <https://doi.org/10.1017/wtc.2022.20>
- Gillette, J. C., & Stephenson, M. L. (2019). Electromyographic Assessment of a Shoulder Support Exoskeleton During on-Site Job Tasks. *IIEE Transactions on Occupational Ergonomics and Human Factors*, 7(3-4), 302–310. <https://doi.org/10.1080/24725838.2019.1665596>
- Gohlke, F. (2002). *Schulter. Orthopädie und Orthopädische Chirurgie*. Georg Thieme Verlag. <https://eref.thieme.de/ebooks/702209> <https://doi.org/10.1055/b-002-7273>
- Gopura, R. A. R. C., & Kiguchi, K. (2009). Mechanical designs of active upper-limb exoskeleton robots: State-of-the-art and design difficulties. In *2009 IEEE International Conference on Rehabilitation Robotics* (pp. 178–187). IEEE. <https://doi.org/10.1109/ICORR.2009.5209630>
- Gray, H., & Lewis, W. H. (1918). *Anatomy of the human body* (20th ed.). Lea & Febiger.
- Grieve, J. R., & Dickerson, C. R. (2008). Overhead work: Identification of evidence-based exposure guidelines. *Occupational Ergonomics*, 8(1), 53–66. <https://doi.org/10.3233/OER-2008-8105>

- Grobe, T., & Bessel, S. (2024). Gesundheitsreport 2024: Arbeitsunfähigkeiten.
- Hagberg, M. (1981). Electromyographic signs of shoulder muscular fatigue in two elevated arm positions. *American Journal of Physical Medicine*, 60(3), 111–121.
- Hagberg, M. (1984). Occupational musculoskeletal stress and disorders of the neck and shoulder: A review of possible pathophysiology. *International Archives of Occupational and Environmental Health*, 53(3), 269–278. <https://doi.org/10.1007/BF00398820>
- Hamamatsu. (2012). *NIRO-200NX: Bedienungsanleitung*.
- Hanavan, E. (1964). A Mathematical Model of the Human Body. Amrl-TR-64-102. *AMRL-TR. Aerospace Medical Research Laboratories (U.S.)*, 1–149.
- Hartmann, V. N., Rinaldi, D. d. M., Taira, C., & Forner-Cordero, A. (2021). Industrial Upper-Limb Exoskeleton Characterization: Paving the Way to New Standards for Benchmarking. *Machines*, 9(12), 362. <https://doi.org/10.3390/machines9120362>
- Herberts, P., & Kadefors, R. (1976). A study of painful shoulder in welders. *Acta Orthopaedica Scandinavica*, 47(4), 381–387. <https://doi.org/10.3109/17453677608988705>
- Herberts, P., Kadefors, R., Högfors, C., & Sigholm, G. (1984). Shoulder pain and heavy manual labor. *Clinical Orthopaedics and Related Research*(191), 166–178.
- Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology*, 10(5), 361–374. [https://doi.org/10.1016/s1050-6411\(00\)00027-4](https://doi.org/10.1016/s1050-6411(00)00027-4)
- Hisaeda, H. O., Shinohara, M., Kouzaki, M., & Fukunaga, T. (2001). Effect of local blood circulation and absolute torque on muscle endurance at two different knee-joint angles in humans. *European Journal of Applied Physiology*, 86(1), 17–23. <https://doi.org/10.1007/s004210100497>
- Iranzo, S., Piedrabuena, A., Iordanov, D., Martinez-Iranzo, U., & Belda-Lois, J.-M. (2020). Ergonomics assessment of passive upper-limb exoskeletons in an automotive assembly plant. *Applied Ergonomics*, 87, 103120. <https://doi.org/10.1016/j.apergo.2020.103120>
- Jäger, M. (2018). Extended compilation of autopsy-material measurements on lumbar ultimate compressive strength for deriving reference values in ergonomic work design: The Revised Dortmund Recommendations. *EXCLI Journal*, 17, 362–385. <https://doi.org/10.17179/excli2018-1206>
- Järvholm, U., Palmerud, G., Herberts, P., Högfors, C., & Kadefors, R. (1989). Intramuscular Pressure and Electromyography in the Supraspinatus Muscle at Shoulder Abduction. *Clinical Orthopaedics and Related Research*.
- Järvholm, U., Palmerud, G., Kadefors, R., & Herberts, P. (1991b). The effect of arm support on supraspinatus muscle load during simulated assembly work and welding. *Ergonomics*, 34(1), 57–66. <https://doi.org/10.1080/00140139108967288>
- Järvholm, U., Palmerud, G., Karlsson, D., Herberts, P., & Kadefors, R. (1991a). Intramuscular pressure and electromyography in four shoulder muscles. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society*, 9(4), 609–619. <https://doi.org/10.1002/jor.1100090418>
- Järvholm, U., Palmerud, G., Styf, J., Herberts, P., & Kadefors, R. (1988a). Intramuscular Pressure in the Supraspinatus Muscle. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society*.

- Järvholm, U., Styf, J., Suurkula, M., & Herberts, P. (1988b). Intramuscular pressure and muscle blood flow in supraspinatus. *European Journal of Applied Physiology*, 58(3), 219–224. <https://doi.org/10.1007/BF00417252>
- Johns, J., Schultes, I., Heinrich, K., Potthast, W., & Glitsch, U. (2024). Biomechanical analysis of different back-supporting exoskeletons regarding musculoskeletal loading during lifting and holding. *Journal of Biomechanics*, 168, 112125. <https://doi.org/10.1016/j.jbiomech.2024.112125>
- Jonsson, B. (1982). Measurement and evaluation of local muscular strain in the shoulder during constrained work. *Journal of Human Ergology*, 11(1), 73–88.
- Jorgensen, M. J., Hakansson, N. A., & Desai, J. (2022b). The impact of passive shoulder exoskeletons during simulated aircraft manufacturing sealing tasks. *International Journal of Industrial Ergonomics*, 91, 103337. <https://doi.org/10.1016/j.ergon.2022.103337>
- Jorgensen, M. J., Hakansson, N. A., & Desai, J. (2022a). Influence of different passive shoulder exoskeletons on shoulder and torso muscle activation during simulated horizontal and vertical aircraft squeeze riveting tasks. *Applied Ergonomics*, 104, 103822. <https://doi.org/10.1016/j.apergo.2022.103822>
- Kadefors, R., Petersén, I., & Herberts, P. (1976). Muscular reaction to welding work: An electromyographic investigation. *Ergonomics*, 19(5), 543–548. <https://doi.org/10.1080/00140137608931568>
- Kim, S., & Nussbaum, M. A. (2019). A Follow-Up Study of the Effects of An Arm Support Exoskeleton on Physical Demands and Task Performance During Simulated Overhead Work. *IIE Transactions on Occupational Ergonomics and Human Factors*, 7(3-4), 163–174. <https://doi.org/10.1080/24725838.2018.1551255>
- Kim, S., Nussbaum, M. A., Mokhlespour Esfahani, M. I., Alemi, M. M., Alabdulkarim, S., & Rashedi, E. (2018). Assessing the influence of a passive, upper extremity exoskeletal vest for tasks requiring arm elevation: Part I - "Expected" effects on discomfort, shoulder muscle activity, and work task performance. *Applied Ergonomics*, 70, 315–322. <https://doi.org/10.1016/j.apergo.2018.02.025>
- Klemm, A.-K., Knieps, F., & Pfaff, H. (2024, December). Spurwechsel Prävention, 2024.
- Kong, Y.-K., Kim, J. H., Shim, H.-H., Shim, J.-W., Park, S.-S., & Choi, K.-H. (2023). Efficacy of passive upper-limb exoskeletons in reducing musculoskeletal load associated with overhead tasks. *Applied Ergonomics*, 109, 103965. <https://doi.org/10.1016/j.apergo.2023.103965>
- Lee, S.-B., & An, K.-N. (2002). Dynamic glenohumeral stability provided by three heads of the deltoid muscle. *Clinical Orthopaedics and Related Research*(400), 40–47. <https://doi.org/10.1097/00003086-200207000-00006>
- Lin, L., Niwayama, M., Shiga, T., Kudo, N., Takahashi, M., & Yamamoto, K. (2000). Influence of a fat on muscle oxygenation measurement using near-IR spectroscopy: Quantitative analysis based on two-layered phantom experiments and Monte Carlo simulation. *Frontiers of Medical and Biological Engineering : The International Journal of the Japan Society of Medical Electronics and Biological Engineering*, 10(1), 43–58. <https://doi.org/10.1163/15685570052061531>
- Lowe, B. D., Billotte, W. G., & Peterson, D. R. (2019). Astm F48 Formation and Standards for Industrial Exoskeletons and Exosuits. *IIE Transactions on Occupational Ergonomics and Human Factors*, 7. <https://doi.org/10.1080/24725838.2019.1579769>

- Maciukiewicz, J. M., Cudlip, A. C., Chopp-Hurley, J. N., & Dickerson, C. R. (2016). Effects of overhead work configuration on muscle activity during a simulated drilling task. *Applied Ergonomics*, 53 Pt A, 10–16. <https://doi.org/10.1016/j.apergo.2015.08.005>
- Mänttari, S., Rauttola, A.-P., Halonen, J., Karkulehto, J., Säynäjäkangas, P., & Oksa, J. (2024). Effects of upper-limb exoskeleton on muscle activity in tasks requiring arm elevation: Part II - In-field experiments in construction industry settings. *Work (Reading, Mass.)*, 79(2), 753–763. <https://doi.org/10.3233/WOR-230218>
- McFarland, T. C., & Fischer, S. L. (2019). Considerations for Industrial Use: A Systematic Review of the Impact of Active and Passive Upper Limb Exoskeletons on Physical Exposures. *IIE Transactions on Occupational Ergonomics and Human Factors*, 7(3-4), 322–347. <https://doi.org/10.1080/24725838.2019.1684399>
- McFarland, T. C., McDonald, A. C., Whittaker, R. L., Callaghan, J. P., & Dickerson, C. R. (2022). Level of exoskeleton support influences shoulder elevation, external rotation and forearm pronation during simulated work tasks in females. *Applied Ergonomics*, 98, 103591. <https://doi.org/10.1016/j.apergo.2021.103591>
- Mehta, J. P., Lavender, S. A., & Jagacinski, R. J. (2014). Physiological and biomechanical responses to a prolonged repetitive asymmetric lifting activity. *Ergonomics*, 57(4), 575–588. <https://doi.org/10.1080/00140139.2014.887788>
- Michaud, B., Jackson, M., Arndt, A., Lundberg, A., & Begon, M. (2016). Determining in vivo sternoclavicular, acromioclavicular and glenohumeral joint centre locations from skin markers, CT-scans and intracortical pins: A comparison study. *Medical Engineering & Physics*, 38(3), 290–296. <https://doi.org/10.1016/j.medengphy.2015.12.004>
- Miranda, H., Viikari-Juntura, E., Heistaro, S., Heliövaara, M., & Riihimäki, H. (2005). A population study on differences in the determinants of a specific shoulder disorder versus nonspecific shoulder pain without clinical findings. *American Journal of Epidemiology*, 161(9), 847–855. <https://doi.org/10.1093/aje/kwi112>
- Moeller, T., Krell-Roesch, J., Woll, A., & Stein, T. (2022). Effects of Upper-Limb Exoskeletons Designed for Use in the Working Environment-A Literature Review. *Frontiers in Robotics and AI*, 9, 858893. <https://doi.org/10.3389/frobt.2022.858893>
- Musso, M., Oliveira, A. S., & Bai, S. (2022). Modeling of a Non-Rigid Passive Exoskeleton-Mathematical Description and Musculoskeletal Simulations. *Robotics*, 11(6), 147. <https://doi.org/10.3390/robotics11060147>
- Musso, M., Oliveira, A. S., & Bai, S. (2024). Influence of an upper limb exoskeleton on muscle activity during various construction and manufacturing tasks. *Applied Ergonomics*, 114, 104158. <https://doi.org/10.1016/j.apergo.2023.104158>
- Ojelade, A., Morris, W., Kim, S., Kelson, D., Srinivasan, D., Smets, M., & Nussbaum, M. A. (2023). Three passive arm-support exoskeletons have inconsistent effects on muscle activity, posture, and perceived exertion during diverse simulated pseudo-static overhead nutrunning tasks. *Applied Ergonomics*, 110, 104015. <https://doi.org/10.1016/j.apergo.2023.104015>
- Openclipart. *Man body silhouette* (Openclipart). Openclipart. <https://openclipart.org/detail/314198/man-body-silhouette>. Retrieved May 28, 2025

- Otten, B. M., Weidner, R., & Argubi-Wollesen, A. (2018). Evaluation of a Novel Active Exoskeleton for Tasks at or Above Head Level. *IEEE Robotics and Automation Letters*, 3(3), 2408–2415. <https://doi.org/10.1109/LRA.2018.2812905>
- Pacifico, I., Aprigliano, F., Parri, A., Cannillo, G., Melandri, I., Sabatini, A. M., Violante, F. S., Molteni, F., Giovacchini, F., Vitiello, N., & Crea, S. (2023). Evaluation of a spring-loaded upper-limb exoskeleton in cleaning activities. *Applied Ergonomics*, 106, 103877. <https://doi.org/10.1016/j.apergo.2022.103877>
- Pacifico, I., Parri, A., Taglione, S., Sabatini, A. M., Violante, F. S., Molteni, F., Giovacchini, F., Vitiello, N., & Crea, S. (2022). Exoskeletons for workers: A case series study in an enclosures production line. *Applied Ergonomics*, 101, 103679. <https://doi.org/10.1016/j.apergo.2022.103679>
- Pacifico, I., Scano, A., Guanziroli, E., Moise, M., Morelli, L., Chiavenna, A., Romo, D., Spada, S., Colombina, G., Molteni, F., Giovacchini, F., Vitiello, N., & Crea, S. (2020). An Experimental Evaluation of the Proto-MATE: A Novel Ergonomic Upper-Limb Exoskeleton to Reduce Workers' Physical Strain. *IEEE Robotics & Automation Magazine*, 27(1), 54–65. <https://doi.org/10.1109/MRA.2019.2954105>
- Page, P. (2011). Shoulder muscle imbalance and subacromial impingement syndrome in overhead athletes. *International Journal of Sports Physical Therapy*, 6(1), 51–58.
- Palmerud, G., Forsman, M., Sporrang, H., Herberts, P., & Kadefors, R. (2000). Intramuscular pressure of the infra- and supraspinatus muscles in relation to hand load and arm posture. *European Journal of Applied Physiology*, 83(2-3), 223–230. <https://doi.org/10.1007/s004210000282>
- Peat, M. (1986). Functional Anatomy of the Shoulder Complex. *Physical Therapy*, 66(12), 1855–1865. <https://doi.org/10.1093/ptj/66.12.1855>
- Pentenga, H. M., Coenen, P., Huysmans, M. A., & Speklé, E. M. (2024). The effects of working with a passive arm-support exoskeleton on objective and self-reported measures during field tasks - a randomised cross-over study. *Ergonomics*, 1–17. <https://doi.org/10.1080/00140139.2024.2392785>
- Perrey, S., Thedon, T., & Bringard, A. (2010a). Application of near-infrared spectroscopy in preventing work-related musculoskeletal disorders: Brief review. *International Journal of Industrial Ergonomics*, 40(2), 180–184. <https://doi.org/10.1016/j.ergon.2009.01.009>
- Perrey, S., Thedon, T., & Rupp, T. (2010b). NIRS in ergonomics: Its application in industry for promotion of health and human performance at work. *International Journal of Industrial Ergonomics*, 40(2), 185–189. <https://doi.org/10.1016/j.ergon.2008.11.002>
- Pinho, J. P., & Forner-Cordero, A. (2022). Shoulder muscle activity and perceived comfort of industry workers using a commercial upper limb exoskeleton for simulated tasks. *Applied Ergonomics*, 101, 103718. <https://doi.org/10.1016/j.apergo.2022.103718>
- Praagman, M., Stokdijk, M., Veeger, H. E., & Visser, B. (2000). Predicting mechanical load of the glenohumeral joint, using net joint moments. *Clinical Biomechanics (Bristol, Avon)*, 15(5), 315–321. [https://doi.org/10.1016/s0268-0033\(99\)00087-x](https://doi.org/10.1016/s0268-0033(99)00087-x)
- Punnett, L., Fine, L. J., Keyserling, W. M., Herrin, G. D., & Chaffin, D. B. (2000). Shoulder disorders and postural stress in automobile assembly work. *Scandinavian Journal of Work, Environment & Health*, 26(4), 283–291. <https://doi.org/10.5271/sjweh.544>

- Rab, G., Petuskey, K., & Bagley, A. (2002). A method for determination of upper extremity kinematics. *Gait & Posture*, 15(2), 113–119. [https://doi.org/10.1016/s0966-6362\(01\)00155-2](https://doi.org/10.1016/s0966-6362(01)00155-2)
- Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. N. (2014). *Research methods in biomechanics* (Second edition). Human Kinetics.
- Rohrer, F. D., & Hyman, C. (1962). Blood flow in arm and finger during muscle contraction and joint position changes. *Journal of Applied Physiology (Bethesda, Md. : 1985)*, 17(5), 819–823. <https://doi.org/10.1152/jappl.1962.17.5.819>
- Schmalz, T., Schändlinger, J., Schuler, M., Bornmann, J., Schirrmeister, B., Kannenberg, A., & Ernst, M. (2019). Biomechanical and Metabolic Effectiveness of an Industrial Exoskeleton for Overhead Work. *International Journal of Environmental Research and Public Health*, 16(23). <https://doi.org/10.3390/ijerph16234792>
- Schrøder Jakobsen, L., Zee, M. de, Samani, A., Desbrosses, K., & Madeleine, P. (2023). Biomechanical changes, acceptance, and usability of a passive shoulder exoskeleton in manual material handling. A field study. *Applied Ergonomics*, 113, 104104. <https://doi.org/10.1016/j.apergo.2023.104104>
- Seidler, A., Bolm-Audorff, U., Petereit-Haack, G., Ball, E., Klupp, M., Krauss, N., & Elsner, G. (2011). Work-related lesions of the supraspinatus tendon: A case-control study. *International Archives of Occupational and Environmental Health*, 84(4), 425–433. <https://doi.org/10.1007/s00420-010-0567-6>
- Seiferheld, B. E., Frost, J., Krog, M., Skals, S., & Andersen, M. S. (2022). Biomechanical investigation of a passive upper-extremity exoskeleton for manual material handling - a computational parameter study and modelling approach. *International Journal of Human Factors Modelling and Simulation*, 7(3/4), Article 124304, 275. <https://doi.org/10.1504/IJHFMS.2022.124304>
- Singleton, M. C. (1966). Functional anatomy of the shoulder. *Physical Therapy*, 46(10), 1043–1051. <https://doi.org/10.1093/ptj/46.10.1043>
- Sobotta, J. (1909). *Atlas and textbook of human anatomy*.
- Sporrong, H., Palmerud, G., Kadefors, R., & Herberts, P. (1998). The effect of light manual precision work on shoulder muscles— an EMG analysis. *Journal of Electromyography and Kinesiology*, 8(3), 177–184. [https://doi.org/10.1016/S1050-6411\(97\)00032-1](https://doi.org/10.1016/S1050-6411(97)00032-1)
- Steinhilber, B., Luger, T., Schwenkreis, P., Middeldorf, S., Bork, H., Mann, B., Glinski, A. von, Schildhauer, T. A., Weiler, S., Schmauder, M., Heinrich, K., Winter, G., Schnalke, G., Frener, P., Schick, R., Wischniewski, S., & Jäger, M. (2020). The use of exoskeletons in the occupational context for primary, secondary, and tertiary prevention of work-related musculoskeletal complaints. *IIE Transactions on Occupational Ergonomics and Human Factors*, 8(3), 132–144. <https://doi.org/10.1080/24725838.2020.1844344>
- Sylla, N., Bonnet, V., Colledani, F., & Fraisse, P. (2014). Ergonomic contribution of ABLE exoskeleton in automotive industry. *International Journal of Industrial Ergonomics*, 44(4), 475–481. <https://doi.org/10.1016/j.ergon.2014.03.008>
- Taylor, W. R., Kornaropoulos, E. I., Duda, G. N., Krutzenstein, S., Ehrig, R. M., Arampatzis, A., & Heller, M. O. (2010). Repeatability and reproducibility of OSSCA, a functional approach for assessing the kinematics of the lower limb. *Gait & Posture*, 32(2), 231–236. <https://doi.org/10.1016/j.gaitpost.2010.05.005>
- Terry, G. C., & Chopp, T. M. (2000). Functional anatomy of the shoulder. *Journal of Athletic Training*, 35(3), 248–255.

- Tetteh, E., Hallbeck, M. S., & Mirka, G. A. (2022). Effects of passive exoskeleton support on EMG measures of the neck, shoulder and trunk muscles while holding simulated surgical postures and performing a simulated surgical procedure. *Applied Ergonomics*, 100, 103646. <https://doi.org/10.1016/j.apergo.2021.103646>
- Theurel, J., & Desbrosses, K. (2019). Occupational Exoskeletons: Overview of Their Benefits and Limitations in Preventing Work-Related Musculoskeletal Disorders. *IIE Transactions on Occupational Ergonomics and Human Factors*, 7(3-4), 264–280. <https://doi.org/10.1080/24725838.2019.1638331>
- Theurel, J., Desbrosses, K., Roux, T., & Savescu, A. (2018). Physiological consequences of using an upper limb exoskeleton during manual handling tasks. *Applied Ergonomics*, 67, 211–217. <https://doi.org/10.1016/j.apergo.2017.10.008>
- Toxiri, S., Koopman, A. S., Lazzaroni, M., Ortiz, J., Power, V., de Looze, M. P., O'Sullivan, L. W., & Caldwell, D. G. (2018). Rationale, Implementation and Evaluation of Assistive Strategies for an Active Back-Support Exoskeleton. *Frontiers in Robotics and AI*, 5, 53. <https://doi.org/10.3389/frobt.2018.00053>
- van der Have, A., Rossini, M., Rodriguez-Guerrero, C., van Rossum, S., & Jonkers, I. (2022). The Exo4Work shoulder exoskeleton effectively reduces muscle and joint loading during simulated occupational tasks above shoulder height. *Applied Ergonomics*, 103, 103800. <https://doi.org/10.1016/j.apergo.2022.103800>
- van Engelhoven, L., Poon, N., Kazerooni, H., Rempel, D., Barr, A., & Harris-Adamson, C. (2019). Experimental Evaluation of a Shoulder-Support Exoskeleton for Overhead Work: Influences of Peak Torque Amplitude, Task, and Tool Mass. *IIE Transactions on Occupational Ergonomics and Human Factors*, 3, 1–14. <https://doi.org/10.1080/24725838.2019.1637799>
- Watterworth, M. W. B., Dharmaputra, R., Porto, R., Cort, J. A., & La Delfa, N. J. (2023). Equations for estimating the static supportive torque provided by upper-limb exoskeletons. *Applied Ergonomics*, 113, 104092. <https://doi.org/10.1016/j.apergo.2023.104092>
- Weston, E. B., Alizadeh, M., Hani, H., Knapik, G. G., Souchereau, R. A., & Marras, W. S. (2022). A physiological and biomechanical investigation of three passive upper-extremity exoskeletons during simulated overhead work. *Ergonomics*, 65(1), 105–117. <https://doi.org/10.1080/00140139.2021.1963490>
- Weston, E. B., Alizadeh, M., Knapik, G. G., Wang, X., & Marras, W. S. (2018). Biomechanical evaluation of exoskeleton use on loading of the lumbar spine. *Applied Ergonomics*, 68, 101–108. <https://doi.org/10.1016/j.apergo.2017.11.006>
- Yang, Y., Soyemi, O. O., Landry, M. R., & Soller, B. R. (2005). Influence of a fat layer on the near infrared spectra of human muscle: Quantitative analysis based on two-layered Monte Carlo simulations and phantom experiments. *Optics Express*, 13(5), 1570–1579. <https://doi.org/10.1364/opex.13.001570>
- Zipp, P. (1982). Recommendations for the standardization of lead positions in surface electromyography. *European Journal of Applied Physiology and Occupational Physiology*, 50(1), 41–54. <https://doi.org/10.1007/BF00952243>

20. Danksagung

Ich möchte diese Gelegenheit nutzen, um mich bei all den Menschen und Institutionen zu bedanken, die mich auf dem Weg meiner Promotion am **Universitätsklinikum Hamburg-Eppendorf (UKE)** begleitet und unterstützt haben.

Herr Professor Doktor Volker Harth, Institutsdirektor des Zentralinstitut für Arbeitsmedizin und Maritime Medizin (ZfAM). Sehr geehrter Herr Professor Harth, ich möchte Ihnen sehr dafür danken, dass Sie mich als mein Doktorvater bei meiner Promotion am UKE betreut und begleitet haben. Dadurch haben Sie mir die Möglichkeit gegeben, diese spannende wissenschaftliche Thematik im Rahmen einer Promotion zu erforschen und meine Expertise und Erfahrungen in der wissenschaftlichen Praxis zu vertiefen. Ich bedanke mich auch für Ihre fachliche und persönliche Unterstützung in den letzten Jahren, die mir in vielen Phasen sehr geholfen hat, in meiner Promotion weiterzukommen. Ihr großes Fachwissen und Ihr interdisziplinärer Blick auf die Thematik haben mich dabei immer wieder inspiriert, mein Wissen – vor allem im medizinischen Bereich – zu vertiefen, um die komplexen Aspekte meines Forschungsthemas besser und tiefergehend zu verstehen.

Herr Professor Doktor Albert Nienhaus, Stellvertretende Institutsdirektor des Instituts für Versorgungsforschung in der Dermatologie und bei Pflegeberufen (IVDP). Sehr geehrter Herr Professor Nienhaus, ich bin Ihnen sehr dankbar dafür, dass Sie mich als ein Mitglied meines Thesis-Komitees bei meiner Promotion begleitet und mich bei meinen Forschungsvorhaben jederzeit unterstützt haben. Ich möchte mich bei Ihnen für Ihr großes Interesse an meinem Forschungsthema, Ihr konstruktives Feedback zu meinen Ergebnissen und Ihren hohen wissenschaftlichen Input bedanken, was mich bei der Fortführung meiner Forschungsarbeiten immer gestärkt hat. Zudem möchte ich Ihnen dafür danken, dass Sie mich im Rahmen meiner Promotion und auf meinem beruflichen Weg – gerade im vergangenen Jahr – immer gefördert und unterstützt haben.

Mein besonderer Dank gilt auch der **Deutschen Gesetzlichen Unfallversicherung e.V. (DGUV)**, die meine Promotion im Rahmen eines Deutschlandstipendiums großzügig gefördert hat.

Ebenso danke ich **Herrn Professor Doktor Rolf Ellegast, dem Direktor des Instituts für Arbeitsschutz (IFA) der DGUV**, für seine kontinuierliche Unterstützung und das Vertrauen, das er mir während meiner Promotionszeit entgegengebracht hat. Ein herzliches Dankeschön geht auch an **Doktor Kai Heinrich, Referatsleiter für Muskel-Skelett-Belastungen**. Lieber Kai, ich danke dir aufrichtig für die langjährige, engagierte Betreuung und Unterstützung im Rahmen meiner Dissertation. Deine fachliche und wissenschaftliche Expertise, dein konstruktiver Rat und deine stets offene Art haben mir nicht nur inhaltlich weitergeholfen, sondern auch meine wissenschaftlichen Fähigkeiten maßgeblich gefördert und geschärft. Besonders deine große Begeisterung für die Forschung und unser gemeinsames Thema haben mich immer wieder motiviert, neue Erkenntnisse zu gewinnen und komplexe Fragestellungen mit Neugier und Tiefe zu durchdringen.

Mein Dank gilt ebenso den aktuellen und ehemaligen Mitarbeitenden des IFAs, die mich während meiner Promotionszeit mit Ihrer fachlichen Kompetenz und wissenschaftlichen Unterstützung begleitet haben. Ihre Hilfe bei der Durchführung meiner Studien und bei der Ausarbeitung meiner wissenschaftlichen Arbeiten war für mich von unschätzbarem Wert: **Herrn PD Dr. Ulrich Glitsch, Herrn Jasper Johns, Herrn Omar El-Edrissi Hörmann, Frau Inga Schultes und Herrn Markus Post** aus dem Referat Muskel-Skelett-Belastungen (Abteilung 4); **Herrn Ingo Hermanns-Truxius und Herrn Robin Grießel** aus der Stabsstelle Gestaltung neuer Arbeitsformen; **Herrn Georg Nischalke-Fehn** aus dem Referat Intelligente Technische Systeme und Arbeitswelt (Abteilung 5) sowie **Frau Dr. Vera Schellewald, Herrn Lasse Hanse und Herrn Lennard Walz**.

Vor allem möchte ich mich auch bei der **Hochschule Koblenz** und dem **Fachbereich Mathematik, Informatik, Technik** dafür bedanken, dass ich während meiner Promotionszeit meine wissenschaftlichen Arbeiten in ihren Räumlichkeiten am Standort in Remagen, **dem RheinAhrCampus**, durchführen durfte.

Insbesondere möchte ich **Herrn Professor Doktor Ulrich Hartmann** danken. Lieber Uli, ich bin dir sehr dankbar für die Zeit, die ich unter deiner Leitung und Betreuung am RheinAhrCampus verbringen durfte. In all den Jahren konnte ich mich immer, auch in schwierigen Phasen, sowohl fachlich als auch persönlich jederzeit uneingeschränkt auf deine Unterstützung verlassen, sei es im Rahmen meiner Promotion oder bei weiteren Projekten, bei denen ich mit dir oder unter deiner Leitung arbeiten durfte.

Ich bin dir sehr dankbar für die ganze Zeit, die du in meine wissenschaftliche und berufliche Entwicklung investiert hast. Danke auch für dein stets offenes, ehrliches und konstruktives Feedback und dass du mir die Möglichkeit gegeben hast, von dir, deinen Erfahrungen und deinem Wissen zu lernen.

Zudem möchte ich mich bei **Frau Doktor. Anika Weber** bedanken. Liebe Anika, ich danke dir für die ganze Zeit, in der wir beide im Rahmen unserer Promotionen und weiterer Forschungsprojekte zusammenarbeiten durften. Dabei hat nicht nur unser fachlicher Austausch sehr dazu beigetragen, dass ich mich als Wissenschaftler weiterentwickelt habe. Auch unsere privaten Gespräche über alltägliche Dinge haben mir sehr dabei geholfen, neue Energie zu gewinnen.

Ich danke auch **Herrn Doktor Daniel Friemert**. Lieber Daniel, vielen Dank für deine ganze Unterstützung und deinen kritischen, aber stets positiven und konstruktiven wissenschaftlichen Input und dein Feedback. Gerade zu Beginn meiner Promotion hat mir dies sehr dabei geholfen, meinen akademischen Weg zu finden und mein Fachwissen zu vertiefen.

Ich bedanke mich außerdem bei allen Teilnehmenden meiner wissenschaftlichen Studien, ohne die diese Promotion nicht hätte durchgeführt werden können.

Ebenso danke ich der **Firma Airbus Defence and Space in Bremen** und den Mitarbeitenden **Herrn Christian Boehm, Frau Vanessa Berg, Herrn Boris Ruschkowski und Frau Madeleine Hirth** dafür, dass ich eine meiner wissenschaftlichen Studien mit den Airbus Mitarbeitenden als Teilnehmenden in einer ihrer Produktionshallen durchführen durfte. Zudem danke ich **Herrn Torsten Wagner** von der **Berufsgenossenschaft Energie Textil Elektro Medienerzeugnisse (BG ETEM)**, welcher diese wissenschaftliche Studie mit unterstützt und begleitet hat.

Zuletzt möchte ich meinen Eltern, **Frank und Regine**, und meiner Verlobten **Svenja** dafür danken, dass sich mich immer unterstützt und mir beigestanden haben, vor allem in den schwierigen Zeiten meiner Promotion. Danke – ohne euch wäre dies alles nicht möglich gewesen.

Lebenslauf
entfällt aus
datenschutzrechtlichen
Gründen

Lebenslauf
entfällt aus
datenschutzrechtlichen
Gründen

22. Eidesstattliche Versicherung

Ich versichere ausdrücklich, dass ich die Arbeit selbständig und ohne fremde Hilfe, insbesondere ohne entgeltliche Hilfe von Vermittlungs- und Beratungsdiensten, verfasst, andere als die von mir angegebenen Quellen und Hilfsmittel nicht benutzt und die aus den benutzten Werken wörtlich oder inhaltlich entnommenen Stellen einzeln nach Ausgabe (Auflage und Jahr des Erscheinens), Band und Seite des benutzten Werkes kenntlich gemacht habe. Das gilt insbesondere auch für alle Informationen aus Internetquellen.

Soweit beim Verfassen der Dissertation KI-basierte Tools („Chatbots“) verwendet wurden, versichere ich ausdrücklich, den daraus generierten Anteil deutlich kenntlich gemacht zu haben. Die „Stellungnahme des Präsidiums der Deutschen Forschungsgemeinschaft (DFG) zum Einfluss generativer Modelle für die Text- und Bilderstellung auf die Wissenschaften und das Förderhandeln der DFG“ aus September 2023 wurde dabei beachtet.

Ferner versichere ich, dass ich die Dissertation bisher nicht einem Fachvertreter an einer anderen Hochschule zur Überprüfung vorgelegt oder mich anderweitig um Zulassung zur Promotion beworben habe.

Ich erkläre mich damit einverstanden, dass meine Dissertation vom Dekanat der Medizinischen Fakultät mit einer gängigen Software zur Erkennung von Plagiaten überprüft werden kann.

Datum

Unterschrift