Using biomechanical data to explore the utility of exoskeleton intervention for work-related musculoskeletal disorders in the surgeons

by

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DEDICATION

I dedicate this dissertation work to my grandma Grace who instilled the attitude of hard work in me at a very tender age.

TABLE OF CONTENTS

Page

LIST OF FIGURES ... vi

LIST OF TABLES ... x

ACKNOWLEDGMENTS ... xii

ABSTRACT ... xiii

CHAPTER 1. GENERAL INTRODUCTION ... 1
- 1.1 Musculoskeletal Disorders in Surgeons ... 1
- 1.2 Contributory Factors to Musculoskeletal Disorders in Surgeons ... 1
- 1.3 Existing Interventions ... 3
- 1.4 Exoskeletons and Their Potential Benefit ... 6
- 1.5 Aim of Research ... 7
- References ... 8

CHAPTER 2. BACKGROUND ... 12
- 2.1 Functional Anatomy of the Neck, Low Back and Shoulders ... 12
  - 2.1.1 Neck ... 12
    - 2.1.1.2 Cervical vertebrae ... 12
    - 2.1.1.3 Muscles of the head-neck ... 13
      - 2.1.1.3.1 Neck extensors ... 13
      - 2.1.1.3.2 Neck flexors ... 14
  - 2.1.2 Low Back ... 16
  - 2.1.3 Shoulder ... 21
    - 2.1.3.1 Bony Anatomy ... 21
      - 2.1.3.1.1 Humerus ... 21
      - 2.1.3.1.2 Scapula ... 23
      - 2.1.3.1.3 Clavicle ... 24
    - 2.1.3.2 Joint articulations ... 25
      - 2.1.3.2.1 Glenohumeral joint ... 25
      - 2.1.3.2.2 Passive mechanism ... 26
    - 2.1.3.3 Dynamic stabilizing mechanism ... 30
      - 2.1.3.3.1 Deltoid muscle ... 30
      - 2.1.3.3.2 Rotator cuff ... 31
- 2.2. General Physical Ergonomic Issues in the Surgical Environment ... 34
- 2.3 Exoskeletons ... 37
  - 2.3.1 Upper Extremity Exoskeletons ... 38
    - 2.3.1.1 Levitate Airframe® ... 38
    - 2.3.1.2 Eksovest ... 39
  - 2.3.2 Trunk Exoskeletons ... 40
    - 2.3.2.1 Laevo v series exoskeleton ... 40
    - 2.3.2.2 Personal lift assist device (PLAD) ... 41

2.3.2.3 Exosuit ... 43
2.4 Occupational Exoskeletons in the Non-surgical Environment ... 44
2.4.1 Upper Extremity Exoskeletons ... 45
2.4.2 Trunk Exoskeletons ... 46
2.5 Occupational Exoskeletons in the Surgical Environment ... 48
2.5.1 Recent Studies on Exoskeletons in the Operating Room ... 50
2.5.1.1 Upper extremity exoskeleton in the operating room ... 50
2.5.1.2 Trunk Postural Support in the Operating Room ... 52
References ... 54

CHAPTER 3. INTRAOPERATIVE POSTURE AND WORKLOAD ASSESSMENT IN VASCULAR SURGERY ... 62
Abstract ... 62
3.1 Introduction ... 63
3.2 Materials and Methods ... 67
3.2.1 Participants ... 67
3.2.2 Data Collection Instrumentation ... 67
3.2.3 Experimental Tasks ... 68
3.2.4 Experimental Design ... 70
3.2.4.1 Independent variables ... 70
3.2.3.2 Dependent variables ... 70
3.2.3.3 Data processing and statistical analysis ... 70
3.3 Results ... 74
3.3.1 Summary Statistics ... 74
3.3.2 Difference Between Open and Non-Open Surgeries ... 77
3.3.3 Difference Between Surgeries With and Without Loupes ... 78
3.3.4 Difference Between Surgeries With and Without RPPE ... 79
3.3.5 Effects of the Surgical Duration ... 80
3.3.6 Correlation Between Subjective and Objective Measurements ... 81
3.4 Discussion ... 83
3.4.1 Basic Statistics and Upper Body Pain Scores ... 83
3.4.2 Difference between Open and Non-Open Surgeries ... 84
3.4.3 Difference Between Surgeries With and Without Loupes ... 84
3.4.4 Difference Between Surgeries With and Without RPPE ... 86
3.4.5 Effects of the Surgical Duration ... 86
3.4.6 Correlation between Subjective and Objective Measurements ... 87
3.4.7 Limitations ... 88
3.5 Conclusions ... 88
References ... 89

CHAPTER 4. AN ALGORITHM FOR PREDICTING THE UTILITY OF EXOSKELETONS FOR DIFFERENT SURGERIES ... 94
Abstract ... 94
4.1 Introduction ... 95
4.2 Methods ... 98
4.2.1 Segmental Postural Data ... 98

4.2.2 Time Domain Independent Variables 98
4.2.2.1 Cumulative postural fatigue risk model (CPFRM) 99
4.2.2.1.2 Biomechanical modeling 100
4.2.2.1.3 Risk coefficients 103
4.2.2.1.4 CPFRS estimation 104
4.2.3 Frequency Domain Variables 105
4.2.4 Dependent Variable 106
4.2.5 Statistical/Discriminant Analysis 106
4.3 Results 108
4.4 Discussion 111
4.5 Conclusions 114
References 115
Appendix: IRB Approval Memo 119

CHAPTER 5. EFFECTS OF PASSIVE EXOSKELETON SUPPORT ON EMG MEASURES OF NECK, SHOULDER, AND TRUNK EXTENSOR MUSCLES WHILE HOLDING SIMULATED SURGICAL POSTURES AND PERFORMING A SIMULATED VASCULAR CATHETERIZATION PROCEDURE 121
Abstract 121
5.1 Introduction 122
5.2 Methodology 125
5.2.1 Participants 125
5.2.2. Apparatus 126
5.2.2.1 EMG 126
5.2.2.2 Inertial measurement units 127
5.2.2.3 Exoskeletons 128
5.2.2.4 Subjective Survey 129
5.2.3 Experimental Design 129
5.2.4 Experimental Tasks 131
5.2.4.1 Maximum voluntary contractions 131
5.2.4.2 Static segmental postures 132
5.2.4.3 Vascular surgery simulation task 133
5.2.5 Study Description 134
5.2.6 Data Processing 136
5.2.7 Statistical Analysis 137
5.3 Results 138
5.3.1 Static Posture 138
5.3.2 Simulation Task 143
5.4 Discussion 147
5.4.1 Static Posture Task 148
5.4.2 Surgical Simulation Task 149
5.5 Conclusion 152
References 153
Appendix: IRB Approval Memo 157

CHAPTER 6. GENERAL CONCLUSION 159

LIST OF FIGURES

Page

Figure 2.1: Superior View of a Cervical Vertebra. (Gray & Lewis, 1918, image 84)................. 12

Figure 2.2: The Trapezius and Levator Scapula. (Gray & Lewis, 1918, image 409).................. 14

Figure 2.3: The Sternocleidomastoid. (Gray & Lewis, 1918, image 385).................................. 15

Figure 2.4: The Anterior Scalene Muscle. (Gray & Lewis, 1918, image 385)............................ 16

Figure 2.5: Three-Dimensional View of the Lumbar Vertebra.(Gray & Lewis, 1918, image 93).. 17

Figure 2.6: Lateral view of the lumbar spine showing the ligaments. (Gray & Lewis, 1918, Image 301).. 19

Figure 2.7: Dorsal View of the Trunk Showing the Erector Spinae Muscles. (Gray & Lewis, 1918, Image 389).. 20

Figure 2.8: Anterior View of the humerus.(Gray & Lewis, 1918, Image 208) 22

Figure 2.9: Dorsal View of the Left Scapula.(Gray & Lewis, 1918, Image 203)......................... 23

Figure 2.10: Superior and Inferior Surface of the Clavicle. (Gray & Lewis, 1918, image 200 & 201).. 25

Figure 2.11: Anterior View of the Glenohumeral Joint.(Gray & Lewis, 1918, Image 327) 26

Figure 2.12: Lateral view of the scapula showing the glenoid fossa. (Gray & Lewis, 1918, Image 205).. 27

Figure 2.13: Lateral View of the Glenoid Cavity. (Witherspoon, Smirnova, & Mciff, 2014) Figure used with permission from Wiley and Sons.. 28

Figure 2.14: Ligaments of the Glenohumeral Joint. (Gray & Lewis, 1918, Image 326).............. 29

Figure 2.15: The Deltoid Muscle of the Shoulder.(Gray & Lewis, 1918, Image 410)................. 31

Figure 2.16: Posterior View of the Shoulder Joint to show the Rotator Cuff Muscles.(Gray & Lewis, 1918, Image 412).. 33

Figure 2.17: The Levitate Airframe (Wilson, 2017) Accessed 05 May 2020. Used with permission from Levitate Technologies Inc. 39

Figure 2.18: Laevo Exoskeleton. (Koopman, et al., 2019) Figure used with permission from Elsevier 41

Figure 2.19: The Personal Lift Assist Device. (Lotz et al., 2009) Figure used with permission from Elsevier 43

Figure 2.20: The Exosuit by VT-Lowe, Inc. (Alemi, Geissinger, Simon, Chang, & Asbeck, 2019). Figure used with permission from Elsevier 44

Figure 3.1: A typical awkward posture in vascular surgery 66

Figure 3.2: IMUs worn by the surgeon. A) Upper back sensor, B) Left upper arm sensor, C) Right upper arm sensor, D) Head sensor 69

Figure 3.3: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of postural angles for upper body parts calculated from IMU data 75

Figure 3.4: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of risk scores for upper body parts calculated from IMU data (Range from 1.0-4.0). The dashed line represents the midpoint of this range. 75

Figure 3.5: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of scores from modified SURG-TLX subscales (Range from 0-20=max). The dashed line represents the midpoint of this range. 76

Figure 3.6: The percentage of the three levels of change in fatigue and pain scores for different body parts including neck, right shoulder, left shoulder, upper back, lower back, and right and left wrists/hands 77

Figure 4.1: Static biomechanical model of the head-neck segment........ 100

Figure 4.2: Static biomechanical model of the trunk 101

Figure 4.3: Static biomechanical model of the upper extremity (shoulder) 102

Figure 4.4: LDA histogram classification plot for head-neck exoskeleton recommendation ... 109

Figure 4.5: LDA histogram classification plot for trunk exoskeleton recommendation 110

Figure 4.6: LDA histogram classification plot for shoulders exoskeleton utility 110

Figure 4.7: QDA Partition plot of exoskeleton classification in the head-neck showing the quadratic decision boundary 111

Figure 5.1: Electrode placement for sEMG measurements 127

Figure 5.2: Delsys trigno avanti EMG+IMU sensor for postural angle measurement 127

Figure 5.3: Subjective fatigue survey 129

Figure 5.4: MVC measurement. left) neck extension, middle) shoulder abduction, right) trunk extension. The red arrow indicates resistance direction, the blue line indicates direction of exertion 131

Figure 5.5: Levitate (left) and Laevo (right) exoskeletons used in the static shoulder abduction and trunk flexion task 132

Figure 5.6: Height-adjustable table with taped tubes for the simulation task............................ 133

Figure 5.7: Participant performing the simulation task with the neck and trunk exoskeletons layered 134

Figure 5.8: Bar plot with error bars (95% confidence interval) showing the mean difference in %MVC between the exoskeleton conditions for NDLES during the static postures. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance. 142

Figure 5.9: Bar plot with error bars (95% confidence interval) showing the mean difference in %MVC between the exoskeleton conditions for DLES during the static postures. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance. 143

Figure 5.10: Bar plot with error bars (95% confidence interval) showing difference in ARV gradients (%MVC/min) between the exoskeleton conditions for the four back muscles. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance 144

Figure 5.11: Side by side plot of the trend of ARV averaged across all participants for the non-dominant lower thoracic erector spinae. Left: exoskeleton condition, right: no exoskeleton condition. (Window 0 = time 0, window 200 =time 25mins) 145

Figure 5.12: Side by side plot of the trend of ARV averaged across all participants for the non-dominant lumbar erector spinae. Left: exoskeleton condition, right: no exoskeleton condition (Window 0 = time 0, window 200 = time 25mins) 145

Figure 5.13: Side by side plot of the trend of ARV averaged across all participants for the dominant lumbar erector spinae. Left: exoskeleton condition, right: no exoskeleton condition. (Window 0 = time 0, window 200 = time 25mins) 146

Figure 5.14: Bar plot with error bars (95% confidence interval) showing the mean difference between the change in subjective fatigue for the body segments between the exoskeleton conditions. Positive bar height indicates exoskeleton showed an increase in comfort. 147

LIST OF TABLES

Page

Table 3.1: Joint angles and their risk-scores 72

Table 3.2: The definitions for exposed and unexposed groups as well as case and control groups for odds ratios calculations 73

Table 3.3: Least square mean (± standard error) for dependent variables with statistically significant differences between open and non-open surgeries 78

Table 3.4: Least square mean (± standard error) for dependent variables with statistically significant differences between surgeries with and without loupes 79

Table 3.5: Least square mean (± standard error) for dependent variables with statistically significant differences between surgeries with and without RPPE 79

Table 3.6: Least square mean (± standard error) for dependent variables with statistically significant differences between short and long surgeries 81

Table 3.7: Correlation (Spearman's rank correlation coefficient (Rs) and prob>|Rs|) between calculated risk scores/postural angles (angles) and change in fatigue and pain scores 82

Table 4.1: Table of postural fatigue risk coefficients and angular thresholds 103

Table 4.2: Sample CPFRS calculation 105

Table 4.3: Independent (predictor) variables and dependent (group) variable 106

Table 4.4: Significant predictor variables and model performance (MsR) for each body segment 108

Table 5.1: EMG dependent variables for static posture and simulation task 130

Table 5.2: Segmental static postural angles 133

Table 5.3: Trunk flexion angles for the vascular simulation task 134

Table 5.4: Two-way ANOVA table for static posture tasks for all eight muscles (bold* indicates a significant difference, GG indicates Greenhouse-Geisser correction) . 139

Table 5.5: Simple effects analysis of the effect of neck exoskeleton utility at respective static neck flexion angles for NDNEM %MVC with fdr correction. (bold* indicates significant difference of exoskeleton usage at these angles).................. 141

Table 5.6: Simple effects analysis of the effect of upper arm exoskeleton utility at respective static arm abduction angles for NDMD %MVC with fdr correction. (bold* indicates significant difference of exoskeleton usage at these angles) 141

Table 5.7: Simple effects analysis of the effect of trunk exoskeleton utility at respective static trunk flexion angles for NDLES %MVC (columns 1 to 4) and DLES %MVC (columns 5:8). (* indicates statistical significance of exoskeleton usage at these angles) 142

Table 5.8: Repeated measure analysis for time and frequency domain-dependent variables for simulation task comparing combined neck and trunk exos vs no exos (positive and negative **est** values in %MVC/min for ARV and Hz/sec for MDF respectively indicates beneficial effects of the exoskeleton and vice versa).......... 143

Table 5.9: Paired t-test results for change in subjective fatigue (bold with asterisk implies significant exoskeleton effect, and positive est means exoskeleton helped to reduce fatigue) 146

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ABSTRACT

This dissertation explores how passive postural support exoskeletons can be used as an ergonomic intervention to mitigate the detrimental effects of work-related musculoskeletal disorders in surgeons. While exoskeletons have shown promising results in automotive manufacturing amongst others, their benefits are yet to be realized in healthcare delivery. For exoskeletons to succeed as an ergonomic intervention in the healthcare environment, they need to be deployed in surgical cases or tasks that predispose the surgeons to the risk factors of work-related musculoskeletal disorders. For this reason, a standardized technique backed by evidence is required to deploy exoskeletons for optimum benefits.

Using vascular surgery as an example (Chapter 3), an intraoperative workload and postural demand study was conducted to identify body segments that are considerably exposed to non-ergonomic conditions. Furthermore, this study explored the types of procedural factors and adjunctive equipment that led to increased discomfort and pain while performing surgery. Inertial Measurement Units (IMUs) were used to collect segmental kinematics posture on 16 vascular surgeons who completed 47 surgeries. Furthermore, subjective pain and discomfort data were obtained before and after each surgical operation. The results from this study showed that the neck and trunk were two body segments with the highest averaged deviation angles (37.1º ±12.7º and 18.1º±6.7º respectively) for a significant percentage of operating time. This led to considerable pain and discomfort scores in those two body segments. These results indicated that the exposure to risk factors of work-related musculoskeletal disorders varies as a function of body segment and procedure type; hence exoskeletons intervention can be targeted accordingly.

Chapter 4 of this dissertation used segmental (head-neck, shoulder, and trunk) kinematics data from 30 surgeries to identify a combination of segmental kinematic variables that can be

used to predict/classify surgeries into two groups: 1) surgeries recommended for exoskeleton intervention and 2) surgeries not recommended for shoulder, trunk and/or neck exoskeleton intervention. Cumulative Postural Fatigue Risk Score (CPFRS) from a newly-developed Cumulative Postural Fatigue Risk Model (CPFRM), in addition to other time and frequency domain kinematic variables, were used as predictor variables in both quadratic and linear discriminant analyses. Stepwise variable selection was used to identify which combination of predictor variables yielded the best classification for each body segment. The model performance was assessed using the Leave-One-Out Cross-Validation (LOOCV) technique. The results showed that the newly developed CPFRS and 10th percentile neck angle were predictive of exoskeleton intervention for the neck, only CPFRS was predictive of the utility of an exoskeleton intervention for the trunk, and CPRFS, mean shoulder deviation angle, and mean frequency (a measure of static posture) were predictive of upper arm exoskeleton intervention. These results suggest that segmental kinematics can be used to develop a standardized technique to indicate exoskeleton interventions.

The final study (Chapter 5) in this dissertation investigated the potential benefit of using exoskeletons on physical demand using electromyography. The study was divided into two phases. The first phase tested the effect of individual segmental (neck, upper extremity, and trunk) exoskeleton on static postures (neck flexion, shoulder deviation, and trunk flexion) typically assumed by surgeons. The second phase was a 30-minute catheter insertion simulation task of five trunk flexion postures typical in vascular surgery. For the first phase, the results showed positive effects of the exoskeleton at reducing the muscle activity in the lumbar extensor muscles, the medial deltoids, and the non-dominant neck extensor muscles. However, significant interactions between the exoskeleton and postural angle implies that exoskeletons may be particularly effective at specific body segment postures. The results from the extended simulation showed that over the 25-min period, the exoskeleton significantly reduced the gradient or slope of the average rectified EMG values (+1.365%MVC/min vs. +0.769%MVC/min for NDLES, $p = 0.0108$: +1.377%MVC/min vs. +0.770%MVC/min for DLES, $p = 0.0196$) for the bilateral lumbar extensors. This reduction in EMG gradient was reflected in the significant decrease (7.34 vs. 4.3) in subjective fatigue on the visual analog scale.

Collectively, the results from the three studies (Chapters 3-5) show that for successful implementation of exoskeleton interventions, body segments exposed significantly exposed to the causative factors of work-related musculoskeletal disorders need to be targeted. This requires the implementation of a standardized technique to identify such body segments, which will in-turn lead to a reduction on the biomechanical demand on such body segments and improvement in pain and fatigue.

CHAPTER 1. GENERAL INTRODUCTION

1.1 Musculoskeletal Disorders in Surgeons

Recent studies have shown that the surgeon population and healthcare systems suffer from symptoms of work-related musculoskeletal disorders (WRMSD). For example, a nationwide survey by the American Society of Vascular Surgeons (Wohlauer et al., 2019) of 775 vascular surgeons showed that the majority experienced somewhat intense pain (4.4 ±2.3 out of 10 on Borg CR10 scale). The pain was predominantly experienced in the neck (45%) and low back (39%). Another survey reported a high prevalence of WRMSDs in the neck (82.9%), shoulder (57.8%), low and upper back (68.1% & 52.6%) in a population of general surgeons (Szeto et al., 2009). These symptoms have been shown to reduce surgeon productivity due to missed work (Davila et al., 2019) and can lead to surgeon burnout (Davila et al., 2019; De Hert, 2020; Dimou, Eckelbarger, & Riall, 2016). Moreover, the cost associated with managing WRMSDs in the healthcare sector is enormous, placing institutions under immense financial burden. In light of the anticipated rise in the need for surgeons in the upcoming decade (BLS, 2018), addressing musculoskeletal disorders in surgeons is vital to sustaining the number and quality of surgical services to be delivered.

1.2 Contributory Factors to Musculoskeletal Disorders in Surgeons

The high prevalence of WRMSDs amongst surgeons can be attributed to workplace risk factors: awkward postures, repetitive exertions, long hours of task performance (exposure time), and the usage of adjunctive equipment (Seagull, 2012; Yang et al., 2020). For example, Yurteri-Kaplan et al. (2018) quantified the duration and frequency of awkward postures between attending and assistant vaginal surgeon while performing procedures. Their results showed a high duration and frequency of lateral trunk bending, neck flexion, and shoulder abduction.

While both the attending and assistant surgeon assumed significant awkward postures, the assistant surgeons spent a significantly higher amount of time in such postures (mean 4% (7.4 min ± 6.8) vs. 2% (2.9 min ± 3.6) of operating time in trunk flexion, 11% (21.4 min ± 26.4) vs. 7% (11.4 min ± 8.3) lateral bending, 41% (71.3 min ± 49.1) vs. 27% (47.2 min ± 29.5) in neck deviation, and 21% (35.1 min ± 36.7) vs. 14% (26.2 min ± 30.4) in left shoulder deviation). Furthermore, a recent study by Yang et al. (2020) used a combination of objective and subjective techniques to assess the physical stressors that predispose surgeons to WRMSD. The results from their objective real-time tracking of neck and trunk posture correlated strongly with the pain and discomfort reported in those two body segments. Furthermore, they found that adjunctive equipment such as loupes tend to induce awkward neck flexion, exacerbating pain in the neck.

The widespread adaptation of minimally invasive surgical techniques (e.g., laparoscopic surgery) as a replacement for some open procedures also poses significant physical risks to surgeons. While these procedures significantly benefit patients in terms of short recovery times and good aesthetic outcomes, the associated awkward segmental postures lead to muscle pain and discomfort. A typical example is laparoscopic surgery, which has been shown to induce significant back, shoulder, and neck static postures (Berguer, Rab, Alarcon, & Chung, 1997; Nguyen et al., 2002). Such static postures reduce blood flow and accelerate fatigue development on those body parts. Furthermore, laparoscopic surgery induces significant shoulder abduction in a static posture. Also, laparoscopic tools tend to induce awkward upper extremity postures, leading to significant fatigue and pain development (Nguyen et al., 2002; Szeto et al., 2012). It is no wonder that a recent survey of laparoscopic minimally invasive surgeons entitled: "Patients Benefit While Surgeons Suffer: An impending Epidemic" reported that of the 272 laparoscopic surgeon participants, 86.9% of them experienced physical pain or discomfort (Park, Lee, Seagull, Meenaghan, & Dexter, 2010) and complaints of such pains are predominant in the neck, low back, and shoulder (Janki, Mulder, IJzermans, & Tran, 2017). Interestingly, these symptoms were weakly correlated with the surgeon's age of length of practice, making laparoscopic surgery a rife field for ergonomic interventions.

1.3 Existing Interventions

While not extensive, some attempts have been made to improve ergonomics in the operating room. Broadly, these interventions are designed to reduce the strain/ pain and discomfort associated with fatigue build-up during surgical task performance. One such intervention is interoperative breaks with or without active stretching routines, also known as work-rest cycles. As the name work-rest implies, this intervention is designed to reduce the continuous exposure time to the risk factors of musculoskeletal disorders such as awkward postures by allowing intermittent breaks during the surgical operation. This strategy has shown improved musculoskeletal discomfort surveys and even increased productivity in non-surgical (data entry) fields (Galinsky et al., 2000; Galinsky et al., 2007). Recently, a few studies have highlighted the effect of such breaks in the operating room. For instance, in a randomized clinical trial, comparing a conventional and an intermittent laparoscopic technique for releasing pneumoperitoneum, Engelmann et al. (2011) reported significantly reduced musculoskeletal pain and strain scores in the intermittent technique, compared to the conventional (0.5 vs. 2.5 for neck, 0.5 vs. 1.7 for arms, 0.6 vs. 2.8 for spine and 0.5 vs. 1.5 for knees). Furthermore, in a multi-hospital institution study, interoperative microbreaks with standardized exercise routines significantly reduced discomfort in the shoulder and hands but not the neck and back even though the intervention was targeted at all four body segments (Hallbeck et al., 2017). Surprisingly, both (Hallbeck et al., 2017) and (Engelmann et al., 2011) reported no significant increase in operating time as a result of the additional resting periods. A similar reduction in muscular fatigue has been reported in other studies with an added advantage of improved task precision (Dorion & Darveau, 2013) and also delayed pain sensation in neck and shoulder muscles (Vijendren et al., 2018). While interoperative microbreaks have shown positive results in fatigue and pain reduction, certain inherent limitations need to be addressed. For example, microbreaks can interrupt the surgical workflow, similar to the error-increasing non-routine events (Blocker et al., 2013). This technique might not be ideal for life-threatening emergency surgeries. Moreover, as reported by (Hallbeck et al., 2017), the microbreak intervention did not positively reduce discomfort in the neck and back as intended. Hence further research is needed to design targeted work-rest routines optimally.

Another intervention technique is the reconfiguration of the layout of equipment in the operating room. This type of intervention is predominant in laparoscopic minimally invasive procedures in which the surgeons' movement is restricted while looking at the monitor screen, leading to musculoskeletal fatigue development from static neck, back, and shoulder postures (Berguer, Forkey, & Smith, 1999). A typical form of intervention is the reconfiguration of the screen to increase visual perception and reduce strain on neck muscles. This intervention has received some attention in the surgical ergonomics literature, even though most of the studies were concerned about productivity increase (Hanna, Shimi, & Cuschieri, 1998; Hernandez, Travascio, Onar-Thomas, & Asfour, 2014; Miura et al., 2019; Rogers, Heath, Uy, Suresh, & Kaber, 2012). Two studies investigated the impact of laparoscopic screen position on neck muscle fatigue and reported reduced muscle activity when the screen was placed directly in the frontal eye axis. For instance, Matern et al. (2005) compared EMG of the neck extensor muscle in three monitor configurations: (A) front at eye level, (B) front in the height of the operating field, and (C) 45° to the right side at eye level. Their results showed that Position A resulted in the least EMG of the neck muscles compared to the other two positions (1.5%MVC vs. 2%MVC vs. 3.5%MVC), even though Position B resulted in the shortest completion time. The second study by Rogers et al. (2012) tested productivity and workload based (NASA-TLX) on three different monitor configurations. The first two configurations were similar to Conditions A and B used in Matern et al. (2005); however, the third configuration was a vertical stack of the two screens. Similar to the previous study, this study showed that collocating the screens with the operating surface as in Condition B of Matern et al. (2005)'s study resulted in the least task completion times as proportions of baseline traditional minimally invasive surgical (MIS) set-up, compared to placing the screen at eye level (0.91 ± 0.14 vs. 0.96 ± 0.13). This advantage came at the expense of a non-ergonomic neck posture (increased EMG of extensor muscle: 3.5%MVC vs. 2%MVC vs. 1.5%MVC) as shown by (Matern et al., 2005). While neck muscle activity was not measured in this study, the monitor position did not impact physical workload. Thus, it seems that the optimal monitor configuration in laparoscopic would be a trade-off between better ergonomics and productivity, requiring further research.

Armrests have also been explored as potential interventions to reduce the fatigue and discomfort associated with quasi-static abducted shoulders during minimally invasive laparoscopic surgeries. Evaluations of these armrests have shown positive results such as reduced shoulder discomfort, error rates, and energy consumption in the form of oxygen uptake (Galleano et al., 2006; Jafri, Brown, et al., 2013; Steinhilber et al., 2015). For instance, using the Fundamentals of Laparoscopic Surgery (FLS) set-up, Galleano et al. (2006), showed that an armrest significantly reduced subjective discomfort in the right (1.6 ± 2.5 vs. 6.1 ± 3.5) and left deltoid muscles (1.7 ± 2.3 vs. 5.8 ± 5.5) over a 10cm visual analog scale (VAS). Furthermore, the study participants thought the task was simplified by the introduction of the armrest. However, these armrests are fixed during surgery, often leading kinematic mismatch between surgeon and support. Additionally, armrests can get in the way of the many tasks that must be completed while performing surgery and it is difficult to put and remove an armrest from the surgical field due to the many infection control limitations (Abdelrahman et al., 2018). Therefore, this intervention is viable only for minimally invasive procedures or very small surgical fields, creating the need to explore other interventions for open-invasive procedures.

1.4 Exoskeletons and Their Potential Benefit

Exoskeletons are wearable man-machine systems that were designed as external mechanical augmentation structures for humans. While they were initially developed for rehabilitation purposes for patients suffering loss of motor function due to spinal or brain injury (Ali, 2014), they have recently seen numerous applications in manual task performance industries; particularly in repetitive lifting and overhead work (Gillette & Stephenson, 2019; Rashedi, Kim, Nussbaum, & Agnew, 2014; Theurel, Desbrosses, Roux, & Savescu, 2018). Exoskeletons are either powered by an electromechanical power source (active exoskeletons) or energy stored in elastic elements (passive exoskeletons) due to the movement of the wearer.

Exoskeletons have shown positive results at being able to reduce the biomechanical demand associated with physical task performance. For example, Theurel et al. (2018) showed significantly reduced deltoid muscle activity in a simulated load-lifting (6%MVC vs. 13%MVC) and stacking task (3%MVC vs. 12%MVC). Furthermore, another study reported up to a 45% reduction in peak shoulder muscle loading in a repetitive overhead drilling task (Kim et al., 2018). The decrease in muscle activation translated to a 20% reduction in task completion time. Similar results have been reported in repetitive lifting tasks for the lumbar region (Bosch, van Eck, Knitel, & de Looze, 2016). This reduction in muscle activity may have the long-term benefit of reducing WRMSD by eliminating the fatigue and discomfort associated with awkward postures while performing surgical tasks.

While exoskeletons have shown promising results in other industries, the healthcare sector mainly, surgeon populations, are yet to see the potential benefits that exoskeletons might have in reducing WRMSDs. Considering that surgeons assume awkward postures while operating, exoskeletons could be a potential solution to lowering WRMSDs in the operating room.

1.5 Aim of Research

As shown in the prior paragraphs, there is the need to address work-related musculoskeletal disorders among healthcare workers, particularly, surgeons; and exoskeletons may be a viable intervention. However, exoskeletons have been relatively unexplored in healthcare delivery, and much research is needed to drive how exoskeleton interventions are deployed for maximized efficiency. Hence, the goal of this research study was to explore the utility of postural support exoskeletons as a means of reducing or eliminating the biomechanical demand on surgeons, which may translate into reduced musculoskeletal disorders. This dissertation explored the topic in three chapters (Chapters 3-5). Using vascular surgery as a model (Chapter 3), a physical and postural workload study was conducted to provide a proof of concept that exoskeleton interventions need to be targeted to specific surgeries and body segments for maximized benefit. (Chapter 4) developed a technique for using segmental kinematics data to identify surgeries that can benefit from exoskeleton interventions. The goal was to provide a standardized technique for the decision to deploy or not to deploy exoskeleton interventions in the operating room. Finally, a two-phase laboratory study was conducted to assess the benefit of exoskeleton on the demand on specific muscles of interest using electromyography and a subjective discomfort survey. This study is described in Chapter 5.

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CHAPTER 2. BACKGROUND

2.1 Functional Anatomy of the Neck, Low Back and Shoulders

2.1.1 Neck

The neck consists of the seven bones (C1-C7) in the cervical region of the spine and the muscles responsible for the movement of the neck in all its planes of motion. The first two cervical vertebrae (C1-C2), in combination with the occipital bone of the skull, form the craniocervical joint (CCJ) while the remaining five vertebrae (C3-C5) for the sub-axial vertebral spine (Kaiser, Reddy, & Lugo-Pico, 2020).

2.1.1.2 Cervical vertebrae

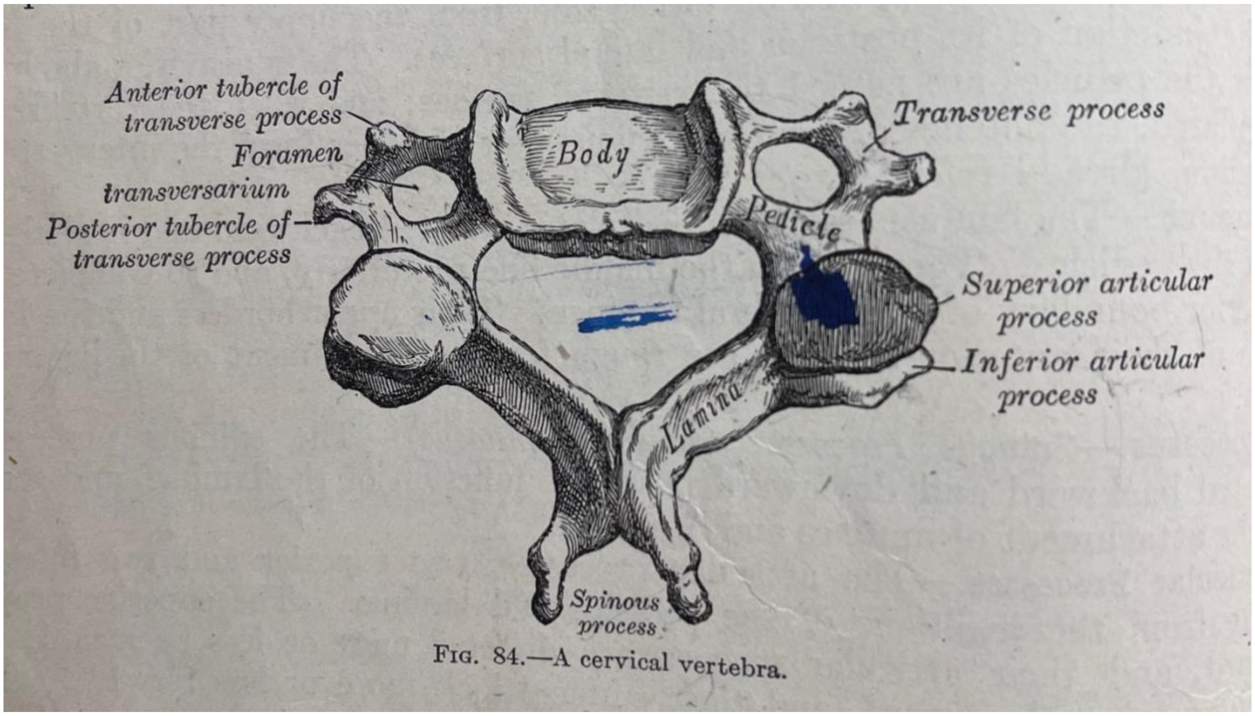

Figure 2.1: Superior View of a Cervical Vertebra. (Gray & Lewis, 1918, image 84)

As seen in Figure 2.1, a typical cervical vertebra has a significantly smaller body than the lumbar vertebrae. Cervical vertebrae can clearly be distinguished from the thoracic and lumbar vertebrae by the foramen in each of their transverse processes and relatively short spinous processes. Furthermore, the superior and inferior articular surfaces and facets are designed to articulate at a posterolateral angle of approximately 45°. This feature, combined with the relatively short spinous process of the cervical vertebrae, enables the significantly more extensive range of motion in the neck than other sections of the spinal column (Watkins & Mathieson, 2009).

2.1.1.3 Muscles of the head-neck

There is an intricate network of muscles that control the motion of the head-neck through its numerous planes of motion. The predominant single plane motion of the head-neck are flexion-extension and rotation. For the proposed study, the discussion of neck muscles was grouped into the type of motion that the muscles initiate.

2.1.1.3.1 Neck extensors

The neck extensor muscles consist of approximately four layers of muscles. Namely: Layer 1: levator scapulae and upper trapezius, Layer 2: Splenius capitis and cervicis, Layer 3: Semispinalis capitis, and Layer 4: Semispinalis and cervicis and multifidus. However, we would classify Layer 1 as superficial and all the other layers as deep muscles.

Layer 1: The trapezius (see Figure 2.2) is a large, diamond-shaped, and superficial muscle that originates from the occiput and extends down to the 12^{th} vertebrae while inserting into multiple points such as distally on the spine of the acromion, scapula, and distal third clavicle (Bakkum & Cramer, 2014). The trapezius muscle is primarily considered as a scapula stabilizer; however, because of its origin from the occiput and attachment to some of the lower cervical bones, it acts as a synergist to the sternocleidomastoid (M. Gatterman, 2012). The levator scapulae (see Figure 2.2) originates from the transverse process' posterior tubercle of the first four cervical vertebrae and inserted into the medial border of the scapula. While this muscle is considered a shoulder elevator, its contraction leads to cervical spine extension when the shoulders are in a fixed position.

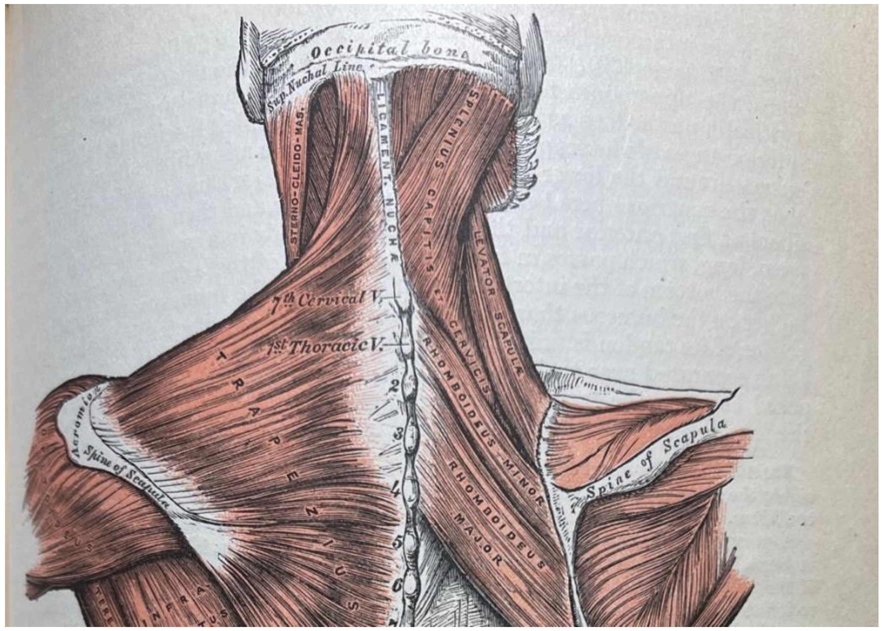

Figure 2.2: The Trapezius and Levator Scapula. (Gray & Lewis, 1918, image 409)

Deep Muscles: The deep muscles of the neck are arranged in layers beneath the superficial muscles. These muscles either act bilaterally to initiate neck extension or unilaterally to initiate lateral flexion.

2.1.1.3.2 Neck flexors

Sternocleidomastoids: These are two pairs of muscles that originate from the upper edge of the sternum and upper face of the clavicle, merge into a single pair of muscles that moves superolaterally on both sides of the neck to insert into the mastoid process of the temporal bone (Figure 2.3). A unilateral contraction of one of the muscles results in ipsilateral flexion and contralateral rotation, while a bilateral contraction results in neck flexion (Bordoni & Varacello, 2018b).

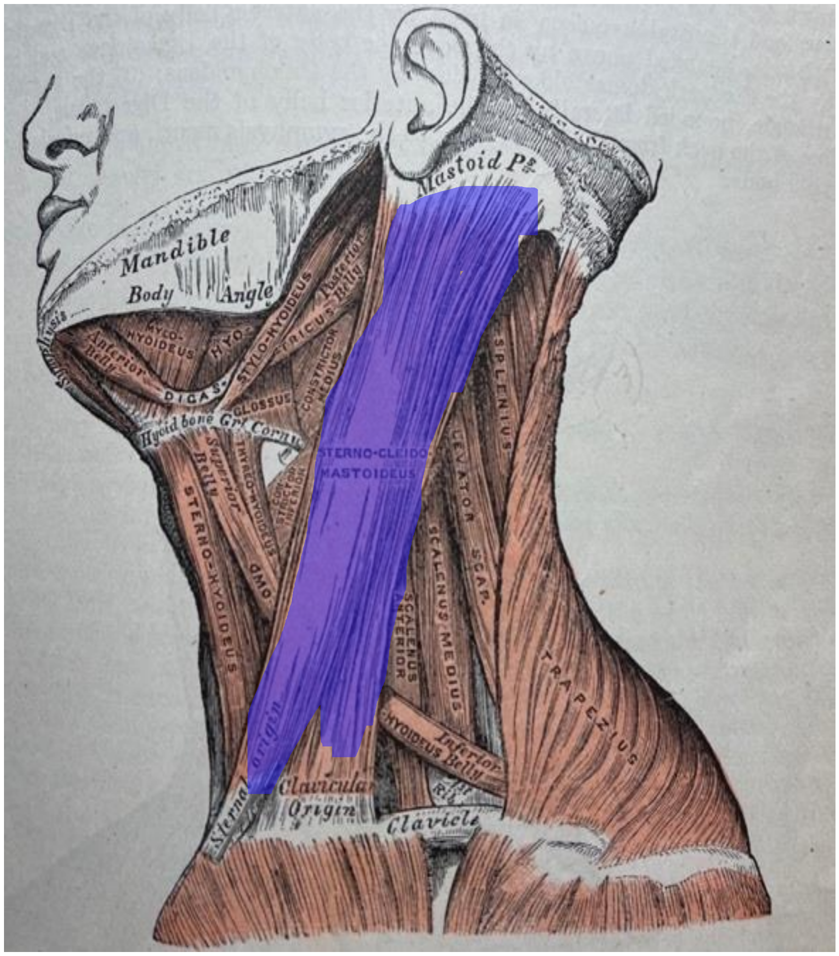

Figure 2.3: The Sternocleidomastoid. (Gray & Lewis, 1918, image 385)

Anterior scalene: The anterior scalene muscles shown in Figure 2.4 are a pair of muscles located deep relative to the sternocleidomastoid. They originate from the transverse processes of the third to the sixth cervical vertebrae on both sides and insert into the upper face of the first rib. When bilaterally contracted, a neck flexion motion occurs (Bordoni & Varacello, 2018a).

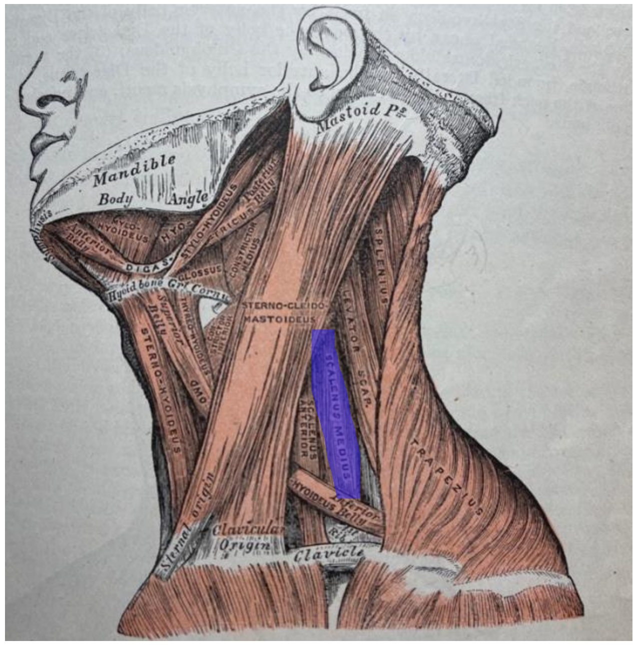

Figure 2.4: The Anterior Scalene Muscle. (Gray & Lewis, 1918, image 385)

2.1.2 Low Back

The low back consists of the five lumbar vertebrae (L1 to L5) and their associated ligaments, musculature, and intervertebral disc. Strain on any of these components can lead to low back pain.

The only bony structures of the low back are the five lumbar vertebrae. Functionally, each vertebra can be divided into three elements, namely: vertebral body, posterior elements, and pedicles connecting the vertebral body to posterior elements (Galbusera & Wilke, 2018). The vertebral body is the main load-bearing (M. A. Adams & Hutton, 1980). It is cylindrical with a kidney-shaped cross-sectional shape and slightly concave cranial and caudal surface for attachment of the intervertebral disc. For resisting compressive physiological loads, the internal architecture of the vertebral body consists of compact bone shells called Corticalis reinforced by vertical and horizontal struts called trabeculae (Ritzel, Amling, Pösl, Hahn, & Delling, 1997). The posterior elements consist of the spinous process, transverse process, laminae, and articular facets, as shown in Figure 2.5. Collectively, the posterior elements guide the lumbar spine's movement and protect the spinal cord in the vertebral foramen. The pedicles are two short bones that connect the vertebral body to the posterior elements.

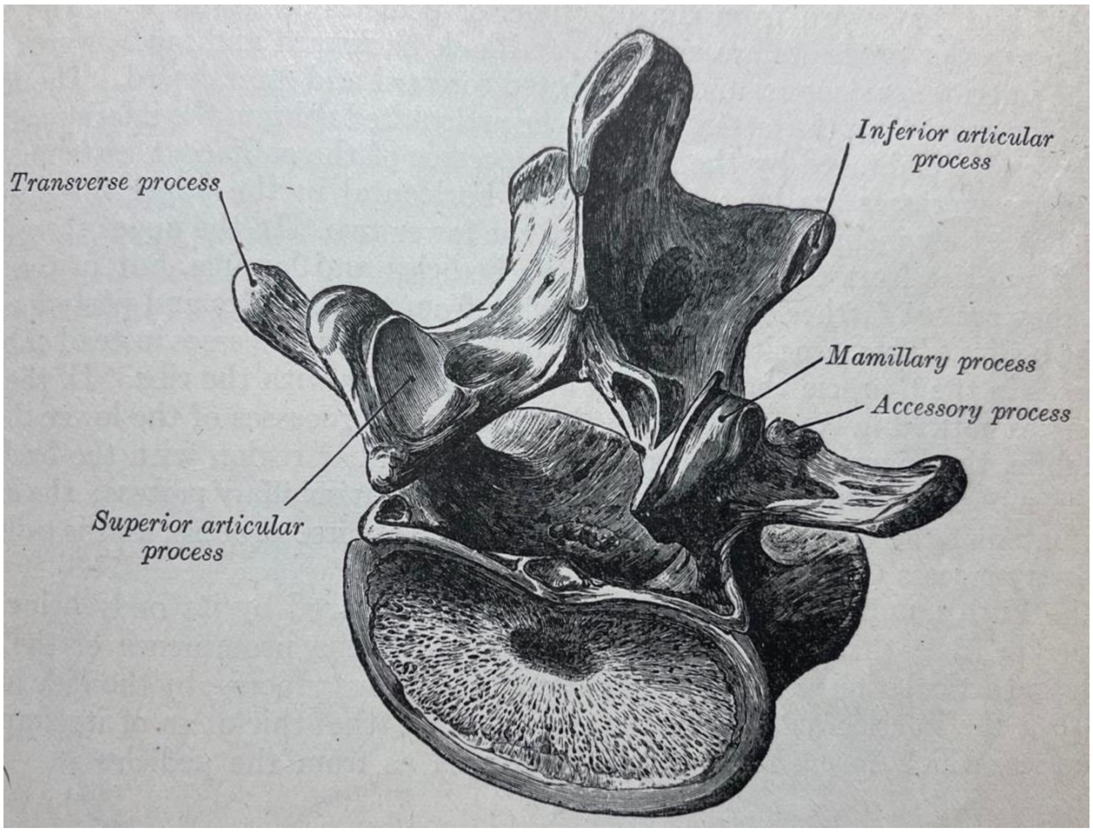

Figure 2.5: Three-Dimensional View of the Lumbar Vertebra.(Gray & Lewis, 1918, image 93)

Adjacent vertebral bodies are connected by a flexible fibrocartilaginous joint called intervertebral discs. Each disc comprises a central nucleus pulposus surrounded by outer annulus fibrosis. The nuclear pulposus is composed mainly of negatively charged proteoglycans, which have the affinity to imbibe water thus, accounting for 80% of the weight of the nuclear pulposus in young individuals (Galbusera & Wilke, 2018; Urban & McMullin, 1988). The annulus fibrosis consists of up 20 collagen Type-1 sheets arranged in layers called lamellae. The annulus fibrosis has fewer proteoglycans than the nuclear pulposus and thus holds approximately 10% less water, and it is less permeable. This difference in water retaining capability between the nuclear pulposus and annulus fibrosis induces an intrinsic pressure of about 0.5 to 1.5MPa in the nuclear pulposus in neutral spine posture. This pressure is vital to the spine's ability to resistive high compressive loads. Furthermore, the unique arrangement of the collagen fibers in the annulus fibrosis provides a tensioning resistance, similar to fiber reinforcement of a vehicle tire during spine flexion, extension, and lateral bending.

A total of seven primary ligaments exist to stabilize the spine in specific motion directions (Galbusera & Wilke, 2018). As shown in Figure 2.6, the anterior longitudinal ligament covers the ventral side of the spine, running over the vertebral bodies and intervertebral discs. Its anterior position primarily resists motion in trunk extension with a marginal effect on other motion directions (Heuer, Schmidt, Klezl, Claes, & Wilke, 2007). The anterior spinal column is covered by the relatively narrow and thin posterior longitudinal ligament. Positioned behind the vertebral body, it limits spinal flexion motion. The dorsal surface of the spinal column is covered by the Ligamentum Flava, which links the laminae of successive vertebrae. It is relatively thicker than all spinal ligaments, and it is composed of mainly elastic fibers that give it mechanically pre-stressed (Galbusera & Wilke, 2018). Its primary function is to resist trunk flexion. Two intertransverse ligaments connect the transverse processes of adjacent vertebrae. These two oppose lateral bending motions of the spine. The interspinous and supraspinous ligaments blend to connect adjacent spinous processes. In the lumbar region, the supraspinous ligament is wider and thicker than the interspinous ligament. Due to their posterior/ dorsal positions, they mainly resist trunk flexion (Heuer et al., 2007).

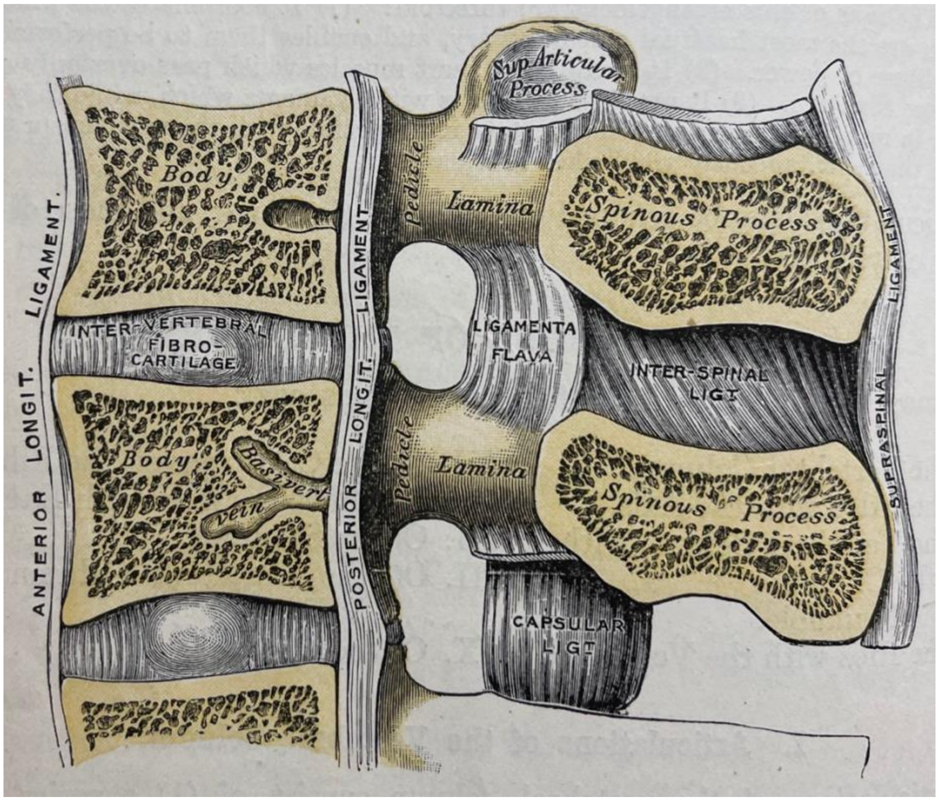

Figure 2.6: Lateral view of the lumbar spine showing the ligaments. (Gray & Lewis, 1918, Image 301)

Muscles of the lumbar region are mainly responsible for trunk extension or holding quasi-static trunk postures. The primary back extensor muscle is the erector spinae group of muscles and will be the only muscle discussed for this research. The erector spinae originates from the iliac crest and lumbar aponeurosis and inserts superiorly into several ribs, as shown in Figure 2.7 (Mayer, Mooney, & Dagenais, 2012). In the lumbar region, the erector spinae originate as one muscle but splits into its constituent three (iliocostalis, longissimus, and spinalis), which insert into various points on the ribs. Considering the small moment arm about the L5-S1 joint in the sagittal plane, a greater force is required to be generated by these muscles to counteract the overall mass of load and torso during trunk extension.

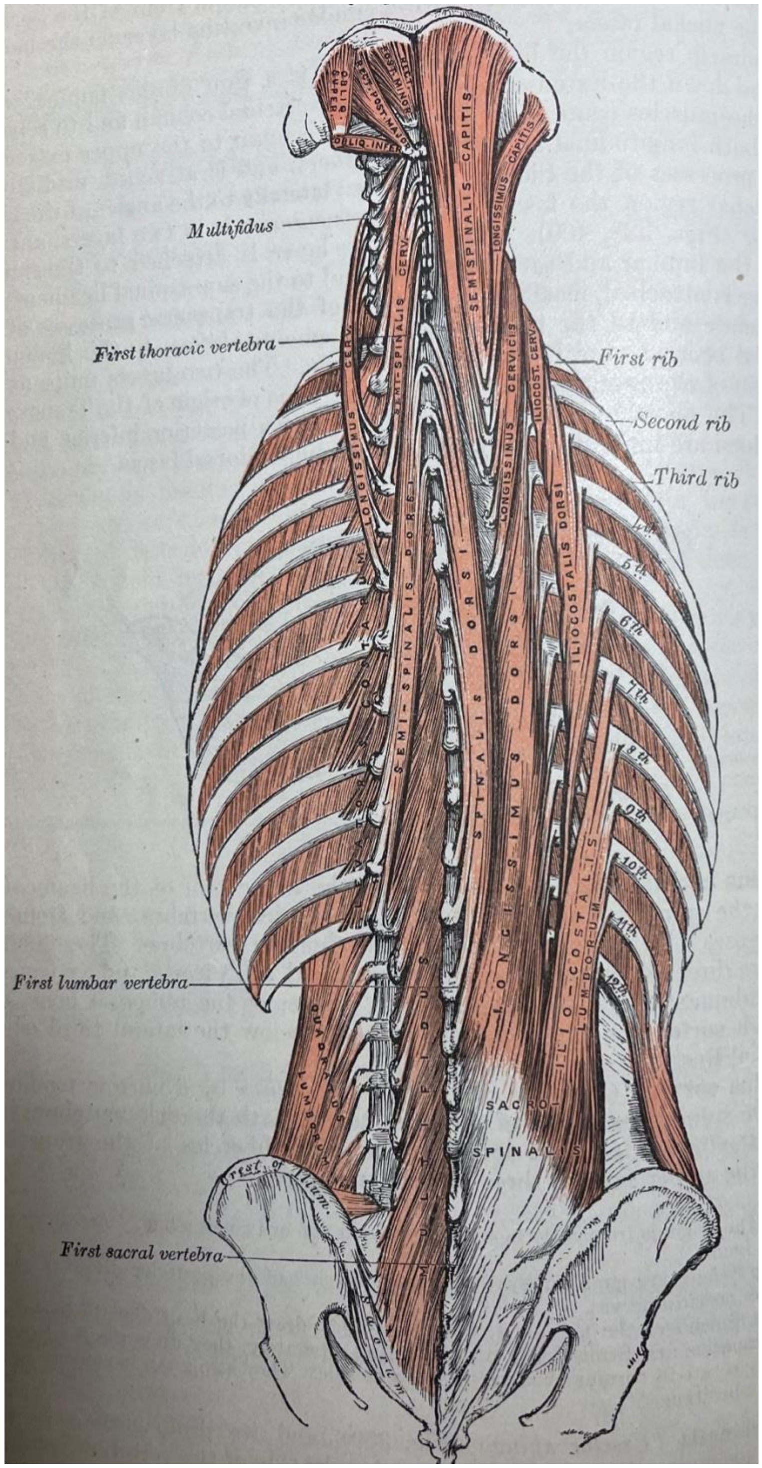

Figure 2.7: Dorsal View of the Trunk Showing the Erector Spinae Muscles. (Gray & Lewis, 1918, Image 389)

2.1.3 Shoulder

The shoulder is one of the most complex joints of the human body due to the interplay of numerous muscles and bony articulations. This joint consists of the clavicle, scapula, humerus, and articulations at the glenohumeral, acromioclavicular, and sternoclavicular joint. For description, the shoulder joint will be discussed from the perspective of its structural components and articulations. i.e., Bony Anatomy, joint articulations, static stabilizers, and muscles (dynamic stabilizers).

2.1.3.1 Bony Anatomy

Three bones, namely: humerus, scapula, and clavicle, make up the shoulder joint.

2.1.3.1.1 Humerus

It is the longest and largest bone of the upper extremity, consisting of a long shaft and two heads on the proximal and distal ends. While the proximal end articulates with the glenoid fossa to form the glenohumeral joint, the distal end articulates with the radius and ulnar to form the elbow joint, as shown in Figure 2.8. The humeral head is inclined to the shaft at an approximate angle of 130° to 150° and just inferior to it is an anatomical neck which divides the two bony prominences called the greater and lesser tubercles as shown in Figure 2.9 (Mostafa & Varacello, 2018; Terry & Chopp, 2000). The greater tubercles consist of three facets, which serve as points of attachment for the supraspinatus, infraspinatus, and teres minor muscles. The lesser tuberosity serves as a point of insertion for the subscapularis muscle. Collectively, these four muscles are called the rotator cuff, and it is responsible for providing dynamic stabilization at the shoulder joint (Maruvada & Varacello, 2018).

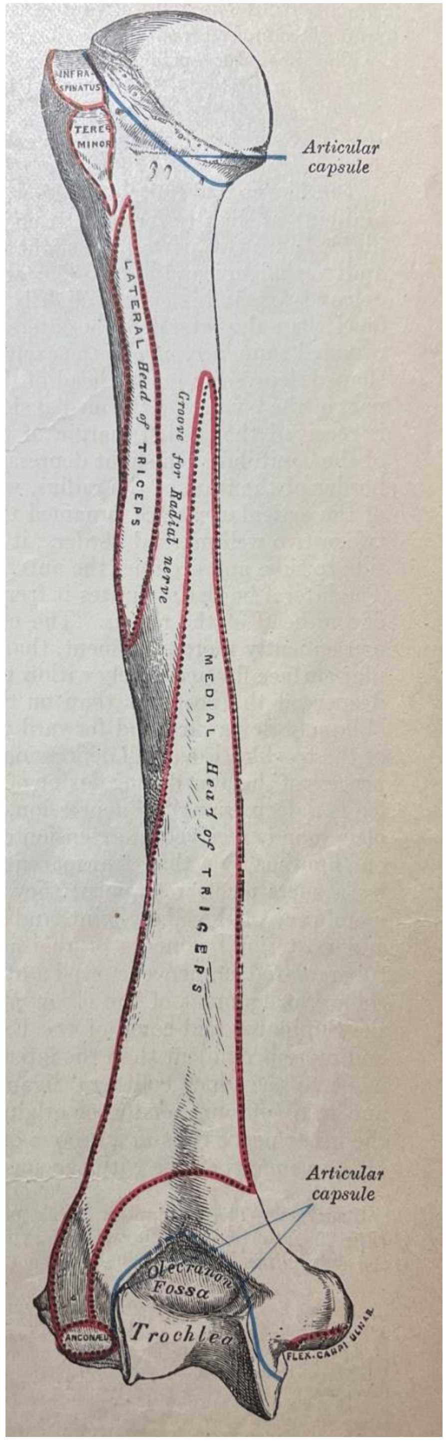

Figure 2.8: Anterior View of the humerus.(Gray & Lewis, 1918, Image 208)

2.1.3.1.2 Scapula

The scapula is a thin, large and triangular-shaped bone that lays over the posterolateral sections of the thoracic cavity from the second to the seventh rib (Terry & Chopp, 2000), as shown in Figure 2.9. Instead of bony or ligamentous attachments to the axial skeleton, the scapula is held in place via pressure from overlaying muscles such as the trapezius, serratus anterior, rhomboid major, and minor muscles (Culham & Peat, 1993).

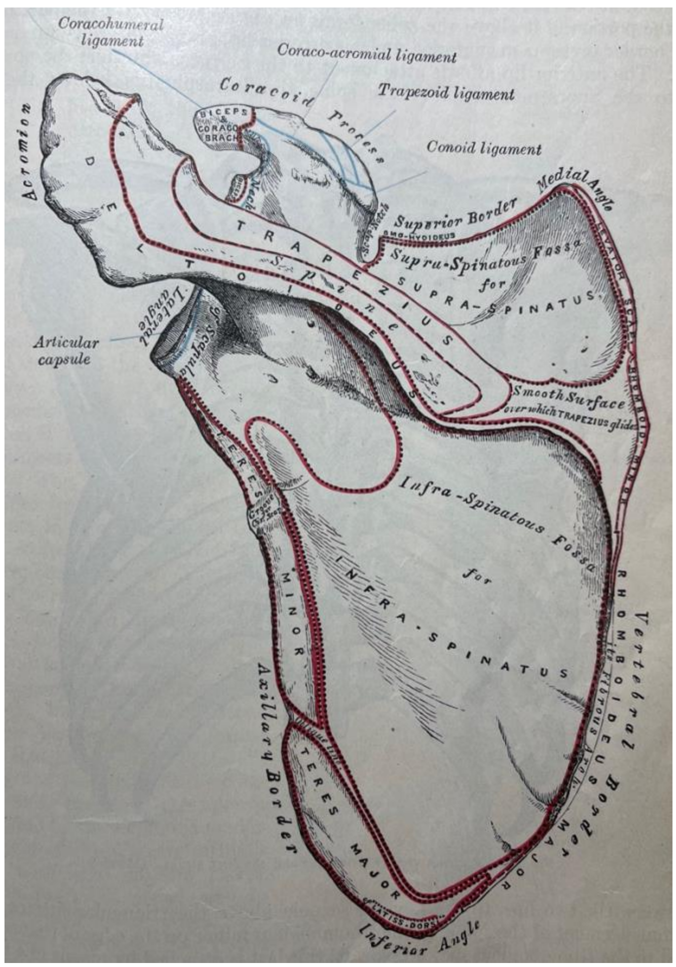

Figure 2.9: Dorsal View of the Left Scapula.(Gray & Lewis, 1918, Image 203)

There is a superior bony prominence called the spine of the scapula, which runs diagonally across the posterior body of the scapula and extends laterally into a bony projection called the acromion. The spine separates the supraspinatus and infraspinatus muscles and also serves as the insertion point for the trapezius muscles and the origin of the posterior deltoid muscles. The outwardly projected acromion serves as a lever arm to maximize the action of the deltoid during shoulder abduction. Furthermore, the acromion articulates with the distal end of the clavicle (acromioclavicular joint).

Just beneath the acromion is a shallow depression called the glenoid fossa, which serves as a point of articulation between the humerus and the scapula. Also, on the lateral edge of the anterosuperior portion of the scapula is a hook-like bony projection called the coracoid process, which serves as a point of attachment for the coracohumeral ligament. This ligament prevents inferior translation of the humeral head during shoulder adduction.

2.1.3.1.3 Clavicle

The clavicle in Figure 2.10, also called the collarbone, acts as a struct to connect the shoulder girdle to the trunk medially via the sternoclavicular joint and laterally via the acromioclavicular joint. Its flat lateral third surface serves as attachment sites for the trapezius and deltoid muscles. The position of the clavicle ensures that the upper extremities are far away, and the shoulder range of motion is unimpeded (Hyland & Varacello, 2019).

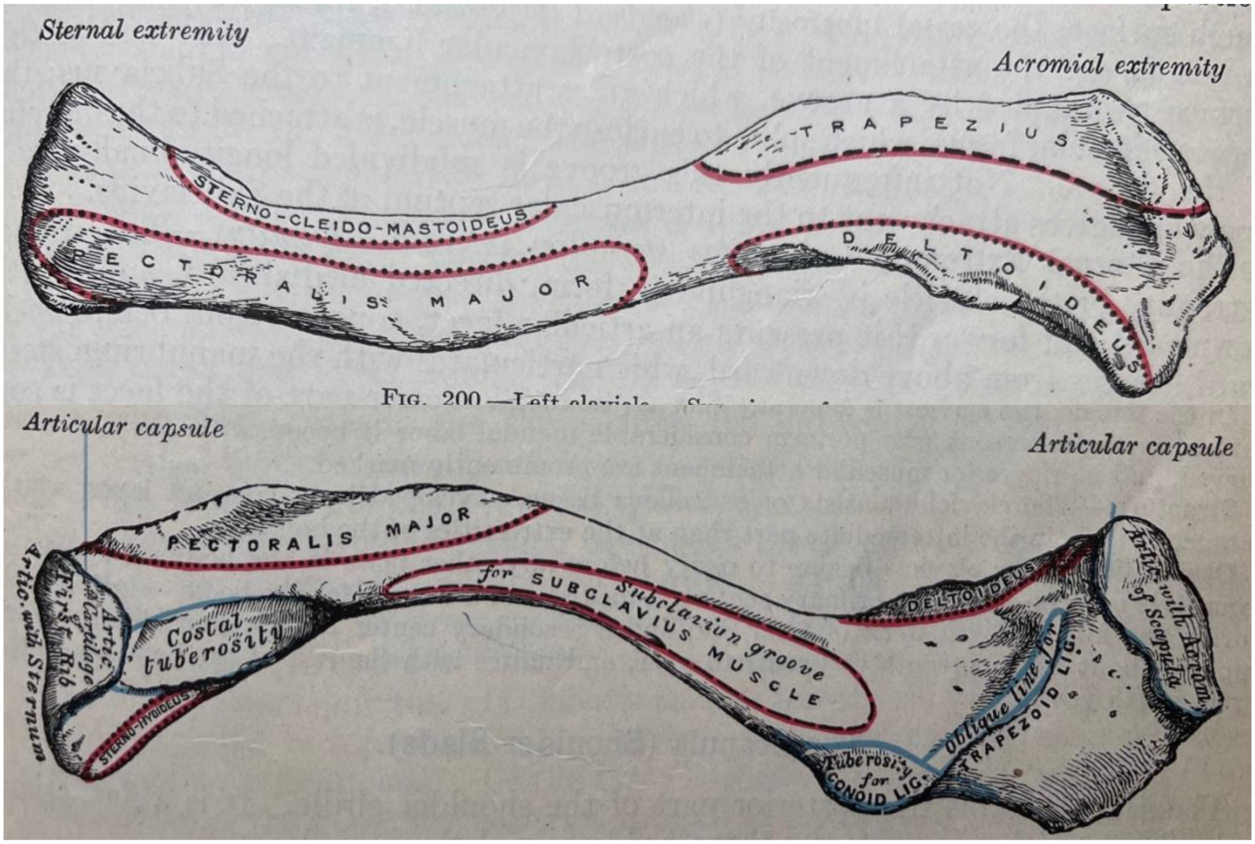

Figure 2.10: Superior and Inferior Surface of the Clavicle. (Gray & Lewis, 1918, image 200 & 201)

2.1.3.2 Joint articulations

2.1.3.2.1 Glenohumeral joint

The glenohumeral joint shown in Figure 2.11 is ball and socket joint which primarily connects the upper extremity to the trunk. The extreme mobility of this joint can be attributed to the mismatched difference between the large humeral head and the small articular surface of the glenoid cavity (Terry & Chopp, 2000). While at any given time, only 25% to 30% of the humeral head makes contact with the glenoid fossa, the glenohumeral joint is highly stable, thanks to its inherent design and complex interplay of passive and dynamic Mechanism.

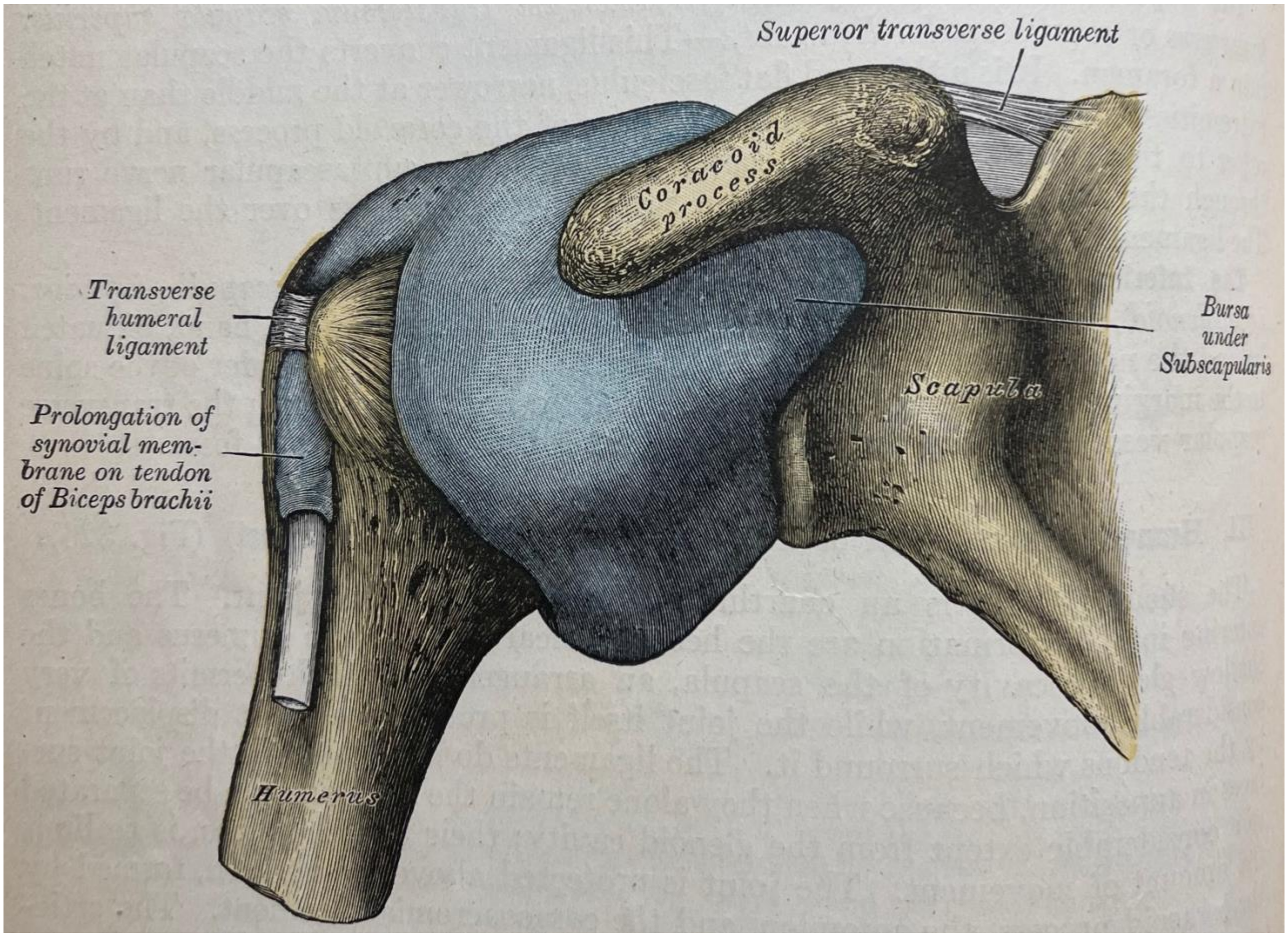

Figure 2.11: Anterior View of the Glenohumeral Joint.(Gray & Lewis, 1918, Image 327)

2.1.3.2.2 Passive mechanism

Articular surface: While the glenoid cavity's curvature (see Figure 2.12) is slightly flattened relative to the humeral head, its peripheral articular cartilage is relatively thicker, creating significant conformity around the humeral head (Soslowsky, Flatow, Bigliani, & Mow, 1992). This conformity, in addition to compression forces from the rotator cuff and surrounding muscles, provides the necessary stability for shoulder function (Lippitt et al., 1993). Additionally, the glenohumeral joint contains approximately 1mL of fluid tightly sealed by the joint capsule. This fluid is under slightly negative pressure, thus providing a suction effect to resist humeral translation.

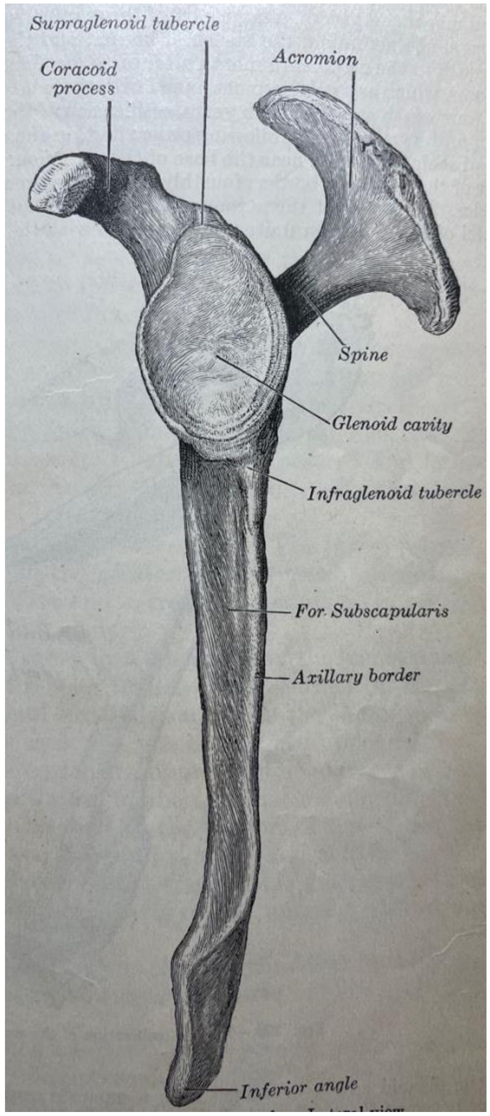

Figure 2.12: Lateral view of the scapula showing the glenoid fossa. (Gray & Lewis, 1918, Image 205)

Glenoid labrum: This is a dense fibrocartilaginous tissue located at the margins of the glenoid fossa (Culham & Peat, 1993), as shown in Figure 2.13. Due to its location, it serves to increase the contact area between the glenoid fossa and the humeral head by deepening the concavity of the glenoid socket by 5mm and 9mm in the anteroposterior and superolateral planes respectively to contribute approximately 50% of the socket depth (Howell & Galinat, 1989). Thus, degeneration in the structure of the glenoid labrum could impact joint stability. In addition to providing stability, the labrum anchors the shoulder ligamentous structures.

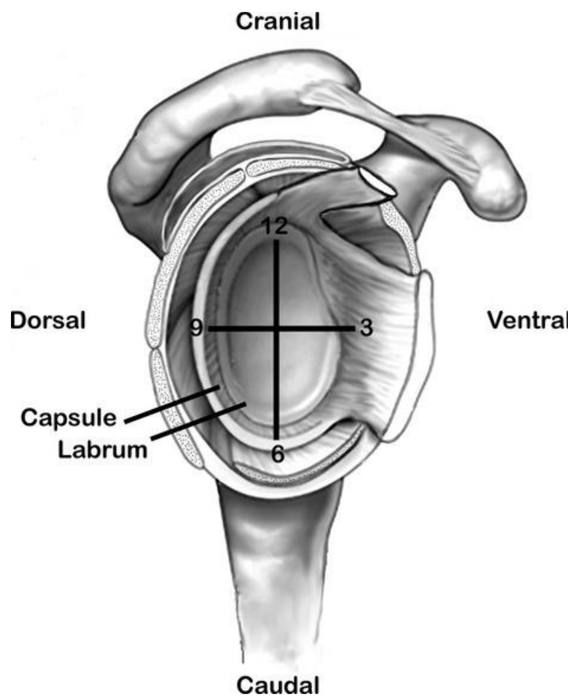

Figure 2.13: Lateral View of the Glenoid Cavity. (Witherspoon, Smirnova, & Mciff, 2014) Figure used with permission from Wiley and Sons.

Joint Capsule: The shoulder capsule medially attaches to the periphery of the glenoid fossa and laterally around the anatomical neck of the humerus. With a surface area approximately twice that of the humerus, it can extend to allow the shoulder its extensive range of motion. The action of the capsule is significantly aided by shoulder ligaments, and they are relatively inactive or lax in midrange motion (Lippitt et al., 1993). However, during extreme shoulder motion such as abduction or rotation, the capsule and ligaments tense up to provide a stabilization effect.

Ligaments: Four main ligaments stabilize the glenohumeral joint. These are the coracohumeral ligament, superior glenoid ligament, middle glenohumeral ligament, and inferior glenohumeral ligament, as shown in Figure 2.14.

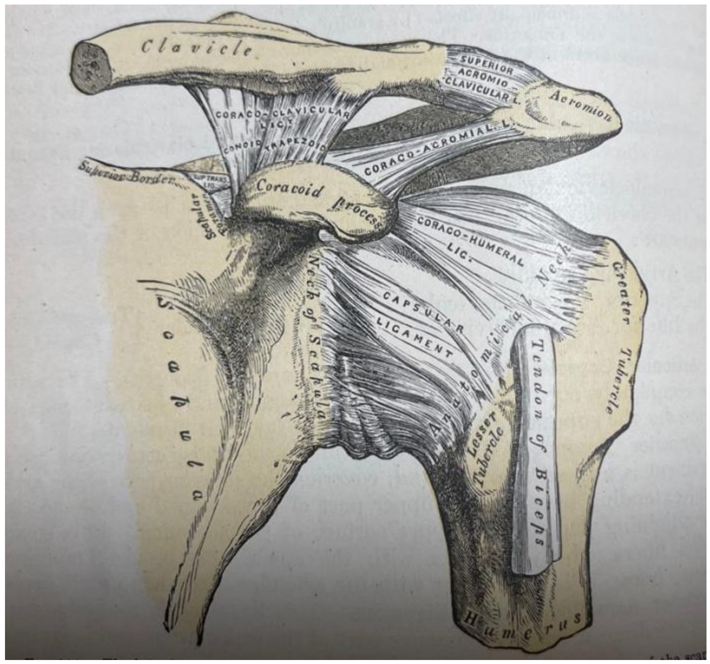

Figure 2.14: Ligaments of the Glenohumeral Joint. (Gray & Lewis, 1918, Image 326)

The coracohumeral ligament originates as a band of a thick capsule from the lateral and base of the scapula coracoid process and inserts into the greater and lesser tubercles of the humerus (Terry & Chopp, 2000). Due to its superior position relative to the other glenohumeral ligaments and its lateral passage across the joint, it tenses up in arm adduction and constrains inferior translation of the humeral head (Warner, Deng, Warren, & Torzilli, 1992).

The superior glenohumeral ligament extends from the anterosuperior periphery of the glenoid and inserts into the top of the lesser tuberosity. The function of this ligament is similar to that of the coracohumeral ligament.

The middle glenohumeral ligament has its origin from the margin of the glenoid cavity and inserts into the anatomical neck and lesser tuberosity of the humerus (Turkel, Panio, Marshall, & Girgis, 1981). Its main function is to prevent anterior humeral head translation during lower range shoulder abduction (Terry & Chopp, 2000).

The inferior glenohumeral ligament originates from the anteroinferior portions of the labrum and glenoid periphery and inserts into the humeral lesser tuberosity. It is the thickest of all glenohumeral joints and functions mainly to resist anterior translation of the humeral head while throwing with shoulder abducted and externally rotated (Terry & Chopp, 2000).

2.1.3.3 Dynamic stabilizing mechanism

2.1.3.3.1 Deltoid muscle

The deltoid (shown in Figure 2.15) is a triangular muscle that originates from the lateral third of the clavicle and scapula’s spine and inserts into the deltoid tuberosity of the lateral side of the humeral body. This muscle divides into three sections (anterior, lateral, and posterior deltoid). The primary role of the deltoid is shoulder (lateral) abduction with the anterior and posterior parts providing important stabilizing function. Furthermore, the anterior deltoid works in tandem with the pectoralis major muscle to flex the shoulder during ambulation (Elzanie & Varacello, 2018).

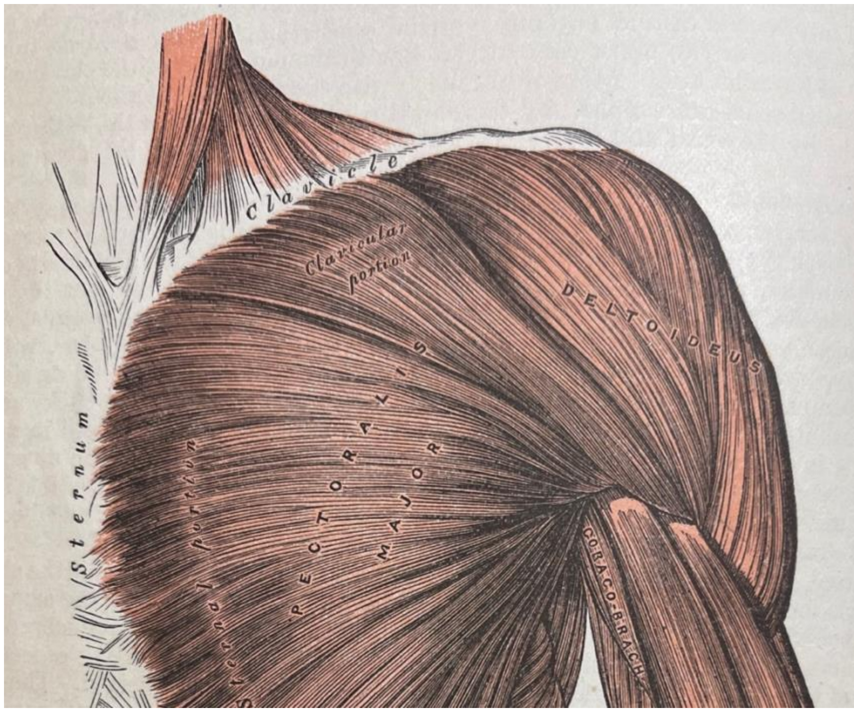

Figure 2.15: The Deltoid Muscle of the Shoulder.(Gray & Lewis, 1918, Image 410)

2.1.3.3.2 Rotator cuff

The Rotator cuff is a group of muscles consisting of the subscapularis, infraspinatus, supraspinatus, and teres minor, as shown in Figure 2.16. These muscles act dynamically to steer the humeral head within the glenoid cavity. The interplay of these muscles and the passive structures discussed in the previous section ensures joint stability in the numerous planes of motion. Compared to the more superficial muscles such as the deltoid, latissimus dorsi, trapezius, and pectoralis, the rotator cuff muscles have a smaller cross-sectional area and moment arm around the humeral center of rotation. Thus, they generate relatively minor forces during glenohumeral movement and much situated for stabilizing the joint (Terry & Chopp, 2000).

The supraspinatus muscle originates from the supraspinal fossa and traverses underneath the coracoacromial joint to insert into the humerus's greater tubercle (Ombregt, 2013). While stabilizing the joint, it also acts in tandem with the deltoid muscle to elevate the shoulder.

The infraspinatus is a thick triangular muscle that originates from the infraspinous fossa of the scapula (origin) and converges into a tendon that inserts into the middle facets of the humeral greater tuberosity (Williams & Obremskey, 2019). It clearly distinguishes itself from the teres minor muscle, as shown in Figure 2.17, and the combined action of the two muscles stabilizes the glenohumeral joint against partial posterior dislocation. Furthermore, the infraspinatus and teres minor are the primary shoulder external rotators (Terry & Chopp, 2000).

The teres minor muscle lies just inferior to the infraspinatus muscle. It arises from the dorsal surface of the scapula and inserts into the most inferior greater tubercle facet (Gatterman & McDowell, 2012; Terry & Chopp, 2000). Its functions are similar to that of the infraspinatus muscle.

The subscapularis is a triangular-shaped muscle that originates from the subscapular fossa of the anterior surface of the scapula and inserts and narrows down into a single tendon to insert into the lesser tuberosity of the humerus. Due to its anterior position relative to the other rotator cuff muscles, its action rotates that shoulder internally (Terry & Chopp, 2000).

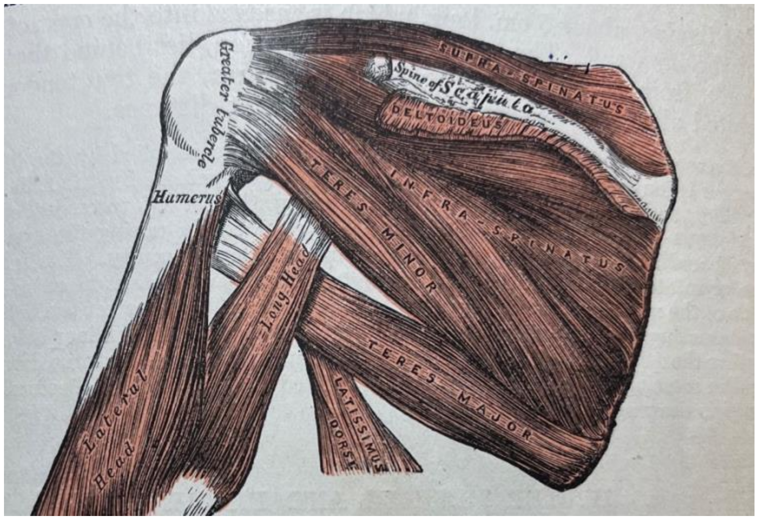

Figure 2.16: Posterior View of the Shoulder Joint to show the Rotator Cuff Muscles.(Gray & Lewis, 1918, Image 412)

Acromioclavicular (AC) joint: The lateral edge of the clavicle and medial edge of the acromion forms a diarthrodial joint called the AC joint, as shown in Figure 2.15 (Terry & Chopp, 2000). This joint is stabilized by a combination of a capsule, intra-articular discs, and ligaments (Beim, 2000). An anterior and superiorly thick capsule surrounds the AC joint, and this is reinforced by AC ligaments with four components (superior, inferior, anterior, and posterior). The superior AC ligament has the strongest fibers, and these fibers blend in with those of the deltoid and trapezius muscles (Terry & Chopp, 2000).

Additional joint stability of the AC joint is provided by the two coracoclavicular ligaments (Lucas, 1973), which connects the superior surface of the coracoid process to the trapezoid and conoid tuberosities of the clavicle, respectively. These two ligaments provide the support to suspend the shoulder girdle from the clavicle; thus, restraining vertical translation (Terry & Chopp, 2000).

2.2. General Physical Ergonomic Issues in the Surgical Environment

The surgeon population is known to suffer significantly from work-related musculoskeletal disorders. For instance, a survey report of 135 general surgeons showed that 82.9%, 68.1%, and 57.8% suffered from neck, low back, and shoulder-related problems, respectively (Szeto et al., 2009). In agreement with this study, Adams, Hacker, McKinney, Elkadry, & Rosenblatt (2013) conducted a cross-sectional survey of 495 gynecological surgeons over twelve months. Their results showed that similar prevalence rates for the back (75.6%), neck (72.9%), and shoulder (66.6%) problems. Furthermore, their report showed that many of the surgeons believed that performing surgery caused or even worsened the symptoms of their musculoskeletal problems. According to a systematic review by Epstein and colleagues (Epstein et al., 2018) on surgeons and interventionists, the estimated career prevalence of degenerative lumbar spine among was 19%, and this appeared to increase over time in interventional cardiologists. It is no wonder that vascular surgeons have been reported to experience significantly higher musculoskeletal pain (4.4 ±2.3 out of 10 on the Borg CR10 scale) after performing procedures (Wohlauer et al., 2019). Moreover, significant pain and high workload amongst this population have been associated with surgeon burnout and inefficiency, and some surgeons even claimed that their pain could lead to their early retirement. Clearly, ergonomic interventions need to be developed to curtail some of these problems.

The high prevalence of WRMSDs amongst surgeons can be attributed to several workplace factors that predispose them to awkward postures, repetitive exertions, and long hours of task performance (exposure time) that causes strain on the surgeon's bodies (Seagull, 2012). For example, Yurteri-Kaplan et al. (2018) quantified the duration and frequency of awkward postures between lead and assistant vaginal surgeon while performing procedures. Their results showed a high duration and frequency of awkward body postures such as lateral trunk bending, neck, and shoulder deviation. Besides, while both lead and assistant surgeon assumed significant awkward postures, the assistant surgeons spent a significantly higher amount of time in such postures.

Also, newer minimally invasive surgical techniques with significantly short patient recovery times are constantly preferred to traditionally invasive surgical techniques. While these procedures significantly benefit patients, they have adverse effects on surgeons. A typical example is laparoscopic surgery, which has been shown to induce significant back and neck static postures (Berguer, Rab, Alarcon, & Chung, 1997; Nguyen et al., 2002). Such static postures reduce blood flow and accelerate fatigue development on those body parts. Furthermore, laparoscopic surgery induces significant shoulder abduction in a static posture. Also, laparoscopic tools tend to induce awkward upper extremity postures, leading to significant fatigue and pain development (Nguyen et al., 2002; Szeto et al., 2012). It is no wonder that a recent survey of laparoscopic minimally invasive surgeons entitled: "Patients Benefit While Surgeons Suffer: An impending Epidemic" reported that of the 272 laparoscopic surgeons, 86.9% of them experienced physical pain or discomfort (Park, Lee, Seagull, Meenaghan, & Dexter, 2010) and complaints of such pains are predominant in the neck, low back and shoulder (Janki, Mulder, IJzermans, & Tran, 2017). Interestingly, these symptoms were weakly correlated with the surgeon's age and length of practice, making laparoscopic surgery a rife field for ergonomic interventions.

There have been attempts to improve operative room ergonomics to reduce the effects of work-related musculoskeletal disorders. These interventions aim to reduce or eliminate cumulative fatigue build-up that leads to strain/pain and discomfort during surgical task performance. One such intervention is interoperative breaks, sometimes referred to as work-rest cycles. The underlying idea of intraoperative breaks is to reduce the exposure time of risk factors of musculoskeletal disorders by allowing breaks during the surgical operation. This strategy has shown improved musculoskeletal discomfort surveys (Engelmann et al., 2011; Hallbeck et al., 2017; Vijendren et al., 2018), improved task precision in laparoscopic simulation tasks (Dorion & Darveau, 2013) without a significant increase in operating time. While interoperative microbreaks have shown positive results in fatigue and pain reduction, it comes at the cost of surgical workflow interruption, which increases the likelihood of surgeon errors (Blocker et al., 2013). Furthermore, this intervention is not applicable for emergency life-threatening surgeries for which a break may prove fatal to the patient. Hence further research is needed to design targeted work-rest routines optimally.

Reconfiguration of the layout of specific operating room equipment is another technique that has been explored. This type of intervention is predominant in laparoscopic minimally invasive procedures in which the surgeons' movement is restricted while looking at the monitor screen, leading to musculoskeletal fatigue development from the static neck, back, and shoulder postures (Berguer, Forkey, & Smith, 1999). One such reconfiguration is the repositioning of monitor screens to allow for easy visualization at ergonomic neck postures. While some studies have been conducted on this intervention technique, the focus has generally been on productivity instead of improving the ergonomics of the surgical tasks (Hanna, Shimi, & Cuschieri, 1998; Hernandez, Travascio, Onar-Thomas, & Asfour, 2014; Miura et al., 2019; Rogers, Heath, Uy, Suresh, & Kaber, 2012), with just a few focusing improved ergonomics (Matern et al., 2005; Rogers et al., 2012). As shown in Chapter 1, the optimal monitor position will likely be determined by good ergonomics and productivity; hence further research is required on this topic.

Another intervention that has been explored in laparoscopic minimally invasive surgery is the use of armrests to relieve shoulder discomfort. While armrests have shown positive results such as reducing shoulder discomfort, error rates, and energy consumption in the form of oxygen uptake (Galleano, Carter, Brown, Frank, & Cuschieri, 2006; Jafri, Brown, Arnold, Abboud, & Wang, 2013; Steinhilber et al., 2015), their benefit is limited to minimally invasive procedures due to their static nature. Thus, surgical procedures that require frequent upper arm out-of-plane movements may result in kinematic mismatches as these arm supports are stationary and do not move with the surgeon's upper extremities. For this reason, other interventions need to be explored for open-invasive procedures.

2.3 Exoskeletons

Exoskeletons (Figures 17-21) are external wearable mechanical devices used to augment the wearer's capability (de Looze et al., 2016). They were mainly designed for rehabilitation purposes (Viteckova, Kutilek, & Jirina, 2013) but have received substantial attention in the industrial setting, particularly in manual tasks for which automation is not a feasible option. Exoskeletons can generally be classified as active or passive, depending on their source of operational energy. Active exoskeletons use motors, actuators, and pneumatic systems to generate assistive torques. This type of exoskeleton uses sensors to capture user and environmental information such as posture and relays it for processing by an onboard control system. The output of the control system is used to optimize the assistive torque generated by the actuators (Koopman, Toxiri, et al., 2019). Passive exoskeletons do not have any actuators or motors but rather use elastic materials such as dampers and springs to accumulate energy from the motion of the wearer's body segment. This stored energy is released to augment the action-specific muscles. A typical example of a passive exoskeleton is the on-body Personal Lift Assist

Device (PLAD), which has been shown to reduce EMG amplitude of the back extensor muscles by up to 27% (Abdoli-E, Agnew, & Stevenson, 2006). Due to predominant trends in work-related musculoskeletal disorders, exoskeletons can also be classified based on the body segment that it augments. The dominant ones are upper extremity exoskeletons and trunk exoskeletons.

2.3.1 Upper Extremity Exoskeletons

As the name suggests, upper extremity exoskeletons are designed to support the arms and shoulders in physical task performance. They are most suitable in overhead tasks, which involve flexed and/or abducted shoulder positions similar to what happens in automotive assembly lines or surgery.

2.3.1.1 Levitate Airframe®

The Levitate Airframe® is a passive lightweight exoskeleton designed to provide upper extremity support for tasks involving static elevated arm posture and/ or repetitive arm motion. It has two arm supports connected via two shoulder harnesses and curved rods to a padded upper back support. As shown in Figure 2.17, a length-adjustable vertical section connects the padded upper back support to a padded waist harness, which can also be adjusted to accommodate varying waist circumferences. This adjustability makes it easy to don and doff.

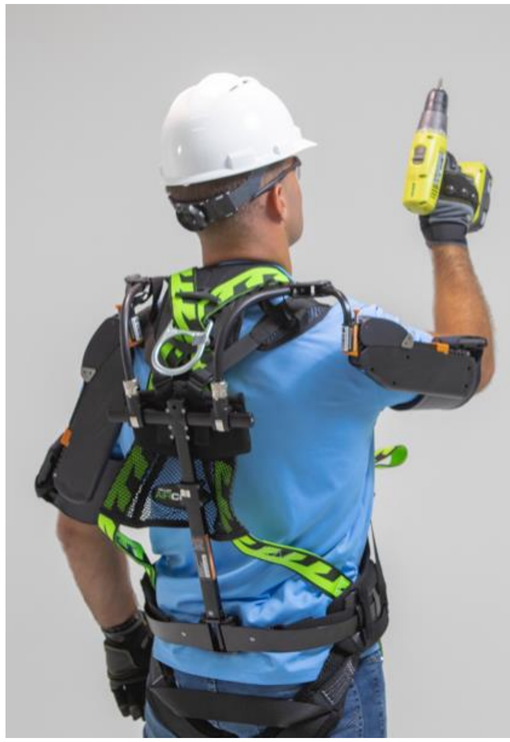

Figure 2.17: The Levitate Airframe (Wilson, 2017) Accessed 05 May 2020. Used with permission from Levitate Technologies Inc.

The Airframe works by evenly redistributing the weight on the shoulder muscles to the outside of the hips. It does this by using a progressive activation mechanical support system that activates based on upper arm elevation in that the degree of support increases as a function of shoulder elevation and abduction. This system deactivates when the arm is in a neutral posture, and the manufacturer claims that it reduces upper arm exertion levels up to 80%. Furthermore, a simulated automobile and surgical study reported significantly less shoulder fatigue/ pain perception when using the exoskeleton (Spada, Ghibaudo, Gilotta, Gastaldi, & Cavatort, 2017).

2.3.1.2 Eksovest

The EksoVest® was designed by Ekso Bionics, a company that designs and manufactures powered exoskeletons bionics for use in military and rehabilitation. Conceptually, the design of EksoVest® is similar to the Levitate Airframe (Kim, Nussbaum, Mokhlespour

Esfahani, Alemi, Alabdulkarim, et al., 2018). However, the main difference is how the upper arms are held in the support structure. While the Levitate Airframe uses an expandable semi-cylindrical structure that fits the upper arm's circumference, The EksoVest uses straps to keep the upper arm in position.

In a two-part study involving a simulated overhead drilling task, the EksoVest reduced normalized EMG of the shoulder muscles by approximately 45% and decreased task completion time by about 20% (Kim, Nussbaum, Mokhlespour Esfahani, Alemi, Alabdulkarim, et al., 2018). Furthermore, it reduced the shoulder range of motion by approximately 10% (Kim, Nussbaum, Mokhlespour Esfahani, Alemi, Jia, et al., 2018); nonetheless, the number of errors incurred during task performance with the exoskeleton increased substantially. The authors recommended further testing, particularly for high-precision tasks.

2.3.2 Trunk Exoskeletons

As the name suggests, trunk exoskeletons are designed to support the back during repetitive lifting and quasi-static awkward postures. While numerous exoskeleton designs from different manufacturers exist, the sub-sections below are devoted to describing the predominant ones on the market and which may be suitable for intervention in the operating room.

2.3.2.1 Laevo v series exoskeleton

The Laevo V series is a passive trunk exoskeleton that works on a loaded spring principle (Bosch, van Eck, Knitel, & de Looze, 2016). As shown in Figure 2.18, it consists of two pads in the chest area for comfort, two leg upper leg pads, and one back pad. Tubes with spring-like characteristics are used to connect the pads on both sides of the body. The exoskeleton is intended to transfer forces from the lower back to the chest and leg pads.

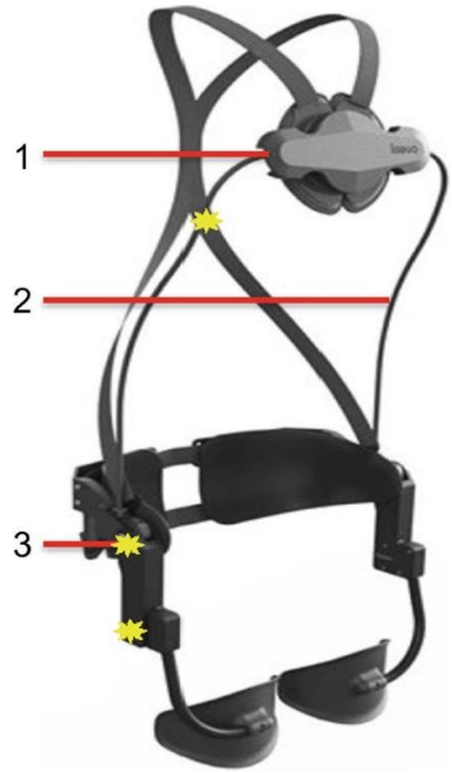

Figure 2.18: Laevo Exoskeleton. (Koopman, et al., 2019) Figure used with permission from Elsevier

The spring-like tubes partially resist trunk motion during forward bending, causing a build-up of elastic energy in them. At the onset of trunk extension, the stored elastic energy is released and applied to the chest and leg to support the trunk extensor muscles. This additional moment has been shown to reduce lower back muscle activity by a 35-38% in a quasi-static forward bending bent assembly task (Bosch et al., 2016) and also reduced peak compression force on the L5-S1 joint by 10% (Koopman, Kingma, de Looze, & van Dieën, 2019).

2.3.2.2 Personal lift assist device (PLAD)

The Personal Lift Assist Device (PLAD) is a passive exoskeleton modeled on the concept of human muscles by applying elastic bands that can be conceived as external muscle power generators (Abdoli-Eramaki, Stevenson, Reid, & Bryant, 2007). Unlike the Laevo exoskeleton, which uses a loaded spring to augment trunk extension, the PLAD uses energy stored in elastic components to augment trunk extension. It has six elastic elements that run over the erector spinae and hamstring muscles and are anchored at the shoulder and knee joints (see Figure 2.19). Four of the six elastic components run over the torso and are anchored to a shoulder strap and waist spacer, while the lower two elastic components connect the waist spacer and the knee strap. To accommodate asymmetric trunk postures, two of the four upper body elastic components are not symmetrically aligned with the trunk.

The PLAD derives its assistive power by storing energy in elastic components when an individual bends the trunk forward to lift a load, and the upper body elastic components are stretched. The energy is released during the trunk extension phase of lifting, reducing the biomechanical demand on the extensor muscles. In a series of studies, the PLAD was reported to have added 23-36Nm of assistive trunk extension torque (Abdoli-Eramaki et al., 2007), reduced lumbar and thoracic erector EMG by 14.4% and 27.6%, respectively in symmetric lifting task (Abdoli-E et al., 2006) and by 23.9% and 24.4% in asymmetric lifting (Abdoli-E & Stevenson, 2008).

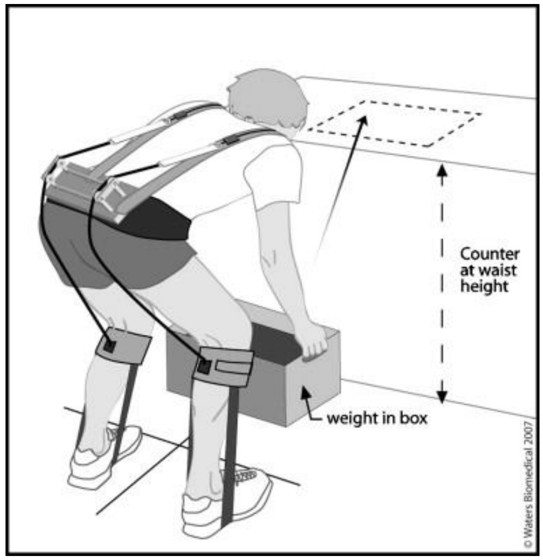

Figure 2.19: The Personal Lift Assist Device. (Lotz et al., 2009) Figure used with permission from Elsevier

2.3.2.3 Exosuit

The Exosuit (Figure 2.20) is a passive exoskeleton developed by a collaborative research partnership between the Assistive Robotics Lab of Virginia Tech and Lowe’s Innovation Lab (Change et al., 2019). Unlike the PLAD, this exoskeleton derives its assistive power from three sets of flexible beams of carbon fiber that run along the center of the back (42 beams) and behind the legs (14 beams each) of the user. To ensure connectivity of these bars, a mounting block is used to hold all three sets of beams at the waist level. This exoskeleton fits on wearers using waist belts with a padded lining, shoulder straps, and thigh pads, as shown in Figure 2.20. For adjustability to accommodate a range of stature and also the kinematic difference between the exoskeleton and human wearer, the shoulder harness and thigh pads are connected to the three sets of carbon fiber beams via low friction sliders.

Similar to the PLAD, when a wearer bends forward to lift a load, the carbon fiber beams flex with the torso, creating elastic energy storage in the beams. As the wearer extends the trunk during the concentric lifting phase, the stored energy is released to reduce the biomechanical demand on the back-extensor muscles. The manufacturers claim that greater than 95% of the stored energy returns at a neutral posture. In a simulated repetitive lifting study, this exoskeleton significantly reduced peak and mean EMG of back extensor muscles (iliocostalis erector spinae and longissimus erector spinae) and leg muscles, both for symmetric and asymmetric lifting (Alemi, Geissinger, Simon, Chang, & Asbeck, 2019).

Figure 2.20: The Exosuit by VT-Lowe, Inc. (Alemi, Geissinger, Simon, Chang, & Asbeck, 2019). Figure used with permission from Elsevier

2.4 Occupational Exoskeletons in the Non-surgical Environment

Exoskeletons in the environment other than the operating room have been targeted at repetitive lifting tasks and overhead work. Hence, the target of such studies and applications are geared towards reducing low back and shoulder pain. Interestingly, these two body areas happen to be the two body areas affected by musculoskeletal disorders among surgeons (Szeto et al., 2009; Wohlauer et al., 2019). So far, only passive exoskeletons have been explored in this space.

2.4.1 Upper Extremity Exoskeletons

Studies on the use of upper extremity exoskeletons in work task performance have shown promising results in reducing biomechanical demand on some upper extremity muscles and cardiac demand. For example, in repetitive lifting, carrying, and box stacking study, Theurel et al. (2018) reported significantly reduced deltoid anterior muscle activity during the lifting and stacking phase with the exoskeleton. However, the EMG of the triceps brachii muscles increased considerably during the carrying phase. They also reported that oxygen consumption was reduced by approximately 14% with the assistance of the exoskeleton, implying that participants needed less energy to perform the stacking task. Furthermore, in a simulated overhead drilling and light assembly task, it was shown that using an upper extremity exoskeleton reduced peak and median nEMG of the right deltoid by 38.4%, left deltoid by 24.5%, and 44.7% for the left trapezius. Collectively, the exoskeleton reduced shoulder muscle activity by 33.5% (Kim et al., 2018). Consequently, this reduction in shoulder EMG resulted in an average decrease in the task completion time of about 20%. Similar results have been reported in other upper extremity tasks (Kim et al., 2018; Alabdulkarim et al., 2019).

In addition to muscle activity, another area of focus for exoskeleton usage is upper extremity kinematics and task precision. In terms of precision, it appears that the impact of exoskeletons is nuanced. For instance, using an exoskeleton negatively impacted task precision in a simulated overhead task, and an attempt to increase precision while using the device resulted in increased upper extremity muscle activity (Alabdulkarim et al., 2019). However, in another study evaluating task precision, participants were instructed to extend their hands while wearing the Levitate passive upper limb exoskeleton and trace arches drawn between two marked points on a paper attached to a billboard (Spada, Ghibaudo, Gilotta, Gastaldi, & Cavatorta, 2018). Subjects were instructed to stop before feeling fatigued as a way to control for fatigue effects.

Their results showed that with the exoskeleton, the number of completed arches increased by 17.5%. Nonetheless, this study didn't report the errors while using the exoskeleton. Consequently, this study measured productivity instead of precision while using the exoskeleton. From a kinematic perspective, exoskeleton studies on upper extremity kinematics have shown reduced shoulder abduction range of motion in overhead drilling tasks (Kim et al., 2018) and lifting tasks (Theurel et al., 2018). Furthermore, using an exoskeleton increased elbow abduction angle in a lift and walking task but reduced elbow and shoulder abduction angle while stacking the load (Theurel et al., 2018). This altered kinematics has implications for joint loading; thus, requiring further studies if exoskeletons are to be implemented as intervention measures for work-related musculoskeletal disorders.

2.4.2 Trunk Exoskeletons

Studies on exoskeleton usage in repetitive lifting tasks have shown beneficial results in reducing biomechanical demand on back extensor muscles. For instance, in a symmetric and asymmetric repetitive lifting study, Alemi et al. (2019) reported that the VT-Lowe Exosuit (see Figure 2.20), regardless of the lifting technique, significantly reduced averaged peak trunk extensor muscle activation. Specifically, an average reduction of 31.5% and 29.3% in iliocostalis and longissimus erector spinae EMG, respectively, were reported. Even though the study was mainly focused on the trunk muscles, they found a substantial reduction in leg muscle (bicep femoris and vastus lateralis) EMG, with stoop lifting depicting the highest reduction compared to other lifting techniques in symmetric lifting (stoop peak=22.8%, mean=18.9%, squat peak=16.5%, mean=9%, freestyle peak=18.1%, mean=14.4%). Using the personal lift assist device (see Figure 2.19), Abdoli-E et al. (2006) reported similar but a lower reduction in lumbar (14.4%) and thoracic (27.6%) erector spinae activity, respectively. Another study of the PLAD reduced the severity of back muscle fatigue via a median frequency shift of 0.33%-0.41%

compared to 12%-26% without the PLAD (Lotz, Agnew, Godwin, & Stevenson, 2009). Interestingly, while Alemi et al. (2019) and Abdoli-E et al. (2006) both recorded a slightly increased abdominal muscle activity, statistical significance was reported by Alemi et al. (2019). They attributed this unexpected result to a high signal-to-noise ratio during the bending phase of the lifting when electrodes interacted with a waist belt and recorded significant noise. A similar reduction in low back muscle activity and fatigue has been reported in awkward quasi-static trunk posture (Bosch et al., 2016), leading to an increased time to fatigue, which in theory can lead to longer hours of task performance and increase productivity.

In addition to erector spinae muscle activity reduction, trunk kinematics and spinal loading have also been of interest to researchers. Using a dynamic spine EMG-assisted biomechanical model in a repetitive lifting task, the PLAD was found to reduce flexion-extension moment by 19.5% (Abdoli-E & Stevenson, 2008). While this reduction in extensor moment concurs with the reduction in trunk muscle activation from previous studies and possible reduction in trunk compression load, Picchiotti et al. (2019) reported contrary results. Their study compared the biomechanical loading of the lumbar spine between two passive trunk exoskeletons and found neither of the devices considerably reduced spine loading. This implied that the trunk extensor moment was not positively affected by the exoskeleton devices. The disparity between these two studies could be attributed to the operating principles of the exoskeletons used in the two studies. The PLAD stores elastic energy during trunk flexion and releases it during the concentric lifting phase to aid trunk flexion. In contrast, the exoskeletons tested by Picchiotti and colleagues were designed to keep the torso straight while using a squat lifting technique. Thus, participants in Picchioti’s study didn't benefit from an assistive moment while performing the lifting task.

Exoskeletons might have secondary effects that need to be addressed for successful implementation at the workplace. One study evaluated the effect that upper extremity exoskeletons might have on loading on the low back (Weston, Alizadeh, Knapik, Wang, & Marras, 2018). Interestingly, their results showed a significant increase in low back EMG, implying a potential shift of load from the upper extremities to the lower back. This unintended consequence may lead to work-related injury in body segments other than those subjected to strain to job demands. Thus, for the successful implementation of exoskeletons, potential secondary effects need to be addressed.

2.5 Occupational Exoskeletons in the Surgical Environment

The utility of exoskeletons in the surgical environment to reduce the long-term effects of work-related musculoskeletal disorders is relatively underexplored. This could be attributed to reasons outlined in a recent study (Cha, Monfared, Stefanidis, Nussbaum, & Yu, 2019). Their study surveyed a population of surgeons and other operating room staff and used content analysis to identify four themes that need to be addressed for the widespread adoption of exoskeletons. These themes are briefly discussed below.

Theme 1: Individual characteristics: This theme identified the perception of the operating room staff towards the implementation of ergonomic interventions. Specifically, most operating room staff considers WRMSDs as a part of the job. Thus, even though they are aware of the potential positive impact that exoskeletons will have on their long-term musculoskeletal health, they are not motivated to make efforts for its implementations. One participant from the Cha et al. (2019) study responded by saying, "I don't think we know what we are missing potentially. I've done it for eight [or] nine years without it". Hence, the implementation of exoskeletons would require a champion (attending surgeon) with some knowledge in ergonomics to lead the effort.

Theme 2: Perceived benefits: This theme identified the role that perceived advantages of using exoskeletons would have on long-term musculoskeletal health. The stakeholders stated that they expect exoskeletons to help retain workers and reduce early retirement, which has been linked to the detrimental effects of musculoskeletal disorders (Wohlauer et al., 2019). Furthermore, stakeholders pointed to the lack of formal ergonomic education in institutions as a reason for not realizing the potential benefits of exoskeletons. While they commended the role of informal safety programs such as fliers and annual online modules, they emphasized the need to include some ergonomics in such modules.

Theme 3: Environmental/ Societal factors: Perhaps this is the critical barrier that needs to be addressed from the design and manufacturing perspective. Stakeholders mentioned that sterility is a central theme in the operating room and that arm cuffs of exoskeletons should not extend beyond the wearer's elbow. Additionally, they were concerned about using the device across different operating room stuff and sterilizing the device between each user.

Also, functionality was of particular importance because study participants experienced frustration with other novel equipment breaking down. They mentioned that company representatives' availability to troubleshoot and fix the device would be crucial to implementation.

Theme 4: Intervention characteristics: This theme focused on the exoskeleton's functionality and interaction with the wearer (Usability). 93% of the stakeholders raised concern about the exoskeleton not hindering the motion of upper extremities in surgical tasks and the wearers' ability to move around freely. Furthermore, the device's weight and anthropometric fit, especially for extremely tall or extremely short individuals, was of primary concern under this theme. They also mentioned that donning and duffing will facilitate easy adoption.

2.5.1 Recent Studies on Exoskeletons in the Operating Room

Recent studies on the utility of exoskeletons in the operating room were targeted at reducing the biomechanical demand of the shoulder and low back due to awkward postures. From a review of the literature, only two studies were found (Albayrak et al., 2007; Liu et al., 2018).

2.5.1.1 Upper extremity exoskeleton in the operating room

The only exploration of the upper extremity exoskeletons in the operating room was from a study by (Liu et al., 2018). This was a three-phase prospective randomized study in which the effect of an upper extremity (Levitate Airframe®) was tested on laparoscopic task dexterity and subjective musculoskeletal discomfort. The first phase of the study documented the impact of wearing the exoskeleton on laparoscopic task dexterity using three standardized tests: the Minnesota Manual Dexterity Test (MMDT), three tests from the Purdue Pegboard Dexterity Test (PPDT), and Fundamentals of Laparoscopic Surgery (FLS). The second phase tested the exoskeleton's impact on subjective musculoskeletal pain while holding a laparoscopic camera in a quasi-static extended arm posture. Finally, the third phase evaluated the effect of using the exoskeleton on subjective musculoskeletal pain and fatigue while performing actual laparoscopic surgeries in the operating room.

Phase one of the study showed no statistically significant difference in completion time for both exoskeleton conditions. While this result has a positive implication for implementing exoskeletons in the operating room, this test didn't report task performance precision in the FLS task. Task completion time and task precision are two different variables that measure fundamentally different quantities. Task completion time measures productivity, while task precision measures the quality of work, and in a delicate task such as surgery, a higher task precision supersedes a short completion time. An inverse relationship exists between these two variables (Fitts, 1954). Furthermore, in a simulated overhead drilling task comparing three different exoskeleton types on physical demand and task precision, Alabdulkarim, Kim, & Nussbaum (2019) found that increasing the task precision lead to a significantly greater number of errors even while using an upper extremity exoskeleton. Clearly, the question of task precision while using an exoskeleton in the operating was not answered by this study. Thus, this requires investigation furthermore.

The results from phase two showed that the exoskeleton significantly reduced fatigue score compared to the control group. Furthermore, their results showed that after ten minutes of holding the laparoscopic camera, subjects wearing the exoskeleton experienced significantly lower arm and shoulder pain compared to those not wearing the exoskeleton. This result was expected and is consistent with findings from previous studies.

In phase two, phase three showed that wearing the exoskeleton while performing laparoscopic procedures significantly reduced perceived shoulder pain after a day of operation. Additionally, whole body composite pain score decreased by approximately 70% when wearing the exoskeleton.

While results from phases two and three show promising signs of using exoskeletons as an intervention tool for long-term musculoskeletal disorders, it must be realized that these results are subjective. Furthermore, upper extremity exoskeleton, in particular, might transfer load from the shoulders unto the lower back (Weston et al., 2018), thus increasing the likelihood of back pain in the long term. Thus, exoskeletons may reduce pain/ fatigue but might have secondary effects that need to be addressed for safe implementation. This issue needs to be addressed in further studies.

2.5.1.2 Trunk Postural Support in the Operating Room

From the review of the literature, two studies were found on the utility of trunk exoskeletons in the healthcare environment (Albayrak et al., 2007; Hwang et al, 2021). The first study (Albayrak et al., 2007) developed and tested the effect of ergonomic upper body support on low back muscle activation. While this is not considered an exoskeleton, it was a postural augmentation device intended to reduce loading on the back muscles. The postural assist system's design was motivated by awkward trunk posture assumed by some surgical team members due to non-optimal table height. Specifically, the height of the operating table is usually adjusted to fit the lead surgeon, leaving other members of the surgical team in an ergonomically poor posture.

Their postural support was designed to meet five criteria listed below

- Support the surgeon’s body in a natural working posture
- Usable both in open and minimally invasive surgical procedures
- Compact construction due to limited space in the operating for mobility
- Comfortably useful by both 5th percentile woman and 95th percentile man, all from the Dutch population
- Should be height adjustable
- Mobility via wheels
- Sufficient space for foot placement during electrosurgery

The prototype has height-adjustable chest support, which activates when the surgeons lean against it. This can easily be removed to create extra space during minimally invasive procedures. Furthermore, there is semi-standing support incorporated into the platform for use when performing minimally invasive procedures.

To test the efficacy of the postural support device, two tests were conducted. The first test was EMG of back muscle (right erector spinae) and lower extremity muscles (semitendinosus and gastrocnemius muscles) in a simulated surgical task, with and without the postural assist device. The second was a questionnaire survey from seven surgeons providing subjective judgment on their comfort level while using the postural assist device.

The EMG study showed that as a percentage of maximum voluntary contraction (MVC), the postural assist device reduced erector spinae activity by 40% and 48% at 15° and 20° trunk flexion, respectively. In addition to the EMG results, the subjective questionnaire respondents indicated that approximately 85% of participants rated postural support as comfortable to use.

While the device seems to provide a means to reduce the biomechanical load on the back muscles during awkward postures, the study results have certain limitations that need to be addressed. For example, no statistical conclusion can be drawn on the EMG data obtained with a single participant. Besides, for open surgery, requiring some dynamic trunk movement, this postural support might restrict movement. This restriction was reported by 50% of the participant in the questionnaire survey.

The second recently published study by (Hwang et al., 2021) tested the effect of three commercially available on back muscle activation and usability during patient transfer. Their results showed that the trunk exoskeletons averagely reduced lumbar EMG by 11.2% and garnered positive usability results from the participants. However, they noticed that the benefit derived from the exoskeleton was dependent on the patient transfer technique as they observed a significant interaction between those two variables.

Collectively, these two studies point to the potential beneficial effects of exoskeletons when introduced into the healthcare field. However, further research is required to understand task/posture-specific benefits so that standardized deployments protocols can be developed to maximize the benefits.

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CHAPTER 3. INTRAOPERATIVE POSTURE AND WORKLOAD ASSESSMENT IN VASCULAR SURGERY

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Abstract

Quantifying the workload and postural demand on vascular surgeons provides valuable information on the physical and cognitive factors that predispose vascular surgeons to musculoskeletal pain and disorders. The aim of this study was to quantify the postural demand, workload, and discomfort experienced by vascular surgeons and to identify procedural factors that influence surgical workload. Both objective (wearable posture sensors) and subjective (surveys) assessment tools were used to evaluate intraoperative workload during 47 vascular surgery procedures. Results demonstrate unfavorable neck and low back postures as well as high

pain scores for those body segments. Additionally, workload from subjective surveys increased significantly as a function of operative duration, and mental workload was high across all procedure types. Neck postural risk exposure and physical demand were among the variables that increased with surgical duration, procedure type, and loupes used by the surgeons. Correlations among postural angles and pain scores showed consistency between the objective assessment and the subjective surveys for neck and trunk. The authors believe that the results of this study notify the need for developing preventive measures such as ergonomic interventions for vascular surgery.

3.1 Introduction

The field of vascular surgery is rapidly evolving due to the advancements in treatment techniques and new technology. For instance, breakthroughs in tissue-engineered blood vessels (TEBV) has led to the development of artificial blood vessel replacements for damaged ones (Row et al., 2017). The invention of advanced imaging techniques (Bredahl et al., 2017; Fillinger, 1999; Garrett et al., 2003) coupled with catheter-based technologies such as angioplasty has changed the approach of most open vascular procedures to minimally invasive therapies (Veith, 2016); while vascular procedures that are impossible to complete minimally invasively are completed via hybrid models (a combination of open and minimally invasive approaches) (Balaz et al., 2012). These advances improve cosmetic and surgical outcomes (Weiss and Lumsden, 1999) and also shorten recovery time significantly. However, the implications for the surgeon's workload and posture when moving from open to minimally invasive approaches, as well as the use of high-tech vessels with their attendant endovascular surgical techniques are unknown.

Technological advancements allow vascular surgeons to perform ever-increasingly complex surgical procedures which enhance patient outcomes. These new advancements will potentially increase surgeon's mental and physical demand, placing them at risk of physical pain and development of musculoskeletal disorders (MSDs). While previous reports on musculoskeletal disorders have focused mostly on general surgeon populations and subspecialties other than vascular surgery (Davis et al., 2014; Dianat et al., 2018; Epstein et al., 2018), two recent surveys have highlighted the plight of vascular surgeons. The first survey by Davila et al. (Davila et al., 2019) on 263 clinical vascular surgeons (United States Society of Clinical Vascular Surgery) reported that approximately 90% of their surgical cases are carried out in a standing position, and that physical discomfort was a significant concern; 54% of survey respondents reported that the pain would eventually impede their capability to perform surgery in the future. The second survey on 775 respondents (United States Society of Vascular Surgery) pointed out moderate (3.9 ± 2.4 out of a max 10) overall pain following a full day of surgery and that 37% of the respondents had sought medical care for their work-related pain (Davila et al., 2019; Wohlauer et al., 2019). Additionally, results from a prospective study highlighted the severe risk (and associated physical pain) in the cervical region for attending vascular surgeons (Davila et al., 2018). Specifically, this study pointed out that the cervical area is at high to severe risk for 67% of operative time. These studies cumulatively point to the physical discomfort and pain experienced by vascular surgeons during practice and the potential detrimental effect on surgeons' careers. It is interesting to note that physical pain may impact cognitive attention resources (Stephenson et al., 2020) and decision-making (Koppel et al., 2017), potentially affecting surgical outcome. Thus, considering the forecasted shortage of vascular surgeons in the United States (Way, 2010), work-related musculoskeletal pain and surgeon workload needs to be addressed.

Workload can be conceptualized as a multi-factorial term that describes the “cost” (physical, mental, temporal demands, etc.) of performing a certain task (Hart, 2006). The subject of workload has recently received some attention across various surgical subspecialties using a variety of assessment tools. One such study was by Lowndes et al., (Lowndes et al., 2020) which used a combination of the subjective National Aeronautics and Space Administration Task Load Index (NASA-TLX) (Hart and Staveland, 1988) and a question from the Surgical Task Load Index (SURG-TLX) (Wilson et al., 2011) to evaluate operative workload across different subspecialties of surgery. Their results revealed that in 22% of the 662 surgeries surveyed, the surgeons perceived the surgical tasks to be more difficult than expected, and this difficulty correlated with the duration of the surgery. Their results also noted a strong positive correlation between mental and physical demand, and surgeon-perceived performance was rated as poor in unexpectedly difficult cases (Lowndes et al., 2020), highlighting the need to reduce the procedural difficulty to boost performance and positively impact surgical outcomes. Additionally, Davila et al., 2018’s postural analysis of 34 cases from 13 vascular surgeons revealed the demanding nature of 83% of the surgeries performed, particularly coupled with awkward postures, such as the one shown in Figure 3.1 (Davila et al., 2018). These awkward and static postures put immense stress on the back, shoulder, and neck muscles; thus, not only increasing work demand but also increasing the risk of work-related MSDs in the long term. Another workload study by Yu et al., (Yu et al., 2016) used validated wearable sensors (Morrow et al., 2017) called Inertial Measurement Units (IMUs) to obtain objective and real-time torso and upper extremity posture of 13 surgeons and 13 assistants while performing robotic prostatectomy. Participants also completed a postoperative SURG-TLX workload body part discomfort questionnaire. Their results illustrated the significant mental demand (40% more) at the surgical robot's console and the physically demanding neck posture required of the assisting surgeon. This study highlighted the capability of combining both subjective workload tools (NASA-TLX and SURG-TLX) and objective measures (postural data from IMUs) to identify potential ergonomic issues during surgery (Yu et al., 2016).

Figure 3.1: A typical awkward posture in vascular surgery

With the rapid procedural and technological advancements made in the field of vascular surgery, coupled with the previous reports of musculoskeletal concerns (Davila et al., 2019; Davila et al., 2018; Wohlauer et al., 2019), quantifying the intraoperative workload and postural demand on vascular surgeons will provide valuable information on the physical and cognitive factors that predispose vascular surgeons to musculoskeletal pain and disorders. Such information will be vital to ergonomists in identifying aspects of the surgical workflow that can be targeted for intervention as an effort to minimize workload and the occurrence of musculoskeletal disorders in vascular surgeons. Therefore, this intraoperative study was undertaken to quantify the workload and discomfort and identify procedural factors that influence vascular surgeons' workload.

3.2 Materials and Methods

3.2.1 Participants

Sixteen vascular surgeons (13 males and 3 females) from a large quaternary academic hospital system participated in this study; 12 were attendings, three were fellows, and one surgeon was a resident. From the 47 data collections for this study, 22 were completed on one campus by nine different surgeons, and 25 were completed on the other campus by seven different surgeons. The number of data collections for each surgeon varied from one to five sessions, with an average of three surgeries per surgeon. The mean ± standard deviation (±SD) of the participants' essential anthropometric characteristics is as follows: age 46 ± 13 (years), height 176.5 ± 7.8 (cm), and weight 78.9 ± 15.9 (kg).

3.2.2 Data Collection Instrumentation

Both subjective (surveys) and objective (wearable posture sensors) assessment tools were used to evaluate different facets of workload in vascular surgeries. The subjective data included two parts. The first part was a body map survey based on standardized Nordic questionnaires (Kuorinka et al., 1987) to evaluate discomfort/pain in surgeons’ neck, right and left shoulders, upper and lower back, as well as right and left wrists/hands, with scores from zero to ten, where zero represented no discomfort/pain, and ten represented extreme discomfort/pain. Also, the Borg CR-10 Scale was used to assess each participant's subjective fatigue level (zero to ten) (Borg, 1982). The discomfort/pain and fatigue questions (SURVEY-1) were completed both before and after the surgical procedure. The differences between the post- and pre-surgical procedure data were calculated and considered as "change in pain score" and "change in fatigue score."

The second part of the subjective assessment was a modified survey (SURVEY-2) based on SURG-TLX (Wilson et al., 2011) (Wilson et al., 2011) and a Global Operative Assessment of

Laparoscopic Skills (GOALS) question (Vassiliou et al., 2005) that was completed only after the surgical procedure was completed. This modified survey contained five subscales (mental demand, physical demand, complexity, distraction, and difficulty) with each subscale ranging from 0 to 20 (Yu et al., 2016).

IMUs were used for objective assessment of surgeon posture for the duration of the surgical procedures. IMUs were attached to the back of the head, upper back, right upper arm, and left upper arm, as shown in Figure 3.2. The IMU (Opal™, APDM, Inc., Portland, OR, USA) used in this study contained an accelerometer, gyroscope, and magnetometer and was 4.8 x 3.7 x 1.4 cm in size. The postural data was collected at a frequency of 128 Hz. These data were used to calculate postural angles and postural risk scores for the neck (on the basis of head segment angles), trunk, right and left shoulders (on the basis of upper right and left arm segments) appertaining to Rapid Upper Limb Assessment (RULA) (McAtamney & Nigel Corlett, 1993).

3.2.3 Experimental Tasks

This study was approved by the Institutional Review Board (IRB), and all participants were vascular surgeons. The data collections were performed in the operating rooms (ORs) of two large academic hospitals in the same hospital system. Before the surgeons entered the OR, a study team member helped the surgeon don the IMUs. The upper back and upper arm sensors were placed in pockets and were securely attached to the surgeon’s scrubs by folding the scrub shirt at these locations to minimize motion of the sensors while ensuring the comfort of the surgeon and the placement and security of the IMU with respect to the surgical field for sterility. The head sensor was taped to the surgeon’s headlight or placed in the pocket of an elasticized headband worn by the surgeon if they did not use a headlight (Figure 3.2). The IMUs were calibrated using a neutral reference posture where the surgeons were asked to stand up straight, facing forward with arms by the sides of the body, fingers pointing downward, and the palms of the hands facing toward the body. Postural parameters were all calculated relative to this "neutral posture." Accelerometer, gyroscope, and magnetometer data were recorded by these IMU sensors during the entire surgical procedure from surgical incision to closure which was the duration of the surgery.

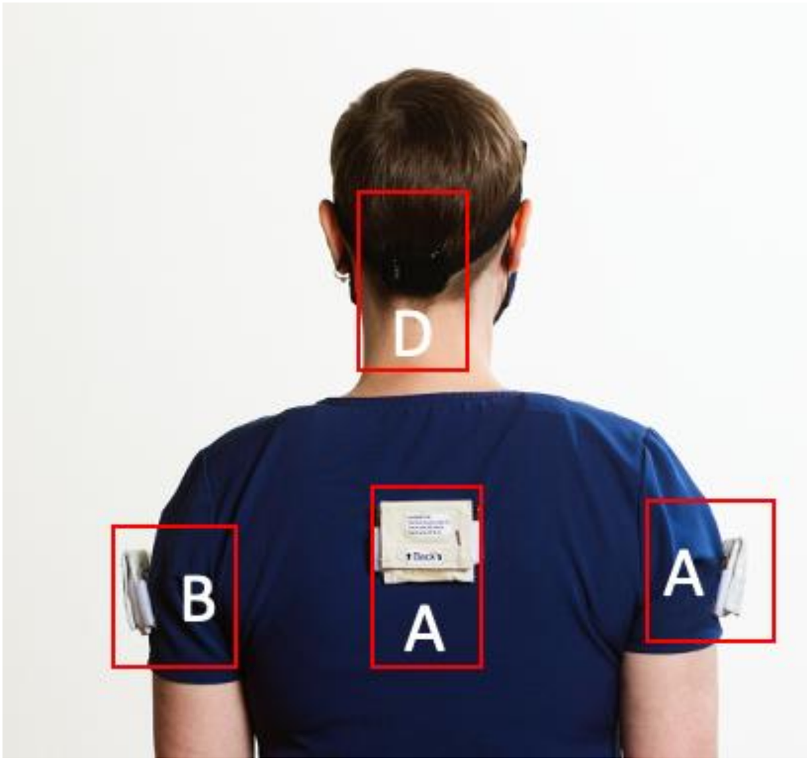


Figure 3.2: IMUs worn by the surgeon. A) Upper back sensor, B) Left upper arm sensor, C) Right upper arm sensor, D) Head sensor

Each surgeon was asked to complete SURVEY-1, which surveyed body parts pain/discomfort and general fatigue prior to surgery. At the end of the surgical procedure, the IMUs were removed, and the surgeon was asked to complete SURVEY-1 for during and after surgery in addition to SURVEY-2. The details of the surgery, including duration, procedure type, and adjunctive equipment used, (such as lead, headlight, and loupe magnification glasses) were recorded for each surgical procedure.

3.2.4 Experimental Design

3.2.4.1 Independent variables

The independent variables in this study were: 1) procedure type, 2) surgical duration, 3) usage of lead as radiation personal protective equipment (RPPE), and 4) usage of loupes by the surgeon. The procedure type had two different levels: open surgeries and non-open surgeries. The non-open surgeries included both endovascular and hybrid surgical procedures. The median duration of all vascular surgeries (240 minutes) was used to split all surgical procedures into two groups. Surgeries with durations higher than the median were considered as long surgeries and vice versa. Thus, the surgical duration had two levels: long and short surgeries. The usage of RPPE had two levels: with and without RPPE. The usage of loupes had two levels: with and without loupes during the surgery.

3.2.3.2 Dependent variables

The dependent variables were: 1) the average angles of neck, trunk, as well as right and left shoulders during surgery calculated from the IMU data; 2) the neck, back, along with right and left shoulders postural risk scores calculated from the postural angles based on RULA (neck risk score, trunk risk score, right shoulder risk score, and left shoulder risk score) (McAtamney and Nigel Corlett, 1993) and combined with a time-weighted average; 3) the change in pain scores between post- and pre-surgical procedures for neck, upper back, lower back, right shoulder, left shoulder, and right and left wrists/hands; 4) the change in the fatigue score between post- and pre-surgical procedures; and 5) the modified SURG-TLX workload subscales, including mental demand, physical demand, complexity, distraction, and difficulty.

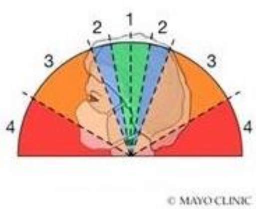
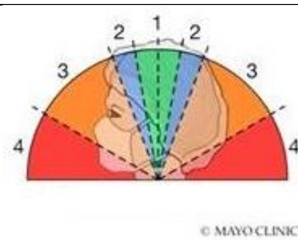
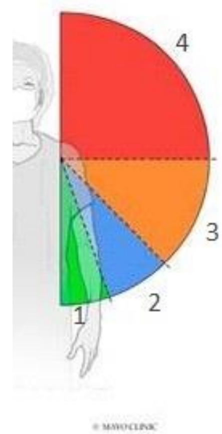
3.2.3.3 Data processing and statistical analysis

Postural data from the IMU sensors were downloaded in Motion Studio software (APDM, Inc., Portland, OR, USA). The data were combined using a custom MATLAB program (R2015b, Mathworks Inc., Natick, MA USA), and the body segment postures and risk levels were defined from the IMUs (Morrow et al., 2017). The angle between the superior-inferior axis of the head at each moment during the surgery, relative to the superior-inferior axis of the head in "neutral posture," was defined as the neck angle at that moment. A similar procedure was employed for upper back as well as right and left upper arm sensors to define the trunk angle in addition to the right and left shoulders angles, respectively. At that juncture, these angles were used to define postural risk scores for neck, trunk, right and left shoulders. For each of the four IMUs, the range of the angle was stratified to four levels. These four levels were defined based on RULA (McAtamney and Nigel Corlett, 1993) (McAtamney and Nigel Corlett, 1993), as shown in Table 3.1. For example, when the neck angle is between 0º and 10º, it was considered as level 1 for neck; when the neck angle is higher than 60º, it was considered as level 4 for the neck. The percentage of time that the neck angle was held in each one of these four levels for the entire surgical procedure was considered as score level 1 (SL1), score level 2 (SL2), score level 3 (SL3), and score level 4 (SL4). These were combined into a total time-weighted, postural risk score for neck defined according to Equation 1.

$$Postural\ Risk\ score\ =\ [4*(SL4)+\ 3*(SL3)+\ 2*(SL2)+\ 1*(SL1)]---\ (1)$$

A similar technique was used to calculate the risk scores for the trunk and shoulders, respectively. The average angles for neck, trunk, right and left shoulders during each surgery were calculated and used for that surgery as neck angle, trunk angle, right shoulder angle, and left shoulder angle, respectively. "Change in pain score" and "change in fatigue score" were negative, zero, or positive values when post-surgical, self-reported scores were compared to the pre-surgical scores, and they represented decrease, no change, and increase in these scores, respectively. In order to obtain a better understanding of the "change in pain score" and "change in fatigue score," these pain scores were categorized into three groups: (A) Pain score 0 or less when the difference showed a decrement or no change in the score, (B) Pain score 1 and 2 when the difference showed an increment of 1 or 2 scores, and (C) Pain score 3+ when the difference showed an increment of 3 or more (Yang et al., 2020); the latter group showed a clinically relevant pain differential.

Table 3.1: Joint angles and their risk-scores

	Neck	**Trunk**	**Right/Left shoulder**
			
Level 1	>0º & <10º	>0º & <10º	>0º & <20º
Level 2	>10º & <20º	>10º & <20º	>20º & <45º
Level 3	>20º & <60º	>20º & <60º	>45º & <90º
Level 4	>60º	>60º	>90º

The effects of each independent variable list on the dependent variable list were analyzed in a mixed-effect model. In this model, the fixed effect was the desired independent variable, and the random factor was always the participant. Also, the least square means (LSM) for dependent variable list were calculated. Another way to look at the effects of each independent variable on the dependent variables is to look at the odds ratios. The odds ratios were calculated for each independent variable separately according to Equation (2), where the definitions (categorization) of exposed and unexposed are shown in Table 3.2.

$$Odds\ ratio\ = \frac{(Exposed\ Cases * Unexposed\ Controls)}{(Unexposed\ Cases * Exposed\ Controls)} --- (2)$$

In this equation, the exposed/unexposed groups represent the two different levels of each independent variable. Also, the dependent variables were split into two groups named as cases and controls. Specifically, a change in pain and fatigue scores of at least three were considered as cases and vice versa. Furthermore, the SURG-TLX subscales of greater than 10 were classified as cases and vice versa.

Table 3.2: The definitions for exposed and unexposed groups as well as case and control groups for odds ratios calculations

Independent variables		
Group	Exposed	Unexposed
Procedure type	Open	Non-open
Surgical duration	Long	Short
Usage of RPPE	With RPPE	Without RPPE
Usage of loupes	With Loupes	Without Loupes
Dependent variables		
Group	Case	Control
Change in pain scores	≥3	<3
Change in fatigue score	≥3	<3
Modified SURG-TLX	>10	≤10

Spearman's rank correlation coefficient was used to investigate the correlation between subjective and objective measurements. All the statistical analyses were done in JMP Pro 14 software package (SAS, Cary, NC).

3.3 Results

3.3.1 Summary Statistics

The summary of the upper body angles, upper body risk scores, and modified SURG-TLX subscales are shown in Figures 3.3-3.5 using quantile plots. These quantile plots illustrate 25^{th}, 50^{th}, and 75^{th} percentiles of the data in addition to the minimum and maximum values. Also, the mean (±SD) was calculated for these variables. The neck was held in a mean (SD) postural angle of 37.1º (±12.7º), while the trunk angle was 18.1º (±6.7º). The postural angle for the right shoulder was 21.4º (±5.5º), and the angle for the left shoulder was 19.1º (±4.5º). The mean (±SD) risk score for the neck was 2.8 (±0.4), while the risk score for the trunk was 2.1 (±0.4), and the risk scores for right and left shoulders were 1.5 (±0.2) and 1.4 (±0.2), respectively (Range from 1.0 min-4.0=max). The modified SURG-TLX mean (±SD) for its five subscales were 14.8 (±4.6) for mental demand, 12.1 (±5.1) for physical demand, 15.3 (±5.0) for complexity, 5.0 (±4.0) for distraction, and 13.8 (±4.8) for difficulty (Range from 0-20=max).

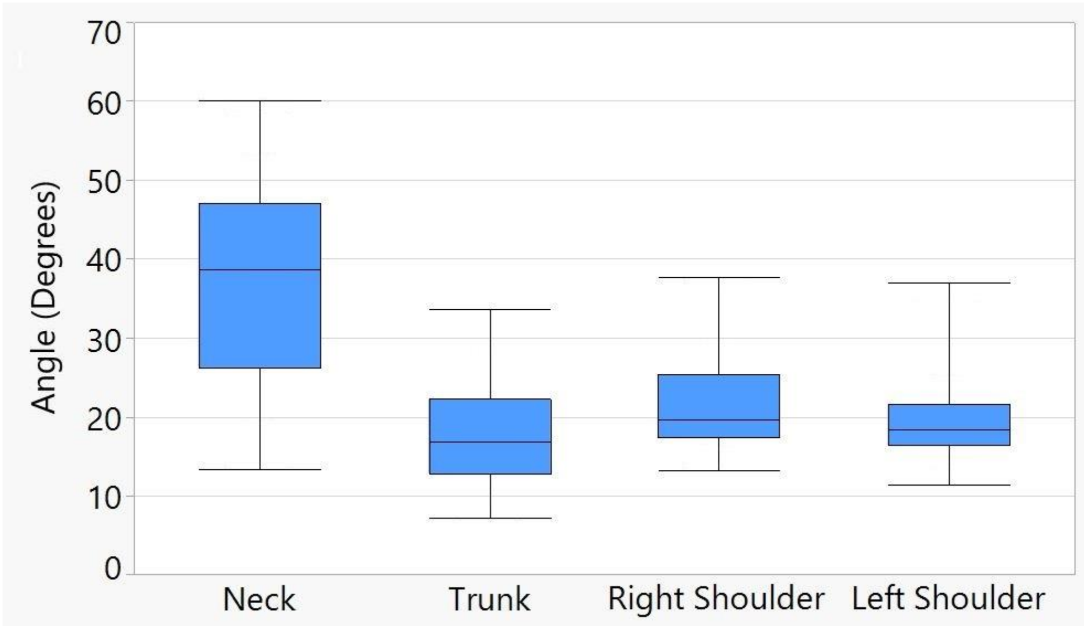


Figure 3.3: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of postural angles for upper body parts calculated from IMU data

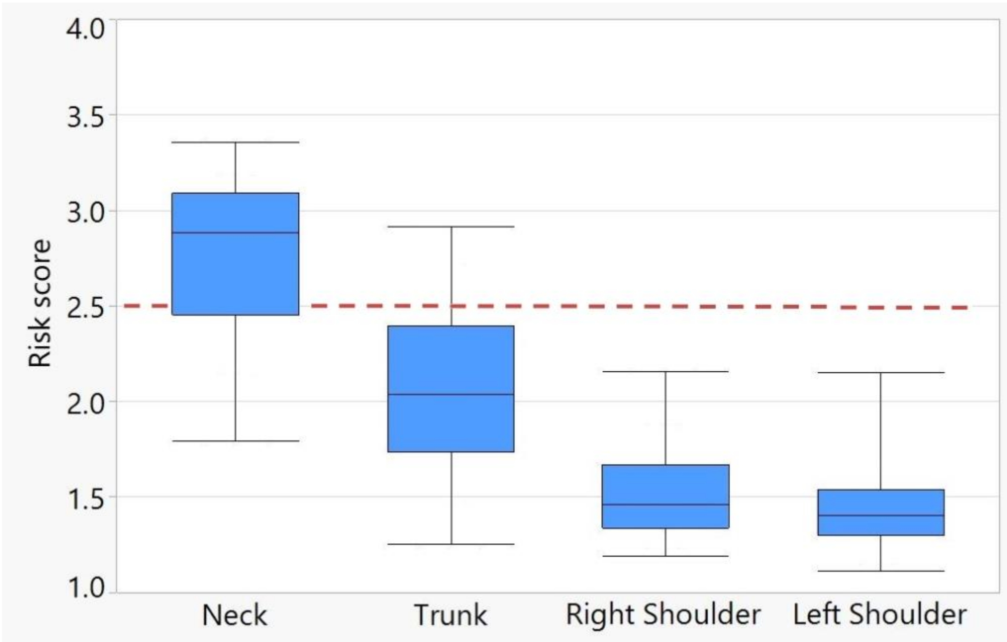


Figure 3.4: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of risk scores for upper body parts calculated from IMU data (Range from 1.0-4.0). The dashed line represents the midpoint of this range.

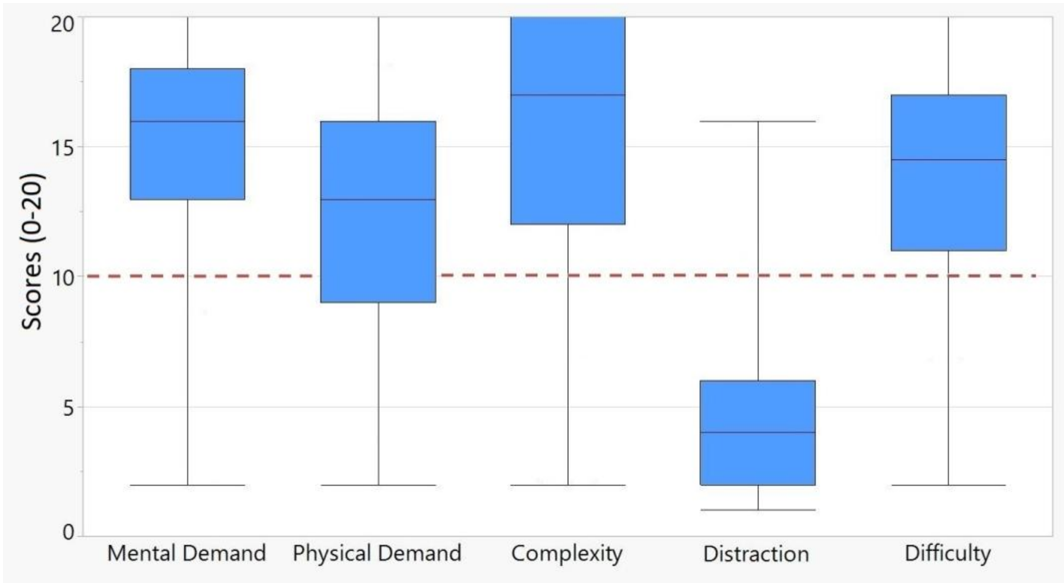


Figure 3.5: 25th, 50th, and 75th percentiles (middle boxes) and the minimum and maximum values of scores from modified SURG-TLX subscales (Range from 0-20=max). The dashed line represents the midpoint of this range.

Categorized "Change in pain score" and "change in fatigue score" by percentage of cases among all surgeries are illustrated in Figure 3.6. The percentage of the surgical operations where the fatigue/pain scores increased by three or more (self-reported pain score of 3+ in Figure 3.6) are 30% for overall fatigue, 36% for the neck, 11% in the right shoulder, 13% in the left shoulder, 30% in the upper back, 30% in the lower back, 4% in the right wrists/hands, and 4% in the left wrists/hands.

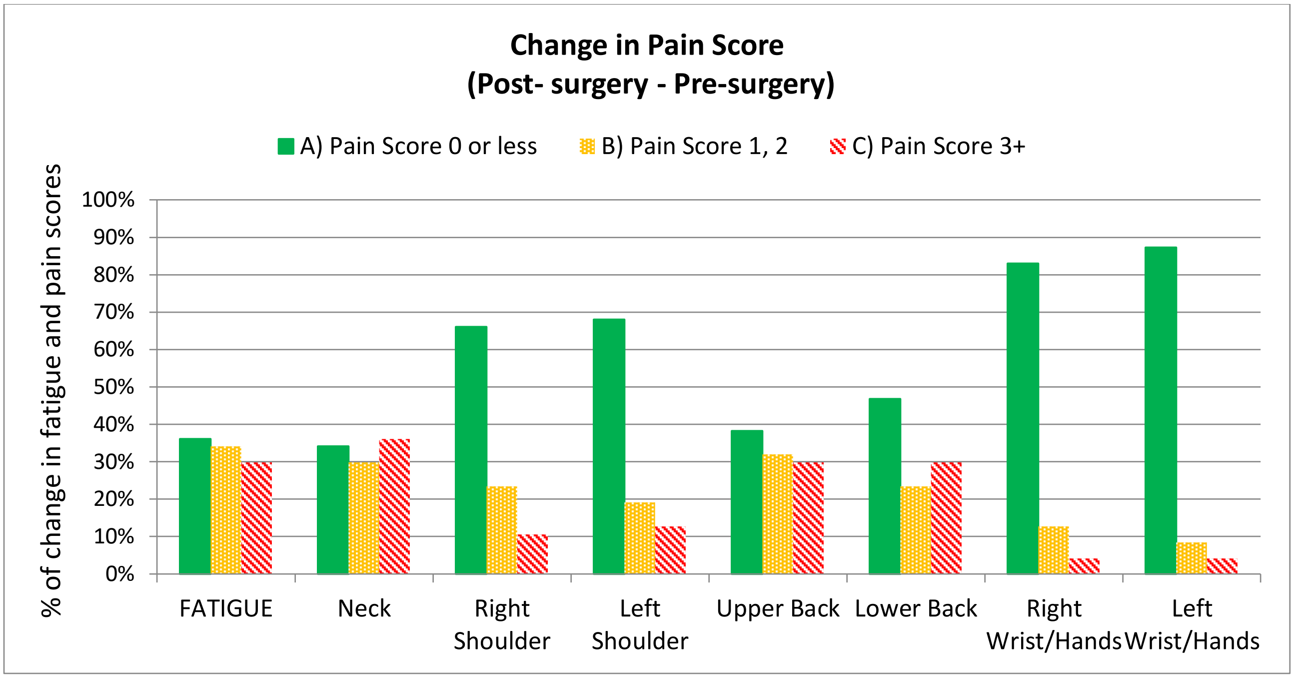


Figure 3.6: The percentage of the three levels of change in fatigue and pain scores for different body parts including neck, right shoulder, left shoulder, upper back, lower back, and right and left wrists/hands

3.3.2 Difference Between Open and Non-Open Surgeries

Thirty-five surgical procedures (74%) were open surgeries. Statistically significant differences were found for neck angle ($p<0.0001$) and right shoulder angle ($p=0.0403$) between open and non-open surgeries. In open surgeries, the LSM for neck and right shoulder angles were, respectively, 54% and 22% larger than their equivalent in non-open surgeries. The effect of the procedure type (open vs. non-open) on risk scores was significant for neck risk score ($p<0.0001$). The LSM of the neck risk score for open surgeries was 24% larger than for non-open surgeries. Over the modified SURG-TLX subscales, a significant effect of procedure type (open vs. non-open) was found only for physical demand ($p=0.0011$). The LSM of the physical demand increased by 69% for the open surgeries in comparison with the non-open surgeries (Table 3.3). The odds ratio [95% CI] for physical demand was 5.6 [1.4-23.1], quantifying how the open surgeries were associated with more cases of physical demand>10 compared to non-open surgeries (Table 3.2). No significant effects were found for other variables of postural angles, risk scores, and modified SURG-TLX subscales. Also, no significant differences were found for the change in fatigue and pain scores between open and non-open surgeries.

Table 3.3: Least square mean (± standard error) for dependent variables with statistically significant differences between open and non-open surgeries

Procedure Type	**Open surgery**	**Non-open surgery**
Neck Angle	41.6º (±2.3º)	27.1º (±3.0º)
Neck Risk Score	2.9 (±0.1)	2.4 (±0.1)
Right Shoulder Angle	22.6º (±0.8º)	18.5º (±1.5º)
Physical Demand	13.4 (±0.8)	8.0 (±1.3)

3.3.3 Difference Between Surgeries With and Without Loupes

The surgeons used loupes during 29 surgeries (62%). Statistically significant effects were found for both neck and trunk angles ($p=0.0116$ and $p=0.0119$, respectively) as well as for neck and trunk risk scores ($p=0.0039$ and $p=0.0042$, respectively) comparing surgeries where loupes were used versus not used. Increases of 24% in LSM of neck angle, 27% in LSM of trunk angle, 13% in LSM of neck risk score, and 16% in LSM of trunk risk score were found for the surgeries where the surgeon used loupes compared to those where loupes were not used. Consistent with these results, the effect of loupes was found to be significant for the change in the subjective neck pain score ($p=0.0227$). In surgeries where loupes were used, the reported neck pain score increased 110% more than surgeries without loupes. The odds ratio [95% CI] for change in neck pain was 4.7 [1.1-19.6], showing that surgeries with loupes were associated with more cases of change in neck pain ≥3 compared to surgeries without loupes (Table 3.2). Also, the LSM of physical demand was 37% larger for surgeries with loupes, which describes the significant effect of loupes on this variable ($p=0.0214$) (Table 3.4). The odds ratio [95% CI] for physical demand was 4.8 [1.3-18.2], indicating that surgeries with loupes were associated with more cases of physical demand >10 compared to surgeries without loupes (Table 3.2). No significant effect was found for other variables of postural angles, risk scores, change in fatigue and pain score, and modified TLX subscales.

Table 3.4: Least square mean (± standard error) for dependent variables with statistically significant differences between surgeries with and without loupes

Usage of Loupes	**With Loupes**	**Without Loupes**
Neck Angle	41.0º (±3.0º)	33.1º (±3.3º)
Neck Risk Score	2.9 (±0.1)	2.6 (±0.1)
Trunk Angle	20.7º (±1.5º)	16.3º (±1.7º)
Trunk Risk Score	2.2 (±0.1)	1.9 (±0.1)
Change in Neck Pain	2.8 (±0.5)	1.3 (±0.6)
Physical Demand	13.6 (±1.0)	9.9 (±1.3)

3.3.4 Difference Between Surgeries With and Without RPPE

In 16 surgeries (34%) out of 47 data collections, RPPE was worn by the surgeon for at least part of the surgical procedure. A statistically significant effect was found only for neck angle (p=0.0005), neck risk score (p=0.0001), and distraction (p=0.0015) among all the variables of postural angles, risk scores, change in fatigue and pain scores, and modified SURG-TLX subscales. The LSM of neck angle and neck risk score were surprisingly smaller (26% and 14%, respectively) in surgeries where RPPE was used. The LSM for the variable of distraction had an increase of 90% when RPPE was used in comparison with surgeries without wearing RPPE (Table 3.5). The odds ratio [95% CI] for distraction was 10.0 [1.0-98.9], showing that surgeries with RPPE were associated with more cases of distraction compared to surgeries without RPPE (Table 3.2).

Table 3.5: Least square mean (± standard error) for dependent variables with statistically significant differences between surgeries with and without RPPE

Usage of RPPE	**With RPPE**	**Without RPPE**
Neck Angle	30.9º (±3.0º)	41.7º (±2.6º)
Neck Risk Score	2.5 (±0.1)	2.9 (±0.1)
Distraction	7.7 (±1.0)	4.0 (±0.9)

3.3.5 Effects of the Surgical Duration

Twenty-three surgeries (49%) were longer than the median duration of all vascular surgeries (240 minutes) and thus they were considered as long surgeries. None of the postural angles was significantly affected by the surgical duration. However, only the neck risk score ($p=0.0469$) was significantly affected by the surgical duration among all the risk scores. The LSM of neck risk score increased by 8% for the long surgeries compared to short surgeries. On the other hand, change in fatigue and pain scores and modified SURG-TLX subscales were more sensitive to surgical duration. The LSM of the change in fatigue ($p=0.0174$) and change in lower back pain score ($p=0.0432$) both increased significantly by 100% and 116%, respectively, for long surgeries in comparison with short ones. No significant effect was found for other pain scores. The odds ratios [95% CI] for increase in fatigue and lower back pain were 3.8 [1.0-14.9] and 2.4 [0.7-8.9] during long surgeries, respectively. These odds ratios quantify how long surgeries were associated with more cases of change in fatigue ($\geq$3) and change in low back pain ($\geq$3) compared to short surgeries (Table 3.2). For modified SURG-TLX subscales, surgical duration had a statistically significant effect on mental demand ($p=0.0005$), physical demand ($p<0.0001$), complexity ($p=0.0003$), and difficulty ($p=0.0001$). These significant increments in LSM were 36% for mental demand, 61% for physical demand, 39% for complexity, and 47% for difficulty during long surgeries compared to short ones (Table 3.6). Also, the odds ratios [95% CI] for mental demand, physical demand, and difficulty were 9.1 [1.0-80.8], 10.5 [2.0-55.0], and 15.7 [1.8-136.5], respectively. These odds ratios show that long surgeries were associated with more cases of mental demand >10, physical demand >10, and complexity >10 compared to short surgeries (Table 3.2). For complexity, all surgeries with long surgical duration (the exposed group) had complexity scores greater than 10 (case group). Thus, odds ratios were not calculated for this variable.

Table 3.6: Least square mean (± standard error) for dependent variables with statistically significant differences between short and long surgeries

The Surgical Duration	Short Surgeries	Long Surgeries
Neck Risk Score	2.7 (±0.1)	2.9 (±0.1)
Change in Fatigue	1.3 (±0.4)	2.5 (±0.4)
Change in Lower Back Pain	1.2 (±0.5)	2.5 (±0.5)
Mental Demand	12.5 (±0.9)	16.9 (±0.9)
Physical Demand	9.3 (±0.9)	15.0 (±0.9)
Complexity	12.8 (±1.0)	17.8 (±1.0)
Difficulty	11.1 (±1.0)	16.2 (±0.9)

3.3.6 Correlation Between Subjective and Objective Measurements

Spearman's rank correlation coefficient (Rs; $-1 \leq Rs \leq +1$) was used to investigate the correlation between subjective and objective measurements. The change in fatigue and pain scores were chosen as subjective variables, while the postural angles and risk scores were the objective variables. Table 3.7 summarizes these findings.

The results revealed that the neck angle was correlated with the change in neck pain score (Rs=0.35), the change in upper back pain score (Rs=0.342), and the physical demand (Rs=0.3212). Similarly, the neck risk score was correlated with the change in neck pain score (Rs=0.341), the change in upper back pain score (Rs=0.3399), and the physical demand (Rs=0.317).

The trunk angle was found to be correlated with the change in neck pain score (Rs=0.3916), the change in lower back pain score (Rs=0.3392), the change in the right shoulder pain score (Rs=0.3013), and physical demand (Rs=0.38). The trunk risk score was correlated with the change in neck pain score (Rs=0.4269), the change in the lower back pain score (Rs=0.3409), and the physical demand (Rs=0.3572).

The right shoulder angle and right shoulder risk score were correlated with the change in fatigue (Rs=0.3368 and Rs=0.3319, respectively). The left shoulder angle was correlated with the change in the left wrist/hand pain score (Rs=0.2994). None of the other Spearman's rank correlation coefficients were statistically significant.

Table 3.7: Correlation (Spearman's rank correlation coefficient (Rs) and prob>|Rs|) between calculated risk scores/postural angles (angles) and change in fatigue and pain scores

	Neck		Trunk		Right Shoulder		Left Shoulder	
	Risk Score	**Angle**	**Risk Score**	**Angle**	**Risk Score**	**Angle**	**Risk Score**	**Angle**
Fatigue*	0.2635 0.0803	0.2650 0.0786	0.2432 0.1074	0.2392 0.1136	0.3319 **0.0278***	0.3368 **0.0254***	0.1719 0.2588	0.1733 0.2550
Neck*	0.3410 **0.0219***	0.3500 **0.0184***	0.4269 **0.0034***	0.3916 **0.0078***	0.1901 0.2165	0.1974 0.1990	0.1341 0.3800	0.1560 0.3063
Right shoulder*	0.1597 0.2948	0.1440 0.3454	0.2890 0.0542	0.3013 0.0443*	0.2164 0.1582	0.2346 0.1253	0.1848 0.2244	0.1849 0.2240
Left shoulder*	0.1427 0.3497	0.1201 0.4322	0.2551 0.0908	0.2789 0.0636	0.2664 0.0805	0.2973 0.0500	0.2078 0.1707	0.2103 0.1656
Upper back*	0.3399 **0.0223***	0.3420 **0.0215***	0.1836 0.2272	0.1478 0.3327	0.0665 0.6682	0.0732 0.6367	0.0606 0.6927	0.0696 0.6495
Lower back*	0.2032 0.1806	0.2304 0.1279	0.3409 **0.0219***	0.3392 **0.0226***	0.0649 0.6755	0.0468 0.7629	0.1000 0.5134	0.0859 0.5748
Right wrist/hand*	0.2084 0.1694	0.1751 0.2498	0.1052 0.4915	0.0717 0.6399	-0.0062 0.9681	0.0073 0.9624	0.1279 0.4026	0.1691 0.2667
Left wrist/hand*	0.2100 0.1663	0.1696 0.2653	0.0232 0.8796	0.0040 0.9794	0.1517 0.3255	0.1557 0.3128	0.2669 0.0763	0.2994 **0.0457***
Mental demand	0.0158 0.9181	0.0161 0.9164	-0.0235 0.8781	-0.0144 0.9250	-0.1709 0.2674	-0.1959 0.2025	- 0.1442 0.3446	-0.1201 0.4320
Physical demand	0.3170 **0.0338***	0.3212 **0.0314***	0.3572 **0.0160***	0.3800 **0.0100***	0.2483 0.1041	0.2506 0.1009	0.1567 0.3041	0.1755 0.2488
Complexity	-0.1304 0.3932	-0.1336 0.3817	-0.1606 0.2920	-0.1384 0.3646	-0.1954 0.2037	-0.2208 0.1498	- 0.1348 0.3774	-0.1251 0.4128
Distraction	-0.1765 0.2461	-0.1900 0.2114	-0.0482 0.7532	-0.0138 0.9281	-0.2197 0.1519	-0.2168 0.1574	- 0.1172 0.4433	-0.1256 0.4109
Difficulty	-0.0067 0.9654	-0.0077 0.9603	-0.0932 0.5472	-0.0759 0.6246	-0.0783 0.6176	-0.1129 0.4711	- 0.1255 0.4170	-0.0982 0.5260

3.4 Discussion

The aim of this study was to quantify the workload and discomfort in vascular surgeons and also identify any procedural factors that might influence surgical workload. The results from this study showed that the neck and trunk are the body segments with the most awkward postures and pain scores. The positive correlation between the ergonomically unfavorable postures and the pain scores experienced by the surgeons in the neck and trunk indicates a higher probability of developing long term musculoskeletal disorders in these two body segments. Furthermore, the discomfort associated with ergonomically risky posture significantly increased as a function of operative duration. Finally, mental workload was significantly high across all surgical procedures.

3.4.1 Basic Statistics and Upper Body Pain Scores

Figure 3.6 demonstrated that the neck is the body part with the highest risk of developing musculoskeletal disorders (mean risk score=2.8/4.0), followed by the trunk (2.1). This is consistent with Figure 3.5 results showing the neck as the most awkward posture (mean angle=37.1º), compared to all other segments examined in this study over the surgical procedures. Interestingly, body pain score results showed that the highest percentage change in pain scores of 3+ were reported in the neck (36%), low back, and upper back (30%). These results cumulatively highlight the strong association connecting awkward posture, musculoskeletal pain, and risks of developing musculoskeletal disorders. The prospective data in this study further support findings in the literature that neck and back pain are two of the most common sites affected by MSD among vascular surgeons (Davila et al., 2018). Further, neck and back pain have also been linked to anticipation of reduced surgeon performance (Davila et al., 2019), which puts patients at risk. This finding implies that ergonomic interventions in vascular surgery should be targeted at reducing awkward neck posture and, secondarily, trunk posture.

3.4.2 Difference between Open and Non-Open Surgeries

The higher neck angle, right shoulder angle and neck pain score in the open procedures compared to the non-open procedures (Table 3.3) can be attributed to procedural differences. For instance, open procedures require larger incisions and mostly require the surgeon to illuminate the surgical field with light from a surgical headlight or magnify the field using a pair of surgical loupes worn on the head. The illumination is achieved via a flexed neck posture, leading to the increased neck angle and pain scores reported in this study. Conversely, a greater portion of non-open vascular procedures involve the visualization of catheters progression through blood vessels displayed on mounted screens resulting in non-awkward neck postures and minimal upper extremity movement. This explanation is underscored by studies from (Van Det et al., al 2008) and (Nguyen et al., 2002) which showed significantly smaller neck flexion angles and more dynamic shoulder movement during laparoscopic minimally invasive procedures compared to open surgeries. This combination of awkward neck flexion angle, pain and relatively large right shoulder angle could potentially be responsible for the significantly higher physical demand associated with open vascular procedures than non-open observed in this study.

3.4.3 Difference Between Surgeries With and Without Loupes

The association among loupe use, surgeon posture, and MSDs is not well-understood. The results of our study revealed that, in procedures where loupes were used, not only the neck and trunk angles and risk scores increased significantly, but there was also a significant increase in the change in the neck pain score and the physical demand, with an odds ratio of 4.7. This may be one of the first studies that looked at the association between loupe use, surgeon posture, and body pain scores. It should be noted that these findings don't necessarily indicate any causal relationship between using loupes and the dependent variables such as pain scores because the surgical procedures with loupes and without loupes were not the same, the surgical duration also differed, and this was a cross-sectional study.

The evidence supporting the effects of surgical loupes on body segmental posture and pain are contradictory. For example, a few studies have shown improvement in posture as an effect of wearing magnification loupes compared to without loupes amongst dental hygienists (Branson et al., 2010; Branson et al., 2004; Maillet et al., 2008). Conversely, the findings of two recent studies are nuanced concerning the effects of wearing loupes on physical well-being, and more research in the future was suggested by the authors (Hayes et al., 2014; Hayes et al., 2016). In a study by Nimbarte and colleagues in 2013, adding loupes and headlamps to a biomechanical model of the microsurgeons' cervical spine led to significant increases in the cervical spine loading (Nimbarte et al., 2013). In a recent systematic review by Plessas and Delgado (2018) (Plessas and Bernardes Delgado, 2018), it was concluded that the use of dental loupes might have positive effects on relieving upper extremity pain, but their effect on neck pain was sparse. There are differences in the postures assumed by the clinicians with respect to patients between dentists and surgeons as well as the duration of their cases, just as there are differences in the postures assumed by different surgical specialties. Vascular surgery can be centered in the chest (many aortic procedures) as well as the extremities. The surgical posture for these different procedures as well as the type of surgery (open, hybrid, endo) make vascular surgery more varied than, for example, breast surgery surgeon postures (Hallbeck et al. 2020). However, the postures for most surgical procedures are extreme (Meltzer, et al 2020) and put the highly trained surgeon at risk for MSDs. Therefore, future studies are thus needed to isolate the impact of using loupes on musculoskeletal loading among surgeons across various specialties and by surgical approach.

3.4.4 Difference Between Surgeries With and Without RPPE

The mixed-effect models showed that the mean neck angle and risk scores were lower for surgical procedures in which RPPE was worn compared to surgical procedures without RPPE. This observation can be ascribed to the procedure types in which RPPE is worn. For the most part, RPPE is used in endovascular procedures, in which surgeons use minimally invasive catheter-based techniques to treat endovascular defects. Furthermore, X-ray fluoroscopy is used at the end of open vascular procedures to ensure complete restoration of blood flow. In these two surgical procedures, surgeons visualize the progress of catheters or blood blow restoration via movable screens mounted on the ceiling, on booms, or wheels. This gives them the flexibility to adjust the screens to eliminate or minimize awkward neck posture, which is reflected in the significantly lower neck flexion and pain score observed in the surgical procedures with RPPE compared to procedures without RPPE (predominantly open vascular procedures). While the extra weight of the RPPEs has been recently shown to increase trunk extensor muscle fatigue (Tetteh et al., 2020) and subsequently could lead to long-term low back disorders, it should be noted that this study solely examined postural risk, not the overall ergonomic risk of using RPPEs.

3.4.5 Effects of the Surgical Duration

In this study, the postural angles and risk scores were not affected by the surgical duration (except for neck risk score; p=0.0469). However, the subjective measurements including fatigue, lower back pain score, mental demand, physical demand, complexity, and difficulty increased significantly for long surgeries. This shows the importance of the subjective measurements combined with objective assessments. Our findings are consistent with survey results from Wells and colleagues in 2019 who reported a positive association between the duration of surgery and endoscopic surgeons' pain/discomfort (Wells et al., 2019). A possible intervention may be taking micro-breaks during the surgical procedure. The findings of a study by Dorion and Darveau (Dorion and Darveau, 2013) and Engelmann, et al. (Engelmann et al., 2011) showed some positive impacts of micro-pauses during surgical procedures on surgeons' fatigue, strength, and accuracy. Additionally, recent studies found that intraoperative micro-breaks with exercises also reduced perceived fatigue and discomfort while causing little distraction from the workflow (Abdelall et al., 2018; Coleman Wood et al., 2018; Hallbeck et al., 2017; Park et al., 2017).

3.4.6 Correlation between Subjective and Objective Measurements

Correlations between postural angles and corresponding pain scores showed the consistency between the IMU data (direct measurement) with the surveys (self-reported data) for neck and trunk. The trunk angle impacts the neck angle. This could be the reason for the correlation between trunk angle and neck pain score. The comparison of IMU data with surveys for the right and left shoulder was difficult to interpret. The upper back and upper arm sensors were securely attached to the surgeon's scrubs, however, there could be artifact motions in these sensors especially in the upper arm sensors due to the nature of the upper arm and hand movements during a surgical procedure. These unavoidable minimal motions could have added variability to the data of these sensors.

Although statistically significant correlations were found between the objective assessments (from the IMU data) and the self-reported surveys, they were not very highly correlated. On the other hand, the risk scores used in this study didn't have consistently better correlations with the survey scores compared to the correlations between postural angles and survey scores. The results from the correlations indicate that these risk scores are relevant; however, there was no proof to show that they worked better than the postural angles. In future studies, these findings may be used as a foundation for creating better and more relevant risk scores. Other factors, such as the effect of the total duration of a surgical procedure, the duration of static postures, and the work-rest cycle may provide better risk scores. Such risk scores may also have higher correlations with self-reported surveys.

3.4.7 Limitations

The data collection inside the OR has its own requirements and challenges (Hallbeck and Paquet, 2019). Although all the data collections were for vascular procedures, it should be considered that these surgical procedures were not the same vascular surgeries, specifically. Even for very similar surgical procedures, it should be noted that each surgery has its own specific conditions, and patient factors and data collections were not seemingly for identical tasks. This adds variability in the data as a limitation of the study. It should be mentioned that, for some potential independent variables, we didn't have enough surgical procedures to evaluate it as a new factor. Also, for some analyzed independent variables, the data were not distributed equally among different levels of that variable; 74% of the surgeries were open procedures while only 26% were non-open surgeries, either endovascular or hybrid procedures. Despite these limitations, this is one of the first studies to investigate different aspects of vascular surgery workload using both subjective and objective assessment tools. This could pave the way for future studies in this area. It can also provide researchers with required details to find the possible underlying risk factors for fatigue, discomfort, and MSDs among surgeons and to develop ergonomic interventions especially for vascular surgeries.

3.5 Conclusions

The results from this study show that the ergonomic risk of performing vascular surgery is high for vascular surgeons' neck and low back. Higher risk was associated with the surgical duration, open procedures, and loupe magnification. Additionally, our results show that vascular surgery procedures are cognitively demanding as well as physically demanding. Thus, future research should explore various ergonomic interventions to mitigate these risk factors. Interventions should be targeted at the surgeon's neck and back for reducing the physical demand and pain.

Author Contributions

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Statistical Analysis: HN, ET, MMM

Interpretation: HN, ET, MMM, EF, AJM, BCM

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CHAPTER 4. AN ALGORITHM FOR PREDICTING THE UTILITY OF EXOSKELETONS FOR DIFFERENT SURGERIES

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Abstract

As exoskeletons are currently being research as a potential intervention to reduce the prevalence of musculoskeletal disorders amongst surgeons, there is the need to develop standardized techniques to deploy the exoskeletons in surgical procedures that will benefit from the intervention. For this reason, this study was conducted to use segmental kinematics data from a number of surgical subspecialties, coupled with expert opinion on exoskeleton appropriateness to develop. Linear and quadratic discriminant analyses were used to identify segmental kinematics variables that classified 30 surgeries into two classes: appropriate for exoskeleton intervention and not appropriate for exoskeleton intervention. The results from this study showed that the cumulative fatigue due to posture in the body segments, alongside several other kinematic variables predicted exoskeleton with minimal misclassification error. This study is the first of its kind to develop an exoskeleton intervention deployment technique for the operating room.

4.1 Introduction

Many surgeons from various surgical subspecialties suffer from work-related musculoskeletal pain and disorders, which affects their career longevity and presents a labor threat to the healthcare system. For example, a survey study by Szeto et al. (2009) of general surgeons reported that 82.9%, 68.1%, and 57.8% of the surveyed surgeons suffered from neck, low back, and shoulder-related problems, respectively. Moreover, a nationwide survey of nearly 800 surgeons by the society of vascular surgeons showed that neck and shoulder pain was experienced by 45% and 35% of the survey respondents, respectively (Wohlauer et al., 2019). Furthermore, Wohlauer et al. (2019) reported that the surgeons experienced a 4.4 ±2.3 out of 10 pain on the Borg CR10 scale after each day of operating. Interestingly, another survey results showed that a significant number (31.4%) of vascular surgeons had to seek medical help for their pain even though only 4.4% of the survey respondents reported their symptoms to employee health resources (Davila et al., 2019). Surgeon musculoskeletal pain has been linked to burnout and inefficiency, and some surgeons feel that their careers might end prematurely due to work-related musculoskeletal pain and disorders (Davila et al., 2019). Thus, as the U.S. forecasts a potential shortage of surgeons (Way, 2010), ergonomic interventions must be developed to prolong the longevity of surgeon careers.

While ergonomic interventions such as intra-operative breaks, operating room reconfiguration, and armrests have been proposed and tested in the operating room with some success and drawbacks (Blocker et al., 2013; Kromberg et al., 2020; Matern et al., 2005; Rogers et al., 2012), exoskeletons are a fledgling intervention in this domain that holds immense potential in alleviating surgeon musculoskeletal pain and disorders. Their ability to reduce musculoskeletal pain in more traditional industrial environments are extensively documented in the ergonomics literature. For instance, Kim et al. (2018) showed that upper extremity exoskeletons reduced shoulder muscle EMG by approximately 33.5% in an overhead drilling task and almost 100% in the anterior deltoid during the lifting phase of a lift-walk-stack activity (Theurel et al., 2018). Furthermore, in an onsite farm machinery assembly plant, wearing an upper extremity exoskeleton reduced deltoid muscle activity in consecutive job cycles (C.I.=14-25%MVC vs 11-17%MVC), resulting in significantly reduced fatigue risk (Gillette & Stephenson, 2019). Moreover, low back exoskeletons have also significantly reduced low back muscle activity (31.5% and 29.3% for iliocostalis and longissimus erector spinae, respectively) in repetitive lifting tasks (Alemi et al., 2019). Such reduction in muscle activity translates to improved muscle discomfort results, a significant reduction in task completion time (Kim et al., 2018), and ultimately a reduction in surgeon musculoskeletal pain and disorders.

Exoskeletons are not currently being implemented in the surgical environment due to several barriers identified by (Cha et al., 2019). Their study found four broad themes (individual characteristics, perceived benefits, intervention characteristics, and environmental/societal factors) that need to be addressed for exoskeletons to be widely adopted in healthcare delivery. Their findings under the theme of perceived benefit questioned the exoskeleton's ability reduce the shoulder muscle and overall body fatigue experienced by surgeons. A recent study using the levitate airframe upper extremity exoskeleton in a series of laparoscopic surgeries (Liu et al., 2018) showed that the exoskeleton reduced subjective shoulder muscle fatigue by 90% (3.11 vs. 5.88). Furthermore, significant post-operative musculoskeletal pain reduction was reported by 85% of study participants.

While the results from (Cha et al., 2019) and (Liu et al., 2018) highlight the barriers and potential benefits of using exoskeleton interventions, these two studies were limited in their scope of coverage. For instance, both studies were restricted to laparoscopic surgery, which requires significant static neck postures in extension (Nguyen et al., 2002; Szeto et al., 2012) and awkward upper extremity movement (Nguyen et al., 2002). However, other surgical subspecialties with different segmental kinematics and postural requirements predispose surgeons to musculoskeletal pain and discomfort. For instance, using Inertial Measurement Units (IMUs) to quantify the postural demand in vascular surgeons, Davila et al. (2018) reported that the surgeons assumed high risk (between 20° and 60° degree neck flexion) and severe neck posture (above 60° neck flexion) for 67% and 10.8% of operating time respectively. Furthermore, a study on the postural requirement for final-year dental surgery students revealed that approximately 95% of the students assumed poor neck and back positioning (flexion) while performing surgery (Ng, Hayes, & Polster, 2016). The results from these two studies highlight the need to extend exoskeleton intervention research beyond laparoscopic surgery.

The direct relation between postural kinematics and muscle fatigue development has been previously used to drive ergonomic interventions and, thus, may be used to inform exoskeleton usage in the operating room potentially. For instance, using a biofeedback postural correction technique, Bazazan et al. (2019) found that the method improved RULA postural scores in the neck, upper back, and shoulder. This improvement in posture resulted in a significant reduction in physical fatigue (12.19 vs. 9.99) and the percentage of musculoskeletal symptoms reported (89% vs. 82.4%) in an Iranian petrochemical processing plant. For exoskeleton interventions to be effective in the operating room, they must be targeted at procedures that require awkward segmental postures and significant segmental fatigue/physical discomfort development. Furthermore, the segmental planes of motion should match that of the exoskeleton. Hence, the aim of this study was to use segmental kinematics data to determine the potential utility of exoskeleton interventions in the surgical environment. It was hypothesized that a combination of time and frequency domain postural variables can be used to create a predictive model capable of identifying which surgeries will be benefit from exoskeleton intervention.

4.2 Methods

A combination of time and frequency domain postural analysis was used to generate a set of independent variables that were used as possible predictors for exoskeleton utility as determined by a healthcare ergonomics expert. Two classification models: Linear and Quadratic Discriminant Analysis (LDA/QDA) were used to create a combination of predictor variables (discriminant function) that best classified the surgical procedures into two groups (0 for no exoskeleton intervention recommended and 1: for exoskeleton intervention recommended) for each body segment. Hence a total of six discriminant functions were created in this study.

4.2.1 Segmental Postural Data

The postural data used for this study was obtained from surgeons performing four different types of surgeries at the Mayo Clinic. Postural data from 30 surgeries were used for this study. These data were collected by attaching Inertial Measurement Units (IMUs) to the head, upper trunk, and shoulder of the participating surgeons. Before starting data collection, baseline segment angles (shoulder, head-neck, trunk, and shoulders) were recorded for each surgeon while in a neutral posture. This served as a reference to all respective segment angles, and deviations from these postures are the angles used for this study. The IMUs were programmed to record postural data at a frequency of 128Hz for the entire duration of each surgery.

4.2.2 Time Domain Independent Variables

The time-domain variables consist of selected postural percentiles for each body segment. These percentiles are the 10^{th}, 25^{th}, $75^{th,}$ and 90^{th} percentiles for neck flexion, shoulder deviation, and trunk flexion for each surgery observed. The mean segment angle and operating time were also included. Finally, the last time domain variable representative of the cumulative awkward posture fatigue was developed from the postural kinematics data. A detailed description of this variable can be found in the next section.

4.2.2.1 Cumulative postural fatigue risk model (CPFRM)

The CPFRM was based on a modified version of the Rapid Upper Limb Assessment (RULA) (McAtamney & Nigel Corlett, 1993). The risk assessment tool was altered in three different ways. First, CPFRM focused only on posture, as opposed to RULA, which generated risk scores based on posture, number of movements, and force exerted. Secondly, instead of a grand risk score found by RULA, CPFRM estimates each body segment's score (CPFRS). This is relevant because different surgeries pose different postural demands for different body segments, and it is appropriate to target body segments with high physical demands. Finally, fatigue recovery was built into the model to account for any recovery from fatigue when each body segment is in a neutral posture during task performance.

Like RULA, postural fatigue risk coefficients were estimated from biomechanical models of the three-body segments based on the moment required to maintain different postural angles. The details of the models are described in Section 4.2.2.1.2. and a demonstration of the CPFRS calculation of is outlined in Section 4.2.2.1.4 and Table 4.2.

4.2.2.1.2 Biomechanical modeling

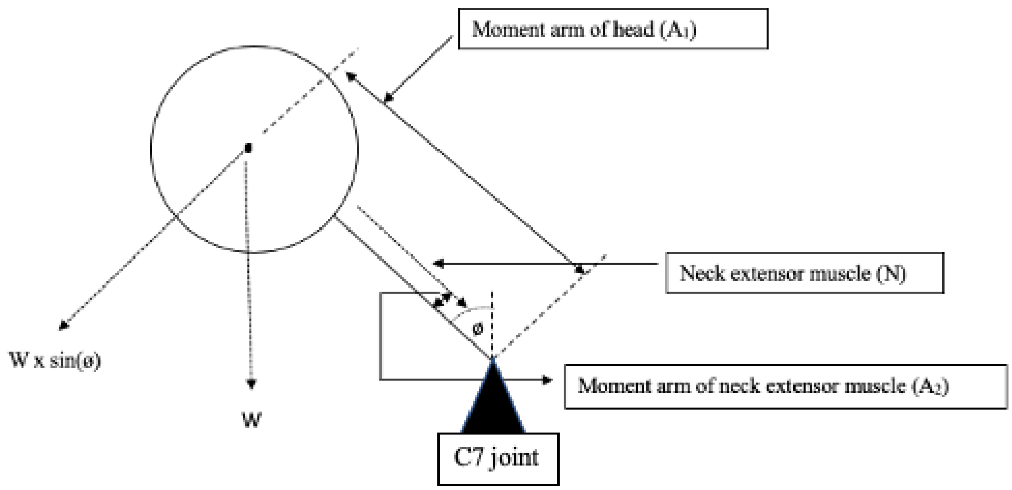

Figure 4.1: Static biomechanical model of the head-neck segment

$W = mass\ of\ 4.54kg$ (Yoganandan, Pintar, Zhang, & Baisden, 2009)

$A_2 = moment\ arm\ of\ neck\ extensor = 14.375mm$ (Ackland, Merritt, & Pandy, 2011)

$A_1 = distance\ from\ C6 - C7\ to\ center\ of\ mass\ of\ the\ head = 12.00cm;$ (Ahmed, Qamar, Imram, & Fahim, 2020)

$$\sum M = 0; NxA_2 = W \times 9.81 \times sin(\emptyset)xA_1$$

$$\sum M = W \times 9.81 \times sin(\emptyset)xA_1$$

$$\sum M = 4.5kg \times 9.81 \times sin(\emptyset)x120.0mm$$

$$\sum M = 5297.40Nmm \times sin(\emptyset) - - - - - (3)$$

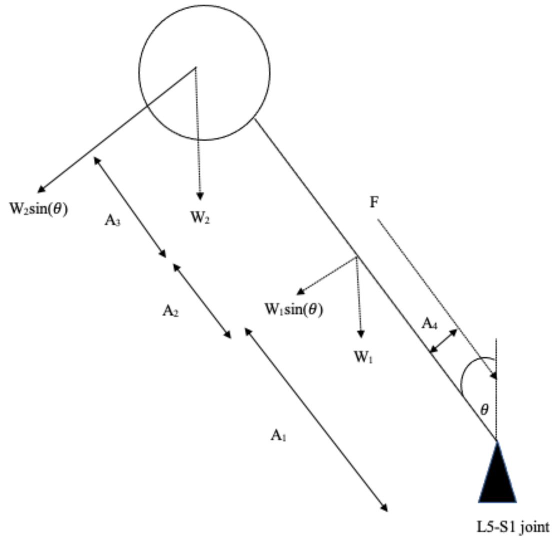

Figure 4.2: Static biomechanical model of the trunk

$W_2 = Mass\ of\ head = 4.5kg$ (Yoganandan et al., 2009)

$W_1 = Mass\ of\ torso = 35kg$

$A_4 = Moment\ arm\ of\ trunk\ extensor\ muscle = 68mm$; (Németh & Ohlsén, 1986)

$A_1 = distance\ from\ L5 - S1\ joint\ to\ center\ of\ mass\ of\ torso = 25.0cm$; (NASA, 2020)

$A_3 = distance\ from\ C6 - C7\ joint\ to\ center\ of\ mass\ of\ head = 12cm$; (Ahmed et al., 2020)

$A_2 = distance\ from\ center\ of\ mass\ of\ torso\ to\ C6 - C7\ joint = 26.8cm$; (NASA, 2020)

$$\sum M = 0; = [W_1 \times 9.81 \times sin(\emptyset) x A_1] + [W_2 \times 9.81 \times sin(\emptyset) \times (A_1 + A_2 + A_3)]$$

$$\sum M = [35 \times 9.81 \times sin(\emptyset) \times 250mm] + [4.5 \times 9.81 \times sin(\emptyset) \times (638mm)]$$

$$\sum M = [85837.5Nmm \times sin(\emptyset)] + [28164.51Nmm \times sin(\emptyset)]$$

$$\sum M = [114002.01Nmm \times sin(\emptyset)] - - - - - (4)$$

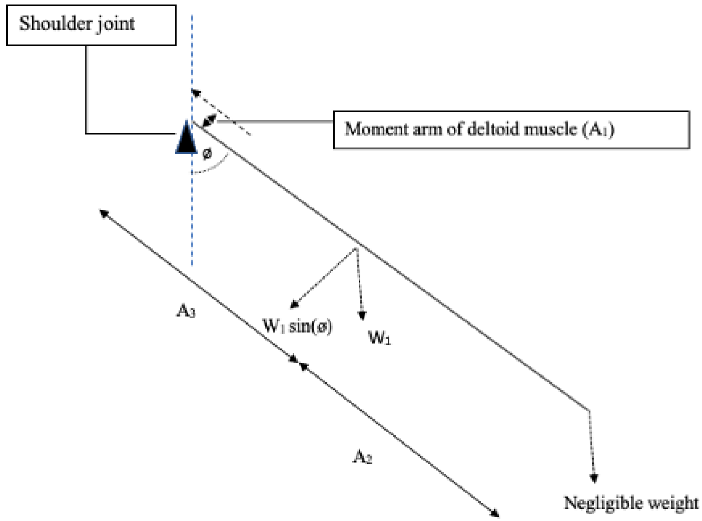

Figure 4.3: Static biomechanical model of the upper extremity (shoulder)

$W_1 = Weight\ of\ the\ arm = 6kg$; (Plagenhoef, Evans, & Abdelnour, 1983)

$A_2 + A_3 = length\ of\ arm = 81.6cm$; (NASA, 2020)

$A_3 = length\ of\ center\ of\ mass\ of\ arm\ from\ shoulder\ joint = 40cm;$ Assumed to be approximately 50% of arm length

$$\sum M = 0; = W_1 \times 9.81 \times sin(\emptyset) \times A_3 + WeightNeg * sin(\emptyset) \times (46cm + A_2)$$

$$\sum M = [6 \times 9.81 \times sin(\emptyset) \times 400mm] + [0 * sin(\emptyset) \times (380mm + 460mm)]$$

$$\sum M = 23544Nmm \times sin(\emptyset) - - - - - (5)$$

The biomechanical models of each of the body segments are shown in Figures (4.1-4.3), and the moment estimation equation as a function of segment angle for the neck extensor muscle, back extensor muscles (erector spinae), and the deltoid muscles are shown in Equations (3-5). Anthropometric data used in the biomechanical models were obtained from the literature.

4.2.2.1.3 Risk coefficients

The moments for each segment across each postural angle were normalized to their respective maximums. The normalized moment values were grouped into bins of 10% sizes for the head, trunk, and neck from 0 to 100 to obtain ten risk categories. Furthermore, the coefficients of the moment estimation equations for the three body segments (neck, shoulder, and trunk) were normalized to the coefficient of the trunk's moment estimation equation to obtain a moment ratio of (head-neck [0.046]: trunk [1.000]: shoulder [0.206]).

Postural angles less than or equal to 10° degrees were considered rest for trunk and shoulder (Tichauer, 1966), while any neck flexion posture below 15° was also classified as rest (Chaffin, 1973). All rest postures were assigned a negated coefficient of risk to signify a recovery from fatigue, as shown in Table 4.1. This ratio was applied to the risk 10% bins of risk categories generated from the normalized moment requirement, and the results are displayed in Table 4.1.

Table 4.1: Table of postural fatigue risk coefficients and angular thresholds

Risk Level	**Head-Neck threshold in degrees**	**Risk coefficient**	**Trunk threshold in degrees**	**Risk coefficients**	**Shoulder threshold in degrees**	**Risk coefficients**
Rest	15	-0.023	10	-0.5	10	-0.103
Risk 1	17	0.023	17	0.5	17	0.103
Risk 2	23	0.046	23	1	23	0.206
Risk 3	30	0.069	30	1.5	30	0.309
Risk 4	36	0.092	36	2	36	0.412
Risk 5	44	0.115	44	2.5	44	0.515
Risk 6	53	0.138	53	3.0	53	0.618
Risk 7	64	0.161	64	3.5	64	0.721
Risk 8	90	0.184	90	4.0	90	0.824

4.2.2.1.4 CPFRS estimation

The Cumulative Postural Fatigue Model accumulates the fatigue risk scores over the entire duration of performing surgery, including rest cycles that might be interspersed in between task performance using the formula below.

$$\text{CPFRS} = \left(\frac{\sum_{b=1}^{n}(\theta_b \times k_b \times t_b)}{T \times 90 \times max(k_b)}\right) ------ (6)$$

Where $T = duration\ of\ task\ performance$

$k = fatigue\ risk\ coefficient\ (see\ Table\ 8)$

$b = bin\ number$

$n = total\ number\ of\ risk\ category\ bins$

$\theta_b = average\ postural\ angle\ within\ bin$

$t_b = time\ elapsed\ in\ each\ bin$

This algorithm groups each body segment's postural angles into successive risk categories based on the segmental thresholds outlined in Table 4.1. For instance, a head-neck postural data set of {18,19,20,16,17,10,11,13,25,25,26,28,27,18,19} with each posture assumed for one (1) second will be grouped according to Table 4.1. This yields a total of five risk bins, making "n" in Equation 6 to be equal to five. The numerator (Equation 6) works like the trapezoidal rule to estimate the area under the time series of segmental angle. This area is normalized (divided) to area when a body segment assumes the most deviation (90°). Hence the CPFRS is a ratio between zero and one, with scores close to one indicative of significant postural fatigue and vice versa.

Table 4.2: Sample CPFRS calculation

Bin number	**Angles**	**mean bin angle**	**Fatigue risk category**	**Fatigue risk score**	**Time (s)**	$\boldsymbol{\theta_b k_b t_b}$
1	18, 19, 20	19	2	0.046	3	2.622
2	16, 17	16.5	1	0.023	2	0.759
3	10, 11	10.5	Rest	-0.023	2	-0.483
4	25, 25, 26, 28, 27	26.2	3	0.069	5	9.143
5	18, 19	18.5	2	0.023	2	0.851
$T = \sum_{i=1}^{5} Time$					14	
$\sum_{b=1}^{n} (\theta_b k_b t_b)$						12.892

$$\text{CPFRS} = \left(\frac{12.892}{14 \times 90 \times 0.184}\right)$$

$$\text{CPFRS} = 0.056$$

4.2.3 Frequency Domain Variables

Spectral analysis of the postural data was conducted to identify the intensity of static posture (a risk factor for muscular fatigue) within the time domain data. Specifically, the Welch’s power spectrum density estimate (pwelch) function in MATLAB was used to compute each frequency component's power spectral density (motion frequency) of the time domain motion data for each body segment. The independent variables extracted from the power spectra are the mean frequency (MF), median frequency (MDF), and cycle time (CT i.e., inverse of the mean frequency) for the kinematic data for each body segment per surgery. A summary of the predictor and response variables is shown in Table 4.3.

Table 4.3: Independent (predictor) variables and dependent (group) variable

Predictor variables for each Segment	**Dependent variable (exoskeleton utility recommendation)**
CPFRS	Level or group 1: Exoskeleton recommended (intervention has significant impact) Level or group 2: No exoskeleton recommended (intervention has low or insignificant impact)
10th percentile angle	
25th percentile angle	
Mean angle	
75th percentile angle	
90th percentile angle	
Operating time	
Mean frequency	
Median frequency	
Cycle time	

4.2.4 Dependent Variable

The dependent variable for this study was segmental exoskeleton utility for all the surgeries observed. This variable has two levels: Level 0 for “no exoskeleton recommended”, and Level 1 for “exoskeleton intervention recommended”. These ratings were obtained from a survey provided by a human factors/ergonomics engineer with expertise in operating room ergonomics and musculoskeletal disorders in surgeons. For each surgical procedure, the expert provided a “yes” or “no” response to the question of exoskeleton suitability as an ergonomic intervention to relieve musculoskeletal discomfort and pain. Thus, a total of 90 “yes/no” responses were obtained for from the survey: one for each body segment for all the 30 surgical procedures.

4.2.5 Statistical/Discriminant Analysis

All statistical analyses were completed in R version 1.1.456. To estimate a discriminant function that classifies the 30 surgeries in this study into these two groups, linear and quadratic discriminant analyses (see Equations 7 and 8) were performed for each of the three body segments. The underlying model assumptions: multivariate normality and homogeneity of covariance were checked before analyses. Both the forward and backward elimination technique was used to identify which combination of variables yielded the best discriminant function for each body segment using the stepclass function. Finally, each body segment's model's performance was evaluated using the Leave-One-Out Cross-Validation (LOOCV) technique. Specifically, the kinematic data for each body segments for 29 out of the 30 surgeries was used to train the classification model and the trained model was used to predict the class of the surgery excluded from the initial model. This was repeated 29 times and mean misclassification rate (MsR) for each was estimated using Equation 9.

$$\delta_i(X) = -\frac{1}{2}log_e|(\varphi_i)| - \frac{1}{2}(X-\mu)^T\varphi_i^{-1}(X-\mu) + log_e(\pi_i) ------ (7)$$

$$\delta_i(X) = -\frac{1}{2}\mu_i{}^T\varphi^{-1}\mu_i \ + \mu_i{}^T\varphi^{-1}X + log_e(\pi_i) \ ------ (8)$$

Where, $i = Number\ of\ classes, in\ this\ case\ 2{:}\ 0\ and\ 1$

Where, $\delta_i = Discriminant\ score$

Where, $X = Matrix\ of\ independent\ variables$

Where, $\mu = Vector\ containing\ means\ of\ each\ variable$

Where, $\varphi_i = covariance\ matrix\ of\ kinematic\ variables\ for\ class\ i$

Where, $\pi_i = prior\ probability\ for\ observation\ belonging\ to\ class\ i$

Where, $T = threashold\ or\ class\ boundary$

MsR: Mean misclassification rate

$$\text{MsR} = \frac{\sum_i^{29} \frac{nP_i - nCP_i}{nP_i}}{29} ------ (9)$$

Where, $\text{MsR} = mean\ misclassification\ rate$

$nP_i = Number\ of\ predictions$

$nCP_i = Number\ of\ correct\ predictions$

$nCEP_i = Number\ of\ correct\ exoskeleton\ predictions$

$nFNEP_i = Number\ of\ false\ "no - exoskeleton"\ predictions$

$nFEP_i = Number\ of\ false\ "exoskeleton"\ predictions$

4.3 Results

The model assumptions test revealed that the predictor variables from each of the classes did not violate the multivariate normality assumption ($p > 0.05$); however, the homogeneity of covariance was violated. While this implied that the QDA was the appropriate classification technique, Huberty (1975) showed that discriminant analysis was relatively robust to model violation. Hence both methods were used, and their performance were compared post-model building.

Table 4.4: Significant predictor variables and model performance (MsR) for each body segment

Body segment	QDA		LDA
	Significant predictor variables	Model performance MsR	Model performance MsR
Head-neck	NeckCPFRM, Neck10th	13.3%	13.3%
Trunk	TrunkCPFRM	23.3%	20%
Upper extremity	RArmCPFRM, RArmMean, RArmMeanF	26.67%	23.3%

A summary of the kinematic variables predictive of exoskeleton usage and prediction performance from the discriminant analyses are shown in Table 4.4. For both LDA and QDA, the cumulative fatigue risk score was predictive of the exoskeleton usage for all body segments. Furthermore, the 10th percentile segment angle was predictive of neck exoskeleton intervention. The decision to intervene with an exoskeleton or not for the trunk was solely determined by the CPFR score, while mean upper arm angle and frequency were predictive of upper extremity exoskeleton usage. It is interesting to note that operating time was not predictive of exoskeleton intervention for either neck, shoulders, or trunk. Finally, Table 4.4 also indicates that the same set of variables were predictive of exoskeleton intervention in both classification techniques. The misclassification rate shows that the head-neck model seemed to perform best, followed by the trunk, and finally the upper extremity. The misclassification rates are reflective of the level of distinction created between the discriminant scores (predicted classes) of the "exoskeleton recommended" and "no exoskeleton recommended" groups for the three body segments (head-neck, trunk, and upper extremities) in Figures 4.4-4.6. Finally, the model performance for the LDA and QDA were very similar even though the LDA seemed to have performed better in the trunk and extremity classification.

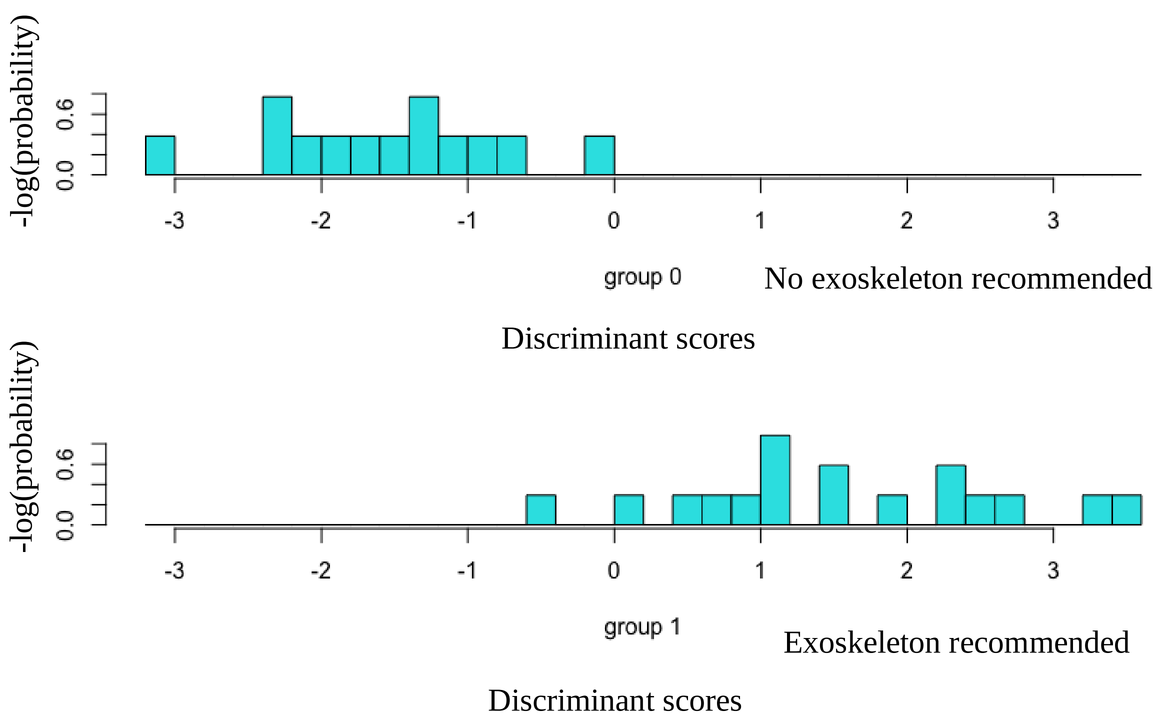

Figure 4.4: LDA histogram classification plot for head-neck exoskeleton recommendation

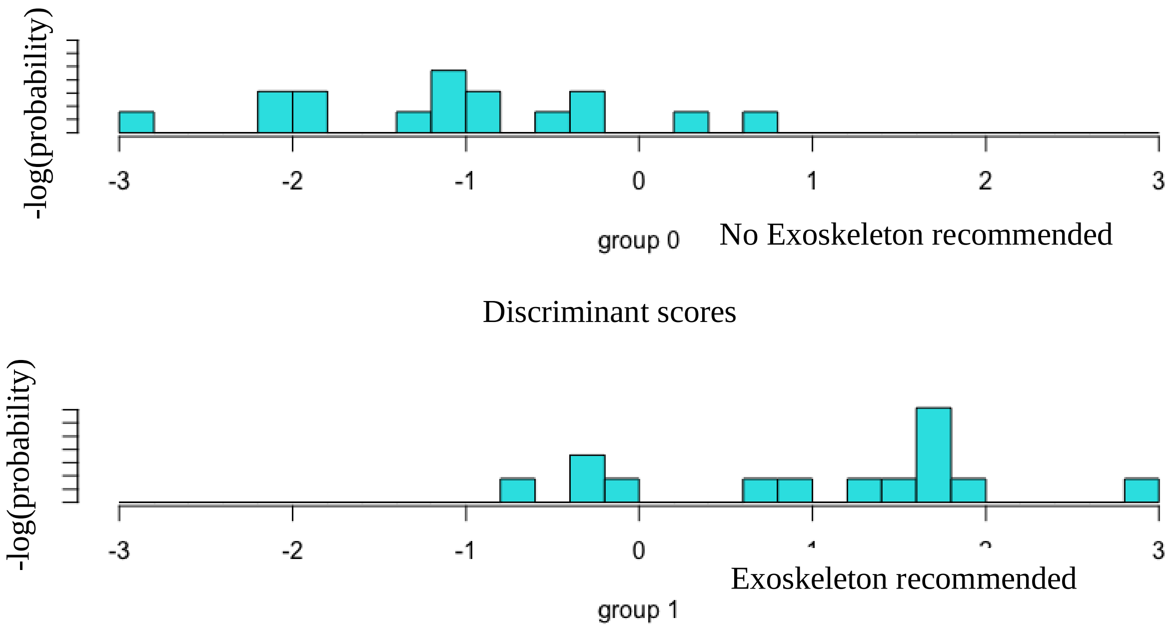

Figure 4.5: LDA histogram classification plot for trunk exoskeleton recommendation

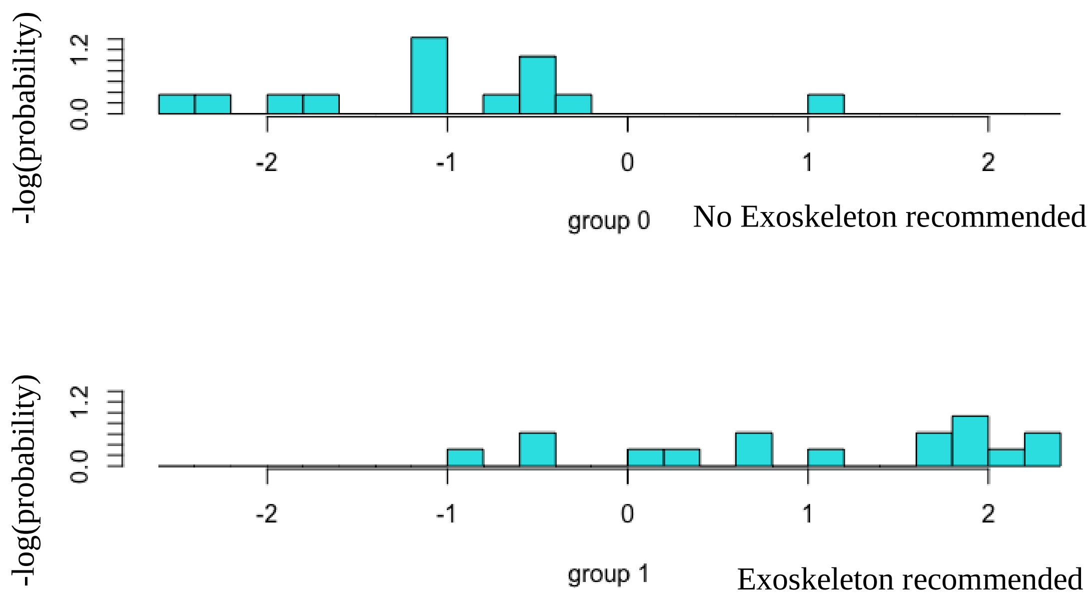

Figure 4.6: LDA histogram classification plot for shoulders exoskeleton utility

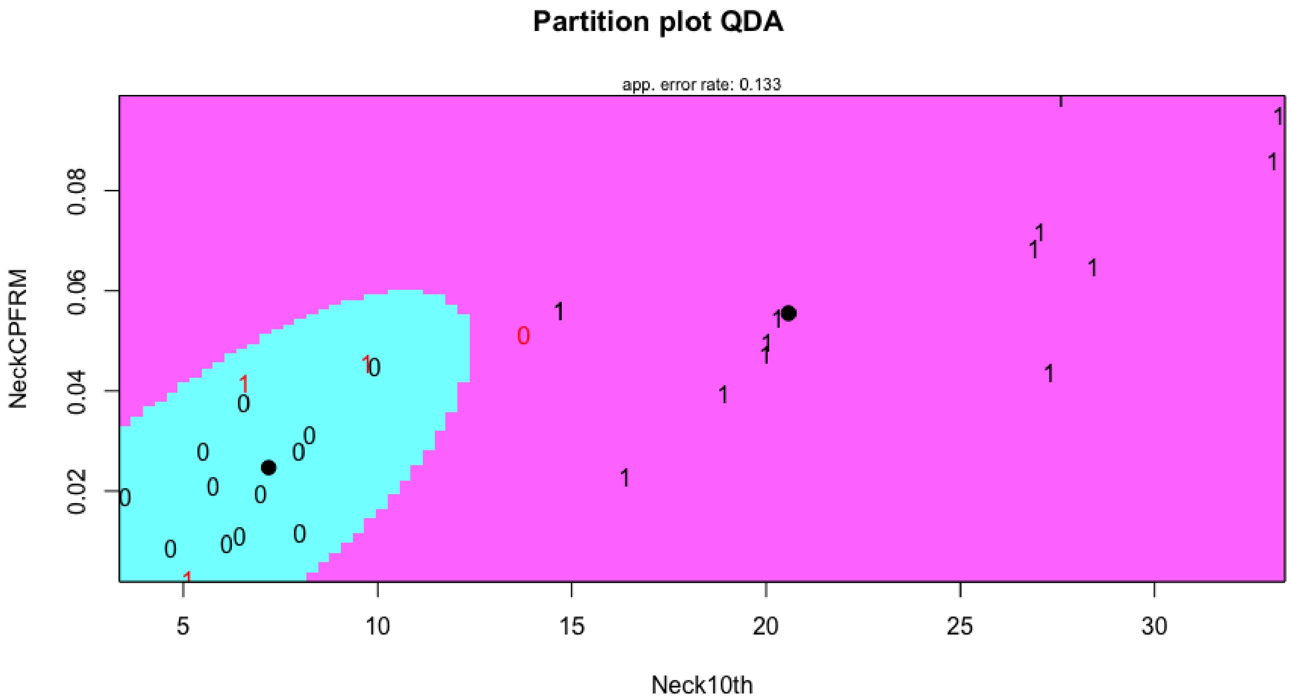

Figure 4.7: QDA Partition plot of exoskeleton classification in the head-neck showing the quadratic decision boundary

4.4 Discussion

The aim of this study was to identify segmental kinematic variables that can be used to inform the use of exoskeleton interventions in the operating room. This study represents the first of its kind to use segmental kinematics to inform the use of exoskeleton intervention in the operating room, complementing the findings of Cha et al., (2019), who identified barriers to be overcome for successful exoskeleton interventions in the operating room. The results from this study showed that the CPFRM score (a measure of cumulative fatigue from awkward posture) was predictive of exoskeleton intervention for all body segments. Furthermore, tenth percentile segment head-neck posture was predictive of the exoskeleton utility for the head-neck. In contrast, mean frequency, indicative of static posture was predictive of exoskeleton intervention for the shoulders.

A cursory look at Figure 4.7 reveals two interesting and consistent patterns concerning the predictive ability of the newly developed CPFR score and the 10^{th} percentile neck flexion angle. The surgical procedures classified as appropriate for exoskeleton intervention tend to have higher CPFR scores and 10^{th} percentile neck angle. In other words, the higher the cumulative segmental fatigue and 10^{th} percentile neck angle, the more appropriate it is for an exoskeleton intervention. It is worthy also to note that the 10^{th} percentile neck angles for procedures classified as beneficial for exoskeleton intervention started from approximately 15° with a majority of such angles over 20°. These two neck angle thresholds have been reported in the literature to correlate with the development of neck pain in physicians and other industrial workers as a result of fatigue development in the neck extensor muscles (Andersen et al., 2003; Kilbom, Persson, & Jonsson, 1986; Mehrdad, Dennerlein, & Morshedizadeh, 2012). Collectively, the CPFR score and trends in the 10^{th} percentile posture of the head-neck are consistent with the main purpose of introducing exoskeletons into the surgical space, which is to minimize or eliminate segmental fatigue and discomfort by providing support for body segments in long duration awkward posture (de Looze, Bosch, Krause, Stadler, & O'Sullivan, 2016).

The classification model for the upper arm exoskeleton intervention identified CPFR score, mean angle, and mean frequency (RArmMeanF) as a variable predictive of exoskeleton intervention. Of particular interest is the RArmMeanF, a factor related to static posture which is predominant in laparoscopic minimally invasive surgery (Szeto et al., 2012). Awkward static postures accelerate the rate of muscle fatigue development, which concurrently degrades cognitive attention resources (Stephenson, Ostrander, Norasi, & Dorneich, 2019). Hence, the support provided by an upper arm exoskeleton will help reduce the associated fatigue which may lead to improvement in cognitive attention resources while performing surgery.

Interestingly, operating duration was not predictive of exoskeleton intervention for any of the body segments. This partially underscores the findings from (Norasi et al., 2021) which reported no difference between the postural risk scores and segmental (trunk and shoulder) angles except neck based on operating duration in vascular surgery. Furthermore, this finding contradicts Wells, et al, (2019) that reported that 85% of the surgeons surveyed experienced pain by the time six hours of surgery was completed, a phenomenon which may warrant the need for exoskeleton intervention to relieve the discomfort. The contradiction can be attributed to the way the results were stated in Wells et al., (2019) without accounting for other factors responsible for the discomfort experienced while operating as well as recall bias since that was an emailed survey. Musculoskeletal disorders have been well documented amongst endoscopic minimally invasive surgeons (Van Det, et al, 2009) and this has been ascribed to the non-ergonomic postures, repetitive movements, combined with extraneously long procedures (Park, et al, 2010). Hence, our results shows that just because a procedure requires long operating time does not necessary mean that the physical demand is high to warrant an exoskeleton intervention.

The result from this study has important implications for exoskeleton design improvement. There are surgical procedures that put more than one body segment in significant awkward posture leading to segment fatigue and discomfort. For instance, numerous vascular surgery procedures require awkward neck and trunk posture (Norasi et al., 2021). Current exoskeleton designs are only targeted at specific body segments ie (upper extremity and trunk) and there is no commercially available exoskeleton designed specifically for the neck. Hence, to address the fatigue development in multiple body segments simultaneously, exoskeletons need to be worn in tandem, a practice, which is not possible with current exoskeleton designs. For this reason, exoskeleton manufacturers should consider expanding their current designs to address multiple segment fatigue development.

There are certain limitations to be considered for the broader application of the results from this study. First, the sample size of data used in this study was segmental kinematics data for 30 surgical procedures (observations) obtained from surgical subspecialties such as vascular, breast and laparoscopic surgery. The limited sample size is a potential reason for the higher misclassification rate in the trunk and upper extremity exoskeleton models. Furthermore, there are numerous surgical subspecialties whose segmental kinematics data were not used in this work due to lack of availability. Hence, it is anticipated increasing the sample size to include other surgical subspecialties will improve the robustness of the predictive model.

4.5 Conclusions

Creating a standardized technique to deploy exoskeletons is the first step in introducing this intervention into the operating room. The result from this study shows that segmental kinematics data acquired from surgeons performing a variety of surgeries can be used to develop such a strategy. However, this technique needs to be improved by expanding it to include kinematics data from surgeries that was not included in this model for efficient performance as well as partitioning the types of surgeries examined and support mechanisms available with passive exoskeletons.

Author Contributions

Conception and design: ET, GM, MSH

Analyses and interpretation: ET, GM, MSH

Data collection: ET

Writing the article: ET

Critical revision of the article: GM, MSH

Final approval of the article: ET, GM, MSH

Overall responsibility: ET

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Appendix: IRB Approval Memo

IOWA STATE UNIVERSITY
OF SCIENCE AND TECHNOLOGY

Institutional Review Board
Office for Responsible Research
Vice President for Research
2420 Lincoln Way, Suite 202
Ames, Iowa 50014
515 294-4566

Date: 07/13/2020

To: Gary Mirka

From: Office for Responsible Research

Title: **Expert opinion on exoskeleton usage**

IRB ID: **20-285**

Submission Type: Initial Submission **Exemption Date:** 07/13/2020

The project referenced above has been declared exempt from most requirements of the human subject protections regulations as described in 45 CFR 46.104 or 21 CFR 56.104 because it meets the following federal requirements for exemption:

2018 - 3 (i.A): Research involving benign behavioral interventions in conjunction with the collection of information from an adult subject through verbal or written responses or audiovisual recording when the subject prospectively agrees to the intervention and information collection and the information obtained is recorded by the investigator in such a manner that the identity of the human subjects cannot readily be ascertained, directly or through identifiers linked to the subjects. - 3 (ii) If research involves deception, it is prospectively authorized by the subject.

The determination of exemption means that:

- **You do not need to submit an application for continuing review. Instead, you will receive a request for a brief status update every three years. The status update is intended to verify that the study is still ongoing.**
- **You must carry out the research as described in the IRB application.** Review by IRB staff is required prior to implementing modifications that may change the exempt status of the research. In general, review is required for any *modifications to the research procedures* (e.g., method of data collection, nature or scope of information to be collected, nature or duration of behavioral interventions, use of deception, etc.), any change in *privacy or confidentiality protections*, modifications that result in the *inclusion of participants from vulnerable populations*, removing plans for informing participants about the study, any *change that may increase the risk or discomfort to participants, and/or* any change such that the revised procedures do not fall into one or more of the regulatory exemption categories. The purpose of review is to determine if the project still meets the federal criteria for exemption.
- All ***changes to key personnel*** must receive prior approval.
- **Promptly inform the IRB of any addition of or change in federal funding for this study.** Approval of the protocol referenced above applies only to funding sources that are specifically identified in the corresponding IRB application.

Detailed information about requirements for submitting modifications for exempt research can be found on our website. For modifications that require prior approval, an amendment to the most recent IRB application must be submitted in IRBManager. A determination of exemption or approval from the IRB must be granted before implementing the proposed changes.

Non-exempt research is subject to many regulatory requirements that must be addressed prior to implementation of the study. Conducting non-exempt research without IRB review and approval may constitute non-compliance with federal regulations and/or academic misconduct according to ISU policy.

Additionally:

- All research involving human participants must be submitted for IRB review. **Only the IRB or its designees may make the determination of exemption**, even if you conduct a study in the future that is exactly like this study.
- **Please inform the IRB if the Principal Investigator and/or Supervising Investigator end their role or involvement with the project** with sufficient time to allow an alternate PI/Supervising Investigator to assume oversight responsibility. Projects must have an eligible PI to remain open.
- **Immediately inform the IRB of** (1) **all serious and/or unexpected adverse experiences** involving risks to subjects or others; and (2) **any other unanticipated problems involving risks** to subjects or others.
- **Approval from other entities may also be needed.** For example, access to data from private records (e.g., student, medical, or employment records, etc.) that are protected by FERPA, HIPAA or other confidentiality policies requires permission from the holders of those records. Similarly, for research conducted in institutions other than ISU (e.g., schools, other colleges or universities, medical facilities, companies, etc.), investigators must obtain permission from the institution(s) as required by their policies. **An IRB determination of exemption in no way implies or guarantees that permission from these other entities will be granted.**
- Your research study may be subject to post-approval monitoring **by Iowa State University's Office for Responsible Research.** In some cases, it may also be subject to formal audit or inspection by federal agencies and study sponsors.
- Upon completion of the project, transfer of IRB oversight to another IRB, or departure of the PI and/or Supervising Investigator, please initiate a Project Closure in IRBManager to officially close the project. For information on instances when a study may be closed, please refer to the IRB Study Closure Policy.

Please don't hesitate to contact us if you have questions or concerns at 515-294-4566 or IRB@iastate.edu.

CHAPTER 5. EFFECTS OF PASSIVE EXOSKELETON SUPPORT ON EMG MEASURES OF NECK, SHOULDER, AND TRUNK EXTENSOR MUSCLES WHILE HOLDING SIMULATED SURGICAL POSTURES AND PERFORMING A SIMULATED VASCULAR CATHETERIZATION PROCEDURE

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Abstract

Exoskeletons have shown significant benefit at reducing the biomechanical demand on muscles during repetitive lifting and overhead tasks in non-healthcare industries. However, the benefits of exoskeletons are yet to be realized in the operating room, particularly as work-related musculoskeletal disorders continue to plague surgeons. This study quantified the effect of using arm, back and trunk exoskeletons on muscle activity while assuming typical postures held in the operating room. Fourteen participants were recruited to hold a series of neck flexion, arm abduction and trunk flexion postures, followed by a simulated surgical task requiring five different trunk flexion posture levels. Participants were required to complete these tasks with and without passive exoskeleton(s). This study showed that even for postures held short time periods, exoskeletons are beneficial at reducing the demand on muscles; however, the benefit depends on

body segment and postural angle. Furthermore, for the simulated surgical task with awkward trunk flexion postures, the trunk exoskeletons showed a significant reduction in the rate of rise in back muscle sEMG (+1.365%MVC/min vs. +0.769%MVC/min for non-dominant lumbar extensor muscles, $p = 0.0108$; +1.377%MVC/min vs. +0.770%MVC/min for the dominant lumbar extensor muscles, $p = 0.0196$) over 25 minutes, consequently resulting in improved subjective discomfort scores (7.34 vs. 4.30, $p < 0.05$). The results from this study indicate that exoskeletons may be a potential intervention to combat the detrimental effects of musculoskeletal disorders in postures modeled from the performance of surgery.

5.1 Introduction

Musculoskeletal pain poses significant health and economic threat to surgeons. For instance, it has been shown, according to a survey of general surgeons by Szeto et al. (2009), that 82.9%, 68.1%, and 57.8% of suffered from neck, low back, and shoulder related problems, respectively. Furthermore, a recent nationwide survey of 775 surgeons by the society of vascular surgeons showed that 45% and 35% of the survey respondents experienced significant pain in their neck and low back, respectively (Wohlauer et al., 2019). The subjective pain score in this survey showed that the surgeons experienced a 4.4 ±2.3 out of 10 pain on the Borg CR10 scale, which is actionable (Wells, Kjellman, Harper, Forsman, & Hallbeck, 2019). Another survey results showed that a significant number (31.4%) of vascular surgeons had to seek medical help for their pain even though only 4.4% of the survey respondents reported their symptoms to employee health resources (Davila et al., 2019). In fact, musculoskeletal pain in surgeons has was linked to surgeon burnout and inefficiency, and some surgeons have even worried that their pain could lead to premature career retirement (Davila et al., 2019). Thus, as the U.S. anticipates a potential shortage of surgeons (Way, 2010), interventions must be developed to prolong existing careers.

Some ergonomic interventions have been explored over the years to address the long- and short-term effects of musculoskeletal disorders. Among such interventions are intraoperative breaks (work-rest cycles), operating room reconfiguration, and arm supports for operations that require a significant amount of quasi-static awkward postures. The principle underlying intra-operative breaks is to reduce the exposure time for the risk factors (awkward static posture and duration) responsible for developing musculoskeletal pain. Thus, surgical teams take intermittent breaks while operating to either relax or even engage in active stretching routines as a method of allowing their muscles to recover from fatigue. This technique has been shown to significantly reduce discomfort in the shoulder and hands but not in the neck in laparoscopic surgery (Engelmann et al., 2011). A similar reduction in muscle fatigue was reported in simulated laboratory studies (Dorion & Darveau, 2013; Vijendren et al., 2018) and in the operating room (Hallbeck et al., 2017) with an added advantage of improved task precision. While this technique has shown positive results at reducing surgeons' pain, a recent study has shown that it did not significantly affect musculoskeletal pain in surgeons performing laparoscopic appendectomy (Kromberg et al., 2020). Furthermore, this technique can interrupt surgical workflow just as the error increasing non-routine events (Blocker et al., 2013), and might not be feasible for emergent life-threatening procedures. Operating room reconfiguration as an ergonomic intervention is predominant in laparoscopic surgery to increase visual perception and reduce strain on neck muscles. Results from (Matern et al., 2005) and (Rogers et al., 2012) showed that keeping the screen at the surgeon's eye level resulted in the least neck muscle discomfort even though that position significantly increased operating time. Thus, it seems that optimal screen configuration will be a trade-off between "good ergonomics" and productivity. Finally, arm supports are also pertinent to laparoscopic surgery to reduce upper extremity fatigue associated with the awkward static postures. While armrests have shown positive results at reducing shoulder discomfort and muscle energy consumption in the form of oxygen uptake (Galleano et al., 2006; Jafri et al., 2013; Steinhilber et al., 2015), their static nature results in kinematic mismatches between the armrest and the surgeon's upper extremities. Furthermore, armrest intervention is limited to only procedures that require awkward upper extremity postures and small surgical fields (Abdelrahman et al., 2018). Considering the drawbacks of current ergonomic interventions, there is a need to explore different techniques that are effective at reducing musculoskeletal pain.

Exoskeletons are external wearable devices that could be an alternative intervention to address the issue of work-related musculoskeletal pain and disorders in surgeons because of the benefits they have shown in non-surgical or healthcare environments. These benefits are manifested in the form of reduced muscle activation patterns and improved muscle discomfort survey results. For example, passive upper extremity exoskeletons have been shown to minimize shoulder muscle electromyography (EMG) by approximately 33.5% in an overhead drilling task (Kim et al., 2018) and by almost a 100% in the anterior deltoid during the lifting phase of a lift-walk-stack activity (Theurel et al., 2018). Similar benefits (C.I.=14-25%MVC vs 11-17%MVC) were reported in an onsite overhead farm machinery assembly plant (Gillette & Stephenson, 2019). Furthermore, trunk exoskeletons have significantly reduced low back muscle activity (31.5% and 29.3% for iliocostalis and longissimus erector spinae, respectively) in repetitive lifting tasks (Alemi et al., 2019). The reduction in muscle activity results in improved muscle discomfort results and a significant decrease in task completion time (Kim et al., 2018), translating to improved productivity.

While passive exoskeletons have shown promising results in other industries, the field of healthcare, particularly in surgery, is yet to fully realize its benefit due to several barriers identified by (Cha et al., 2019). Their study used content analysis to identify four themes: individual characteristics, perceived benefits, environmental/societal factors, and intervention characteristics, as concerns that need to be addressed for widespread adaptation of exoskeleton interventions in the operating room. One concern that was highlighted under the perceived benefit theme in this study was that there is no evidence that passive exoskeletons can indeed reduce shoulder (or other body areas) and overall body fatigue due to performing surgery. However, a recent discomfort survey study by (Liu et al., 2018) on laparoscopic surgeons performing with and without an upper extremity exoskeleton revealed that the exoskeleton significantly reduced upper extremity fatigue by 90% (3.11 vs. 5.88). Moreover, 85% of the study participants reported pain reduction post-operation. While these results are encouraging, this study was limited to only laparoscopic surgery and seven surgeons. Extrapolating the benefits of this study to other surgical procedures needs to be premised on empirical data. Additionally, this study did not have objective measures and participants' subjective discomfort could be biased, thus using a combination of subjective assessment and sEMG could provide objective information to influence the decision to use a passive exoskeleton intervention for different surgeries. Therefore, the aim of this study was to use sEMG, combined with a subjective fatigue survey to explore the potential benefits of exoskeleton(s) in reducing the biomechanical demand associated with postures assumed in a variety of surgical procedures. It was hypothesized that the exoskeleton interventions would reduce the subjective fatigue and induce changes in sEMG measures consistent with muscle fatigue reduction.

5.2 Methodology

5.2.1 Participants

A total of 14 participants (ten males and four females) from the community within and around Iowa State University were recruited to participate in this study. The sample size was estimated with a Cohen's effect size of 0.65, an alpha value of 0.05 and a power of 0.8. The participants' ages, stature, and body masses reported as means and standard deviations are (24 ± 4 years), stature (178.0 ± 8.9 cm), body mass (72.7 ± 17.8 kg) respectively. Eleven out of the 14 participants were right-handed. These participants had no experience performing surgery but were instructed to hold postures assumed by surgeons while operating. Before participating, all participants provided written informed consent approved by Iowa State University Institutional Review Board (20-288-00). Individuals younger than 18 and older than 65 years, as well as anyone with a history of chronic back pain or current pain in the back, shoulder, neck, and arms, were ineligible to participate.

5.2.2. Apparatus

5.2.2.1 EMG

The DELSYS® Bagnoli-16 sEMG system and DE-2.1 sensors (Delsys Inc., MA) were used to sample muscle activity from the following muscles: non-dominant and dominant pairs of lumbar erector spinae (NDLES, DLES), non-dominant and dominant pairs of the lower thoracic erector spinae (NDLTES and DLTES), non-dominant and dominant pairs of the medial deltoid muscles (NDMD and DMD), and non-dominant and dominant pairs of the splenius capitis (Neck extensor muscles) (NDNEM and DNEM) as shown in Figure 5.1. Dominance was based on handedness; thus, right-handed participant had their right side as dominant and vice versa. These data were collected at a frequency of 1024Hz. The electrode sensors placement was done according to SENIAM recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). A single reference electrode (UltraStim X, Axelgaard Manuf. Co. Ltd, CA) was placed on the iliac crest.

Figure 5.1: Electrode placement for sEMG measurements

5.2.2.2 Inertial measurement units

Figure 5.2: Delsys trigno avanti EMG+IMU sensor for postural angle measurement

The Delsys trigno avanti wireless EMG system with onboard inertial measurement unit (IMU) was used to measure segment angles relative to neutral posture (see Figure 5.2). A single sensor was attached to the back of a firmly fitting baseball cap to measure neck angle, on the lateral side of the elbow joint to measure shoulder abduction angle, and in between the shoulder blades to measure trunk flexion angle.

5.2.2.3 Exoskeletons

Three exoskeletons, one for each body segment, were used in the first phase of this study. The upper arm exoskeleton (Levitate Airframe®) is a passive lightweight exoskeleton designed to provide upper extremity support for tasks involving static elevated arm posture and/ or repetitive arm motion. It has two arm supports connected via two shoulder harnesses and curved rods to a padded upper back support (Figure 5.4 left). A length-adjustable vertical section connects the padded upper back support to a padded waist harness, which can also be adjusted to accommodate varying waist circumferences. The Airframe works by evenly redistributing the weight on the shoulder muscles to the outside of the hips. It does this by using a progressive activation mechanical support system (cassette) that activates based on upper arm elevation in that the degree of support increases as a function of shoulder elevation and abduction. This system deactivates when the arm is in a neutral posture. For this study, cassette level three was used for all participants.

The Laevo V series is a passive trunk exoskeleton that works on a loaded spring principle (Bosch, van Eck, Knitel, & de Looze, 2016). It consists of two pads in the chest area for comfort, two leg upper leg pads, and one back pad. Tubes with spring-like characteristics are used to connect the pads on both sides of the body. The exoskeleton is intended to transfer forces from the lower back to the chest and leg pads, as shown in Figure 5.4 (right). The spring-like tubes partially resist trunk motion during forward bending, causing a build-up of elastic energy in them. At the onset of trunk extension, the stored elastic energy is released and applied to the chest and leg to support the trunk extensor muscles.

This study's neck exoskeleton was an early prototype built for the study researchers by the same company that manufactures the upper extremity exoskeleton. It consists of a headpiece with a connection behind it that slides over a flexible spring-like rod. The rod sits on top of the back piece of the upper extremity exoskeleton, as shown in Figure 5.7. As the head flexes away from the neutral posture, the spring-like rod flexes with the head creating a build-up of elastic energy that provides support for the head to reduce the demand on the neck extensor muscles.

5.2.2.4 Subjective Survey

The subjective survey (Figure 5.3) used for this study consisted of three 16.5 cm continuous scales between two extremes: on the left is no or insignificant body segment fatigue, and on the right significant body segment fatigue. Each scale represented one body segment (neck, shoulders, and trunk). Study participants were required to place a check mark to represent each body segment's fatigue before and after completing the study task.

Post task performance

Neck

I--|

No Fatigue Extreme Fatigue

Low back

I--|

No Fatigue Extreme Fatigue

Shoulder

I--|

No Fatigue Extreme Fatigue

Figure 5.3: Subjective fatigue survey

5.2.3 Experimental Design

This study was designed as a counterbalanced repeated measure experiment with at least 72-hours between the conditions to reduce the potential carryover effect of fatigue residuals from the previous condition. For the static posture task, the independent variables were exoskeleton intervention and segment deviation (neck flexion, trunk flexion, and shoulder abduction) angle that varied by body segment. The exoskeleton intervention had two levels: 1) performing the task while wearing the exoskeletons and 2) performing the same task without the exoskeletons, while the segment deviation angle had levels shown in Table 5.2. The dependent variable was the percentage of maximum voluntary contraction (%MVC) of the respective muscles of interest. For the simulation task, the independent variable was exoskeleton intervention, with levels similar to that of the static posture task. The dependent variables in the simulation task were the gradient (Δ) of the downward shift in the median frequency (MDF) of the EMG spectrum, the slope (Δ) of the increase in average rectified value (ARV) of the time domain EMG for each of the sampled muscles, and change (pre-to-post) in the subjective assessment of segment muscle fatigue. The muscles were classified as dominant and non-dominant based on each participant's dominant hand. A summary of the muscles of interest and the EMG dependent variables is shown in Table 5.1.

Table 5.1: EMG dependent variables for static posture and simulation task

Muscle name	Simulation task dependent variables	Static posture dependent variables
Non-dominant neck extensor muscle (NDNEM)	ΔMDF and ΔARV	%MVC
Dominant neck extensor muscle (DNEM)	ΔMDF and ΔARV	%MVC
Non-dominant medial deltoid (NDMD)	ΔMDF and ΔARV	%MVC
Dominant medial deltoid (DMD)	ΔMDF and ΔARV	%MVC
Non-dominant lower thoracic erector spinae (NDLTES)	ΔMDF and ΔARV	%MVC
Dominant lower thoracic erector spinae (DLTES)	ΔMDF and ΔARV	%MVC
Non-dominant lumbar erector spinae (NDLES)	ΔMDF and ΔARV	%MVC
Dominant lumbar erector spinae (DLES)	ΔMDF and ΔARV	%MVC

5.2.4 Experimental Tasks

The experimental tasks for this study were split into three phases. The first phase was to measure and record EMG for maximum voluntary contraction (MVC). The second phase was measure muscle activity for the segmental static postures, and the last phase was the simulated surgical task.

5.2.4.1 Maximum voluntary contractions

The maximum voluntary contraction was repeated twice for the three body segments (see Figure 5.4). For the neck extension and shoulder abduction MVCs, the participants were comfortably seated into a Kin-Com dynamometer chair before the MVCs. Each participant was instructed to extend their neck against a resistance placed behind their head for the neck extensor MVC (see Figure 5.4 left). They were instructed to abduct their shoulder maximally against a resistance at a shoulder abduction angle of approximately 90° for the shoulder MVC (see Figure 5.4 middle). The trunk MVC was completed on a roman chair in which each participant's upper lower body (below the waist) was fixed to the roman chair while extending the back against resistance (see Figure 5.4 right). Two MVCs were completed for each body segment, and EMG was concurrently sampled from the muscles of interest at a frequency of 1024 Hz. Each exertion lasted for four seconds.

Figure 5.4: MVC measurement. left) neck extension, middle) shoulder abduction, right) trunk extension. The red arrow indicates resistance direction, the blue line indicates direction of exertion

5.2.4.2 Static segmental postures

The second phase of this study involved measuring and recording the muscle activity of the neck, shoulders, and back muscles while assuming a series of static neck flexion, shoulder abduction or trunk flexion postures. These postures (see Table 5.2) were extracted from postural data from the Mayo Clinic on four surgical subspecialties: vascular surgery, laparoscopic, breast, and general surgery (see Chapter 3). These angles approximate the 25th, 50th, 75^{th,} and 95th percentile segment postures for each surgical subspecialty. Table 5.2 below shows the postural angles for each body segment. For each segment postures, the participant was instructed to either flex the neck or flex the trunk or abduct (shoulders) to the required angle and hold the pose in a relaxed manner while EMG data were collected at that posture for five seconds. The order of presentation was head-neck, followed by the upper arms, and finally, the trunk. Within these presentations, the magnitude of postural angles were randomized for each segment. The onboard inertial measurement unit ensured postural accuracy (Figure 5.2) attached at specific points (at the back of a hat for neck postures, on the lateral surface of the elbow joint for upper arm posture, and in between the shoulder blades (T8) for trunk posture). For the exoskeleton condition, participants were fitted with the exoskeleton as shown in Figures 5.5 and 5.7 and they performed these static holding task in the same manner.

Figure 5.5: Levitate (left) and Laevo (right) exoskeletons used in the static shoulder abduction and trunk flexion task

Table 5.2: Segmental static postural angles

Head-neck flexion angles	Trunk flexion angles	Bilateral shoulder abduction angles
10	5	10
15	10	15
25	15	20
30	20	25
35	25	30
40	20	40
45	35	45
55	40	50
60	45	55
70	65	60
75		65
80		70
		75
		85

5.2.4.3 Vascular surgery simulation task

The experimental task and setup simulated a vascular catheterization procedure adapted from a previous study (Tetteh et al., 2020). An adjustable height table (Figure 5.6) was used as the operating table, adjusted to each participant's elbow height for each trunk angle simulated in this study. A transparent silicone tube was taped to the table to represent the blood vessel, and marks were placed at intervals to mark insertion progress.

Figure 5.6: Height-adjustable table with taped tubes for the simulation task

Each participant was required to assume the trunk flexion postures shown in Table 5.3 for the specified durations with a minute break between task trials. The task with the intermittent rest mimicked realistic workflow as observed from catheter advancement in minimally invasive surgeries in the operating room.

Figure 5.7: Participant performing the simulation task with the neck and trunk exoskeletons layered

Table 5.3: Trunk flexion angles for the vascular simulation task

Trunk flexion angles	Duration (mins)
10	5
20	10
30	5
45	5
65	5

5.2.5 Study Description

A brief description of the study was provided, and their informed consent was obtained as the participant arrived at the lab. This was followed by research assistants gathering anthropometric data (whole-body mass, stature, elbow height, and standing knee height). Next, research assistants led the participant through a short routine stretching and warm-up session. The participants were then asked to sit in a relaxed manner while alcohol was used to clean the skin over the muscles of interest before the EMG electrodes and IMUs were attached. After the electrodes were attached over the muscles of interest, the MVCs were performed in the order described in Section 5.2.4.1. A one-minute break was provided, following each MVC exertion. During the static posture data collection, a 30-second rest was allowed between successive static postures. The participants were fitted with their respective segmental exoskeleton before completing the task on the day of the exoskeleton condition.

Participants were allowed a five-minute break between the static posture phase and the vascular catheterization simulation phase. For trial-to-trial and day-to-day repeatability, participants were required to select a comfortable foot position by the height-adjustable table, and this was marked with tape on the floor. A study assistant then instructed the participant to flex their trunk to the required angle. The participant was then signaled to begin sliding the metal cable into the flexible plastic tube of six inches every 30 seconds. They held the silicone tube with their non-dominant hand while the dominant hand was used to slide the wire through (Figure 5.6). The sliding continued until the target time was reached, and the participant was instructed to stand upright for one minute and then assume the next posture for the corresponding duration (see Table 5.3). Concurrently, the trunk angle was monitored and corrected using the IMU (see Figure 5.2) placed at the back of the participant while EMG data were sampled from all eight muscles for five seconds, twice per minute according to the configuration of the Delsys EMG system. Participants completed a subjective fatigue assessment (head-neck, trunk, and shoulders) before and immediately after completing the simulation task. This was a checkmark placed on a 16.5cm continuous visual analog scale. For the exoskeleton condition, the torso and neck exoskeleton were worn simultaneously (Figure 5.7) by each participant as they performed the simulation.

5.2.6 Data Processing

A custom-developed program in MATLAB (MathWorks Inc., MA) was used to process all EMG data. The EMG signals for each participant's muscle were rectified, and the Fast Fourier Transform (FFT) was used to transform signals from the time domain to the frequency domain, after which signals below 10Hz and above 400Hz were bandpass filtered. Finally, extraneous powerline noises were eliminated by applying notch filters at frequencies of 60Hz and harmonics. For the MVC, the signal was transformed back to the time domain, and the root mean square voltage (RMS) voltage was calculated for each muscle using a sliding window of size 512 ms (Chowdhury & Nimbarte, 2017) for the first two seconds of the EMG data. The maximum RMS voltage of all the respective muscles' windows was selected as the MVC EMG. The MVC EMG was used to normalize the EMG of the respective muscles of interest. Hence all time-domain values are expressed as a percentage of MVC. For the static postures, the EMG data for each signal was transformed back to the time domain, and the average was calculated as the dependent variable. For the simulation task, preliminary EMG analyses showed that most study participants experienced flexion relaxation of the lumbar muscles at the 65° trunk flexion angle. Therefore, data for the first four flexion angles (25-mins) were used. The median frequency (MF) was determined using a fixed window of 1024ms for the 50 data collections during the first 25-minutes of the 30-minute task. The frequency-domain data was transformed back to the time domain, and the average was calculated using the same fixed window size of 1024ms. Finally, to compute the dependent variables, a least square regression line was fitted to both the median frequencies and the normalized average rectified values, and the gradient of the fitted line for the

ARV and MF over the 25-min duration was taken as the dependent variable. A positive and negative gradient for ARV and MF respectively indicates muscle fatigue development (Cifrek, Medved, Tonković, & Ostojić, 2009).

The difference between the initial and final subjective fatigue checkmarks was considered as the change in subjective fatigue while performing the task. The distance between each of the check marks and the "no-fatigue mark" was measured using a ruler. Finally, the distance from the "no-fatigue mark" before the simulation task was subtracted from the distance from the "no fatigue mark" post-completion to obtain the change in subjective fatigue for all three body segments (neck, shoulder, and low back).

5.2.7 Statistical Analysis

All statistical analysis was done in **R studio** (version 3.5.1 "Feather Spray"). It was suspected that the body segment angle might interact with the exoskeleton condition. Thus, a two-way repeated measure analysis of variance (ANOVA) with interaction (between exoskeleton usage and segment angle) was used to determine the effect of exoskeleton usage and segment angle (neck flexion angle, shoulder abduction angle, and trunk flexion angle) on the %MVC for the static segmental posture tasks. Before conducting the ANOVA, the normality and sphericity assumptions were checked for each combination of independent variables. Logit transformation was applied for normality violation, and the Greenhouse-Geisser (GG) correction (Greenhouse & Geisser, 1959) was applied to degrees of freedom used to estimate the F-statistic when sphericity was violated. For any muscle of interest with a significant difference in the dependent variable between the exoskeleton conditions and interaction effect (between exoskeleton and segment angle) from the ANOVA, the post-hoc paired t-test was used to compare the exoskeleton effect at each level of segmental angle. The differences between %MVC at the various segment angles were back-transformed. The familywise Type 1 error rate was controlled using the false discovery rate (FDR) minimization method by Benjamin-Hochberg to adjust the p-values. Finally, for any muscle without a significant interaction effect but with a significant exoskeleton effect, paired t-test was used to test the exoskeleton's impact.

For the 30-minute simulation task, the normality of residual assumption was checked using the Shapiro-Wilk test. Consequently, the paired t-test was used to explore the effects of the exoskeleton usage for those dependent variables that did not violate the normality assumption, while the Non-Parametric Wilcoxon Signed-Rank Test was used for those dependent variables that violated the normality assumption. In all cases, the benchmark value for significance was set at $\alpha=0.05$. A similar analysis was done for the subjective segmental fatigue assessment.

For the static posture task, the normality test result showed that normality was violated for the %MVC values of the NDNEM, DNEM, NDMD, and DLTES for the static posture task; thus, the logit transformation was applied to the data from these muscles. Furthermore, except for DMD, the sphericity assumption was violated for all the muscles. Thus, the F-statistic degrees of freedom were adjusted using the GG correction (Greenhouse & Geisser, 1959).

5.3 Results

5.3.1 Static Posture

Table 5.4 shows the results from the two-way repeated measure analysis of variance for the segmental static postures. The results indicate that the neck exoskeleton provided a significant reduction of neck muscle activity on the non-dominant neck extensor muscles; however, this reduction was restricted to a range of neck flexion angles as indicated by the significant exoskeleton-flexion angle interaction. Further exploration of the effects (see Table 5.5) revealed that the neck exoskeleton prototype significantly reduced the non-dominant neck extensor activity at flexion angles from 30° to 55°.

Table 5.4: Two-way ANOVA table for static posture tasks for all eight muscles (bold* indicates a significant difference, GG indicates Greenhouse-Geisser correction)

Segment posture	Muscle	Effect	DFn	DFd	F	p-value
Neck flexion static postures	NDNEM	Cond (neck exo vs no neck exo)	1	12	6.288	**0.028***
		angle^{GG}	3.69	44.25	1.796	0.153
		(Cond*angle)	11	132	4.334	<**0.001***
	DNEM	Cond (neck exo vs no neck exo)	1	13	1.618	0.226
		angle^{GG}	3.51	45.57	1.690	0.175
		(Cond*angle) ^{GG}	3.45	44.87	2.526	0.062
Shoulder abduction static postures	NDMD	Cond (upper arm exos vs no upper arm exos)	1	13	15.218	**0.002***
		angle^{GG}	3.56	46.29	106.029	< **0.001***
		(Cond*angle) ^{GG}	3.56	46.29	5.568	< **0.001***
	DMD	Cond (upper arm exos vs no upper arm exos)	1	12	17.119	**0.001***
		angle^{GG}	13	156	76.174	< **0.001***
		(Cond*angle) ^{GG}	13	156	1.390	0.169
Trunk flexion static postures	NDLTES	Cond (trunk exos vs no trunk exos)	1	13	0.435	0.521
		angle^{GG}	2.65	34.46	27.449	< **0.001***
		(Cond*angle) ^{GG}	3.49	45.31	1.691	0.176
	DLTES	Cond (trunk exos vs no trunk exos)	1	13	2.300	0.153
		angle^{GG}	2.76	35.91	35.859	< **0.001***
		(Cond*angle) ^{GG}	3.64	47.29	2.413	0.067
	NDLES	Cond (trunk exos vs no trunk exos)	1	10	11.491	< **0.001***
		angle^{GG}	2.00	19.98	15.927	< **0.001***
		(Cond*angle) ^{GG}	2.22	22.18	3.381	<**0.048***
	DLES	Cond (trunk exos vs no trunk exos)	1	12	18.258	**0.001***
		angle^{GG}	1.84	22.09	18.193	< **0.001***
		(Cond*angle) ^{GG}	2.71	32.56	5.931	<**0.001***

The upper extremity exoskeleton (Levitate Airframe^{TM}) significantly reduced medial deltoid muscle activity with significant interaction between exoskeleton utility and abduction angle on the non-dominant medial deltoids. A similar result was observed on the dominant medial deltoids; however, the interaction effect did not reach significance. Averaged across all participants, the upper extremity exoskeleton reduced non-dominant and dominant medial deltoid activity by 31.9% (4.7%MVC vs. 3.2% MVC) and 26.5% (4.9%MVC vs. 3.6%MVC), respectively. Post-hoc pairwise t-tests for the non-dominant arm (Table 5.6) revealed that exoskeleton reduced %MCV for all angles 70° and below.

Compared to the lower thoracic extensors, the trunk exoskeleton (Laevo V) seems to provide significant support for the low lumbar extensor muscles. Using the trunk exoskeleton resulted in a 12% (4. 3%MVC vs 4.9% MVC) and 16% (4.0%MVC vs 4.8% MVC) non-significant ($p > 0.05$) muscle activity reduction in the non-dominant and dominant lower thoracic extensors respectively. Conversely, an average reduction of 33% (5.7%MVC vs. 3.8%MVC) and 32% (5.8%MVC vs. 3.9% MVC) were observed for the dominant-nondominant pair of lumbar extensors respectively. However, the significant interaction (see Table 5.4) between trunk flexion angle and exoskeleton intervention indicates that the significant reduction in lumbar muscle activity was not consistent across all angles. The post-hoc analyses (Table 5.7 & Figures 5.8-5.9) showed that the exoskeleton significantly reduce bilateral lumbar muscle activity when trunk flexion angle was above at least 15°.

Table 5.5: Simple effects analysis of the effect of neck exoskeleton utility at respective static neck flexion angles for NDNEM %MVC with fdr correction. (bold* indicates significant difference of exoskeleton usage at these angles)

NDNEM angle	F-value	p-value	p.adj
10	0.329	0.577	0.577
15	0.630	0.443	0.532
25	4.490	0.056	0.112
30	7.071	0.021	**0.050***
35	16.913	0.001	**0.012***
40	11.391	0.006	**0.018***
45	13.124	0.003	**0.012***
55	13.472	0.003	**0.012***
60	1.111	0.313	0.470
70	0.708	0.416	0.532
75	1.231	0.289	0.470
80	0.358	0.561	0.577

Table 5.6: Simple effects analysis of the effect of upper arm exoskeleton utility at respective static arm abduction angles for NDMD %MVC with fdr correction. (bold* indicates significant difference of exoskeleton usage at these angles)

NDMD angle	F-value	p-value	p.adj
10	10.798	0.006	**0.011***
15	12.706	0.003	**0.010***
20	14.257	0.002	**0.010***
25	8.826	0.011	**0.014***
30	17.872	<0.001	**0.010***
40	12.147	0.004	**0.010***
45	9.178	0.010	**0.014***
50	11.707	0.005	**0.010***
55	11.370	0.005	**0.010***
60	0.799	0.388	0.388
65	9.111	0.010	**0.014***
70	11.598	0.005	**0.010***
75	3.216	0.096	0.103
85	4.658	0.050	0.058

Table 5.7: Simple effects analysis of the effect of trunk exoskeleton utility at respective static trunk flexion angles for NDLES %MVC (columns 1 to 4) and DLES %MVC (columns 5:8). (* indicates statistical significance of exoskeleton usage at these angles)

NDLES angle	F	p	p.adj	DLES angle	F	p	p.adj
5	1.346	0.273	0.273	5	0.814	0.385	0.385
10	3.646	0.085	0.109	10	2.972	0.110	0.123
15	6.238	0.032	**0.048***	15	11.898	0.005	**0.0075***
20	2.450	0.149	0.168	20	9.180	0.010	**0.0129***
25	10.097	0.010	**0.018***	25	11.717	0.005	**0.0075***
35	11.496	0.007	**0.018***	35	15.585	0.002	**0.0045***
40	20.030	0.001	**0.009***	40	20.831	<0.0007	**0.0029***
45	10.634	0.009	**0.018***	45	24.394	<0.0003	**0.0029***
65	13.388	0.008	**0.004***	65	16.993	0.001	**0.0030***

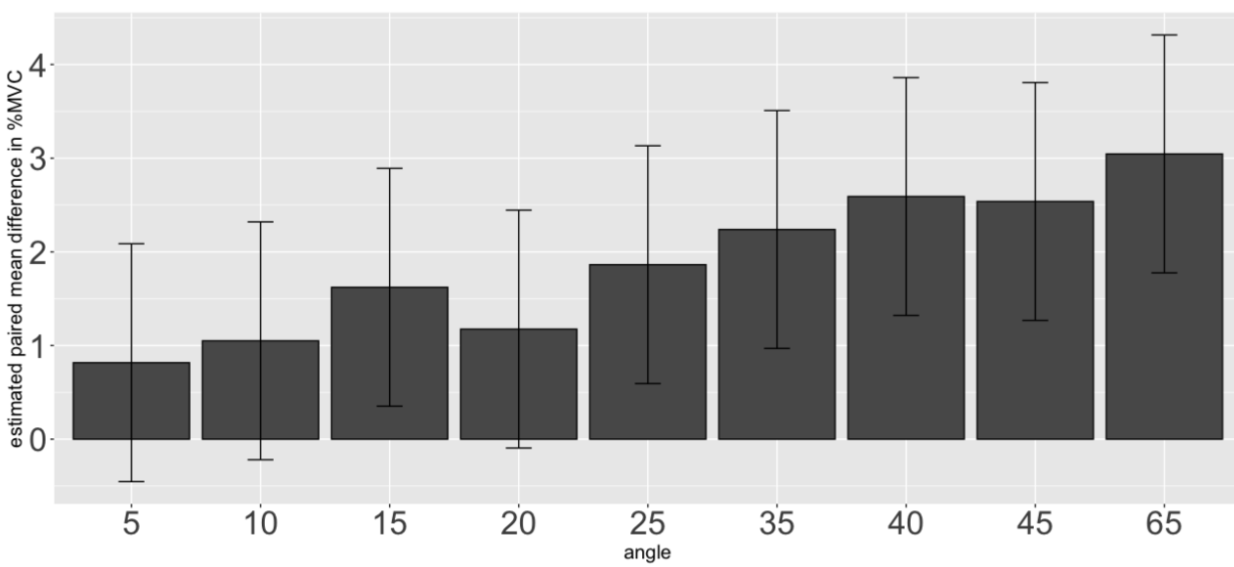

Figure 5.8: Bar plot with error bars (95% confidence interval) showing the mean difference in %MVC between the exoskeleton conditions for NDLES during the static postures. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance.

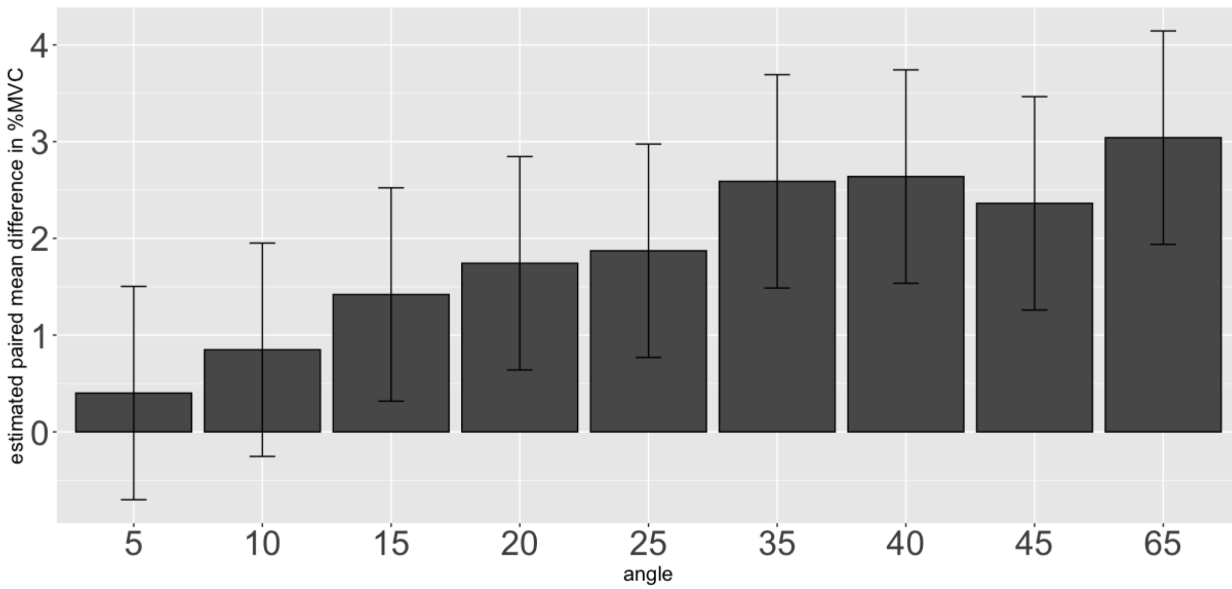

Figure 5.9: Bar plot with error bars (95% confidence interval) showing the mean difference in %MVC between the exoskeleton conditions for DLES during the static postures. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance.

5.3.2 Simulation Task

Table 5.8: Repeated measure analysis for time and frequency domain-dependent variables for simulation task comparing combined neck and trunk exos vs no exos (positive and negative **est** values in %MVC/min for ARV and Hz/sec for MDF respectively indicates beneficial effects of the exoskeleton and vice versa)

Body segment	Muscles	Differences in ARV gradient (exos vs no exos)	p-value	Differences in MDF gradient (exos vs no exos)	p-value
Neck	NDNEM	-0.1815	0.4513	0.007	0.616
	DNEM	-0.1095	0.2196	-0.018	0.688
Shoulders	NDMD	0.0180	0.2490	-0.020	0.393
	DMD	0.0177	0.5404	-0.023	0.117
Trunk	NDLTES	**0.4866**	**0.0320**	-0.014	0.299
	DLTES	0.4213	0.2453	0.019	0.315
	NDLES	**0.6388**	**0.0108**	-0.029	0.248
	DLES	**0.6500**	**0.0196**	-0.009	0.725

Table 5.8 shows that the difference in MDF gradients between the two exoskeleton conditions (head-neck and trunk exoskeletons vs no exoskeletons) was not statistically different for all the muscles of interest. In contrast, the time-domain EMG (Figures 5.10, 5.12-5.13) showed that throughout the 25-minutes (out of the 30-minutes) task performance, the

exoskeleton utility resulted in a significantly reduced upward rise in EMG in the dominant and non-dominant lumbar extensor muscles (+1.365%MVC/min vs. +0.769%MVC/min for NDLES, $p = 0.0108$: +1.377%MVC/min vs. +0.770%MVC/min for DLES, $p = 0.0196$) with the difference in exoskeleton effect seeming slightly higher for the non-dominant lumbar extensors compared to the dominant. Additionally, a similar observation was recorded in the bilateral lower thoracic erector spinae (+1.482%MVC/min vs. +0.996%MVC/min for NDLTES, $p = 0.032$: +1.427%MVC/min vs. +1.005%MVC/min for DLTES, $p = 0.245$). However, the effect of the exoskeleton was not significant on the dominant lower thoracic extensor side. All the other muscles did not demonstrate any significant differences between the two levels of our independent variable.

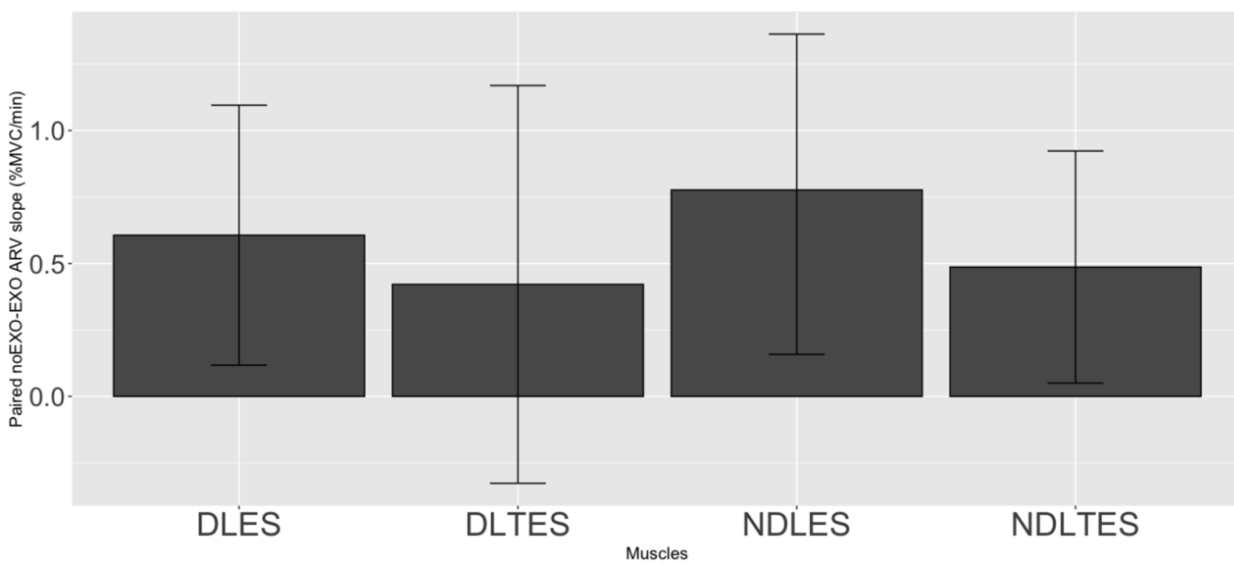

Figure 5.10: Bar plot with error bars (95% confidence interval) showing difference in ARV gradients (%MVC/min) between the exoskeleton conditions for the four back muscles. Positive bar height indicates beneficial effect of exoskeleton, and confidence interval crossing zero implies no statistical significance

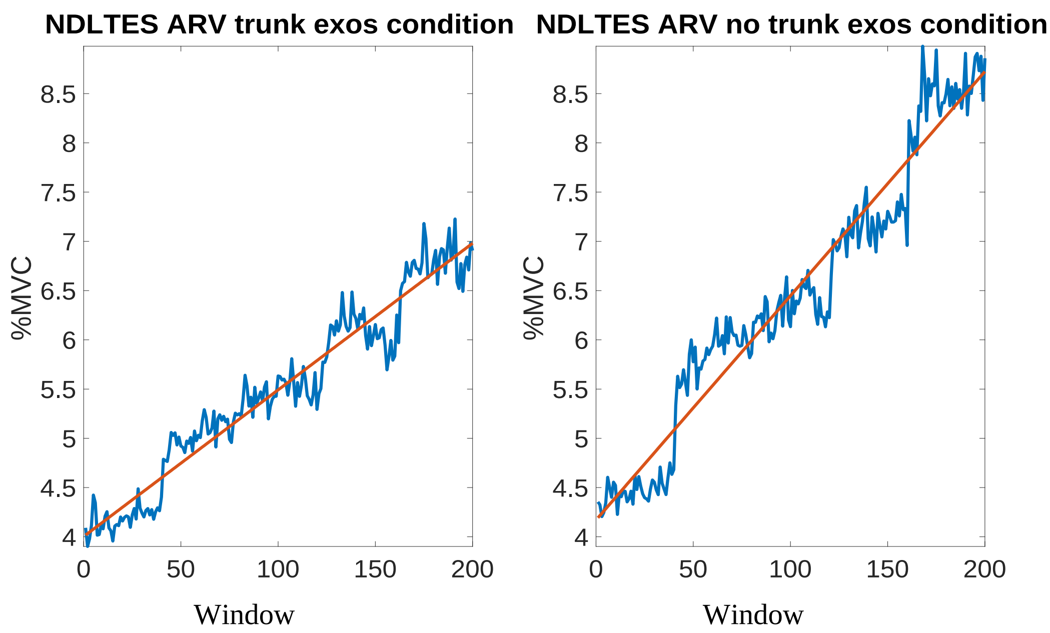

Figure 5.11: Side by side plot of the trend of ARV averaged across all participants for the non-dominant lower thoracic erector spinae. Left: exoskeleton condition, right: no exoskeleton condition. (Window 0 = time 0, window 200 =time 25mins)

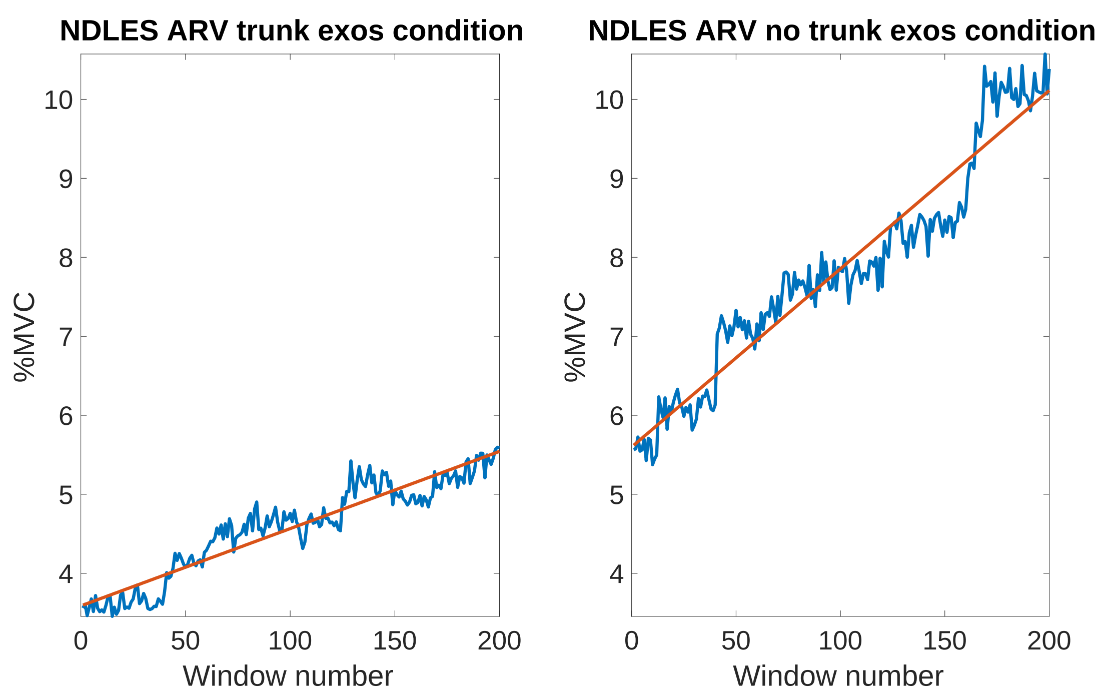

Figure 5.12: Side by side plot of the trend of ARV averaged across all participants for the non-dominant lumbar erector spinae. Left: exoskeleton condition, right: no exoskeleton condition (Window 0 = time 0, window 200 = time 25mins)

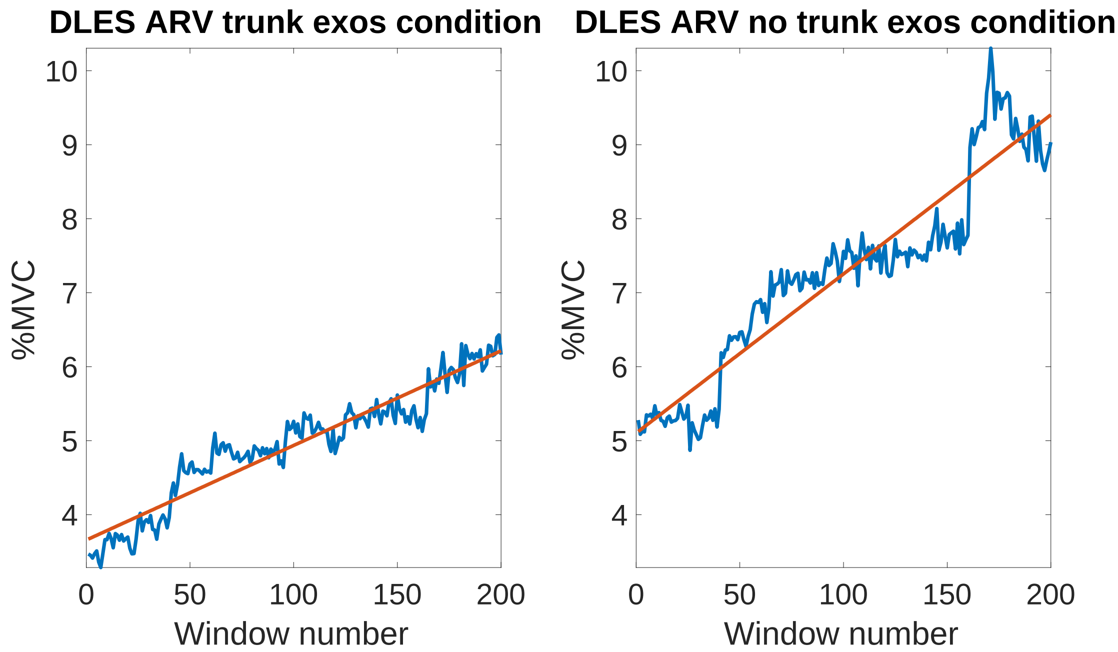

Figure 5.13: Side by side plot of the trend of ARV averaged across all participants for the dominant lumbar erector spinae. Left: exoskeleton condition, right: no exoskeleton condition. (Window 0 = time 0, window 200 = time 25mins)

Table 5.9: Paired t-test results for change in subjective fatigue (bold with asterisk implies significant exoskeleton effect, and positive est means exoskeleton helped to reduce fatigue)

	Neck	Upper arms	Trunk
est (mean difference in subjective fatigue ratings)	0.686	1.343	3.025
p-val	0.345	**0.024***	**0.003***
Lower CI	0.979	2.559	3.622
Upper CI	-0.828	0.209	1.221

The subjective fatigue survey (Table 5.9 and Figure 5.14) showed that our study participants experienced less fatigue in the shoulders and trunk while wearing the exoskeletons to perform the task compared to while not wearing it.

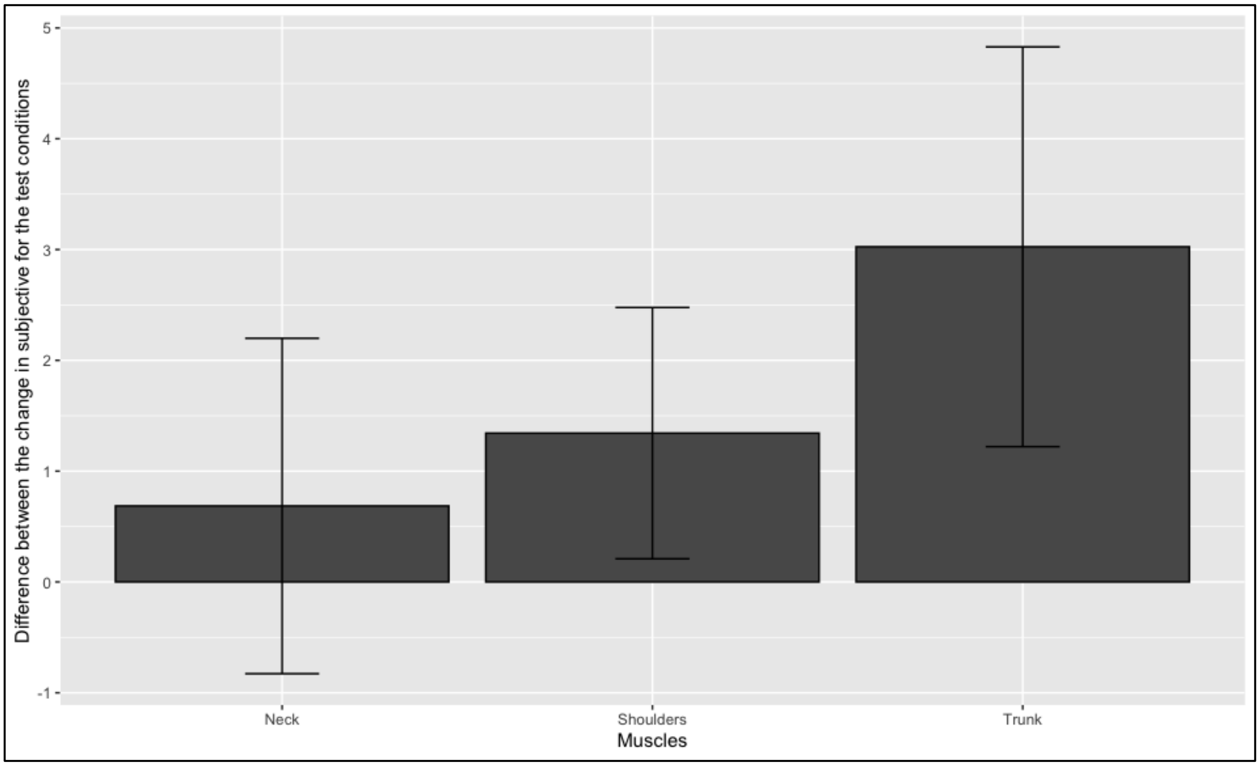

Figure 5.14: Bar plot with error bars (95% confidence interval) showing the mean difference between the change in subjective fatigue for the body segments between the exoskeleton conditions. Positive bar height indicates exoskeleton showed an increase in comfort.

5.4 Discussion

The aim of this study was to explore the ability of segmental exoskeletons to reduce the demand on muscles of different body segments in postures assumed in a variety of surgical procedures. This study shows that even for posture held a short period of time, the benefit of exoskeleton utility depends on the muscle of interest and the postural angle assumed. Furthermore, for trunk postures held for a relatively extended period, the trunk exoskeleton reduces the demand on the muscle by providing support for the torso. Collectively, these results highlight the potentially positive impact exoskeletons will have on reducing the biomechanical demand on muscles which may translate to a reduction in musculoskeletal disorders in surgeons when deployed appropriately.

5.4.1 Static Posture Task

The neck extensor muscles showed that the neck exoskeleton prototype provided support for only the non-dominant side of the neck extensor muscles; this support was limited to flexion angles ranging from 30° to 55°. The one-sided benefit provided by the exoskeleton to the non-dominant extensors could be attributed to the prototype neck support device's design. The headpiece is designed to have a metal component behind that rides over the connection between the mast and headpiece to allow for clockwise and counterclockwise neck rotation. It was observed that while flexing the neck, the metal component tended to slide in the counterclockwise direction. This observation could be the reason behind the one-sided. Hence an improved design may help to restore support for both neck extensors.

As shown in the neck flexion static postures row of the two-way ANOVA in Table 5.4 and the post hoc analyses in Table 5.5, the neck exoskeleton's benefit was limited to flexion angles ranging from 30° to 55° only for the non-dominant neck extensor muscle. This limited benefit could be ascribed to the flexion-relaxation phenomenon (FRP), a phenomenon in which the neck extensor electric activity goes silent towards maximum flexion (Maroufi, Ahmadi, & Mousavi Khatir, 2013). As shown by (Nimbarte, Zreiqat, & Chowdhury, 2014), this phenomenon occurs around 60° neck flexion, making it impossible to isolate any significant difference electromyography as observed in this study. Furthermore, neck flexion angles below 15° and below have been shown to elicit no considerable electromyography (Chaffin, 1973), hence showing no significant difference in neck extensor EMG in those angles with and without support.

While the shoulder exoskeleton showed beneficial effects at reducing shoulder muscle activity, the benefit seemed to differ between the dominant and non-dominant arms. Specifically, a significant exoskeleton-angle interaction effect was observed for non-dominant arm but not for the dominant hand. This observation may be attributed to difference in muscle activation patterns between dominant and non-dominant arms. The dominant hand is used for most tasks and thus has evolved to have fine and consistent muscle activation patterns, compared to the non-dominant side which may have variable muscle activation patterns. This difference in muscle activation patterns could be a possible reason for the difference in benefit observed from the exoskeleton support for the dominant and non-dominant shoulders. While static postures were not held for an extended period, the results underscore and complement the finding from (Liu et al., 2018), who tested the levitate Airframe in the operating room and reported an 85% reduction in subjective shoulder fatigue.

It is interesting to note that the trunk exoskeletons did not significantly reduce the activity of the lower thoracic trunk extensor muscles; however, a significant exoskeleton effect and interaction were found for the lumbar muscles (Table 5.4). Previous studies have shown that occupational low back pain predominantly affects the lumbar region (Szeto et al., 2009); and one way of preventing its occurrence is to reduce the physical demand of occupational tasks through work redesign or provision of assistive equipment (Al-Otaibi, 2015). The significant reduction in biomechanical demand of the lumbar muscles aligns with the goal of reducing low back pain and highlights the role exoskeletons can play in the short term to curb low back pain

5.4.2 Surgical Simulation Task

In the simulated vascular task, the results indicate that using the trunk exoskeleton significantly reduced the rise in time domain EMG of the dominant and non-dominant lumbar extensor muscles by approximately 44%. This finding points to trunk support exoskeletons' capability to reduce trunk bending moment in non-neutral postures (Kazerooni, Tung, & Pillai, 2019, and aligns with previous studies that have attempted to quantify the benefits of trunk exoskeletons. For instance, a study by (Bosch et al., 2016) on the Laevo trunk exoskeleton's effect while performing an assembly task at a 40° trunk posture showed that the exoskeleton reduced the trunk EMG by 35%-38%, which translated to a three-fold increase in endurance time. Furthermore, in a simulated patient transfer task (Hwang et al., 2021) using three different transfer techniques, trunk support exoskeletons were found to, on average reduce lumbar EMG by 11.2%. The relatively low percentage difference in EMG can be attributed to the study task. The patient transfer task is physically demanding compared to a wire insertion task in the current study and the Purdue pegboard assembly task (Bosch et al., 2016). Thus, the potential benefit of exoskeletons may be related to the task being performed.

The time domain EMG comparison between the dominant and non-dominant lower thoracic extensors showed that the exoskeleton significantly reduced the rise in time domain EMG on the non-dominant side compared to the dominant ($p = 0.032$ vs. 0.245 for NDLTES vs. DLTES). Interestingly, a similar asymmetry pattern was observed in a previous study from which the current study task was based on (Tetteh et al., 2020). The authors reported an asymmetric impact of fatigue development and attributed it to the slight torso twist while holding the tube with the left hand and sliding the wire with the right hand. This led to the left side of the trunk extensor muscles bearing most of the extensor moments, hence fatiguing quickly. While the Tetteh et al., (2020) speculated about the impact of handedness on the significant asymmetry because data on handedness was not recorded during the data collection phase, The current study recorded that information, providing empirical evidence on the influence of hand dominance on potential benefit of trunk exoskeleton intervention.

The frequency-domain analyses, while showing a negative estimate (indicative of beneficial effects of the exoskeleton, see Table 5.8), did not exhibit statistical significance. This could be attributed to the design of the experimental task. The trunk flexion angle increased with each bout of task performance, increasing the trunk extensors' initial moment to maintain the posture. This leads to an initial increase in median frequency (Cifrek, Medved, Tonković, & Ostojić, 2009), followed by a drop as potential fatigue sets in. This initial rise median frequency at the start of each bout potentially confounds the decline in median frequency when the posture was held, and fatigue begins to develop. Furthermore, it is hypothesized that the predominantly 5-minute bout may not have allowed enough time for the median frequency to drop considerably to indicate the trunk exoskeleton's effect, hence the insignificant difference in median frequency drop across the 25 mins task performance.

The trunk exoskeleton’s significant effect on time-domain EMG rise (see Figure 5.10) was reflected in the analysis of the results from the back subjective fatigue survey (see Fig 5.14), which shows that the participants thought that while fitted with the trunk support, they experienced less low back fatigue than when not wearing the device. These survey results underscore recently published literature on the attempt to introduce exoskeleton interventions into healthcare. As reported under the perceived benefit theme of identifying barriers for successful exoskeleton implementation in the operating room, Cha et al. (2019), healthcare workers will adopt exoskeletons if they think the intervention will reduce the physical fatigue and ultimately reduce musculoskeletal symptoms associated with their work. Moreover, a recent survey of Finnish nurses (Turja et al., 2020) reported that the success of exoskeleton usage in the healthcare sector would depend on the perceived benefit associated with its use. Interestingly, Bosch et al. (2016) reported subjective discomfort/fatigue results similar to that of this study when testing a trunk exoskeleton's effects in an assembly task requiring a 40° trunk posture. Thus, this study's subjective result indicates that the exoskeletons may have a high chance of succeeding as an ergonomic intervention in the healthcare sector.

Certain limitations need to be considered when generalizing the results of this study. First, the study was conducted in a laboratory study where environmental conditions can be controlled to isolate the exoskeleton's effect. This might not be the case in an operating room where environmental conditions are difficult to control. Furthermore, college-aged students were used in the study instead of actual surgeons. While this population may differ from surgeons, the repeated measure study design focused on quantifying the segmental exoskeletons' beneficial effects.

5.5 Conclusion

This study shows that for static segmental postures held for short periods, exoskeletons help reduce the demand of respective muscles; and the benefit is a function of the postural angle assumed. This beneficial effect ranges from ranges from 1.6%MVC in small deviation angles of the body segments to about 3.2%MVC in extreme deviation angles. Furthermore, for an extended task involving awkward trunk postures with a head-neck and trunk exoskeleton intervention, the trunk exoskeleton reduced the demand on the low back muscles. The reduction in demand seemed to be influenced by handedness. Collectively, these results indicate that exoskeleton interventions could be key assets in the fight against work-related musculoskeletal disorders amongst surgeons. Nonetheless, their deployment should be targeted at procedural tasks that expose surgeons to the causal factors of work-related musculoskeletal disorders.

Author Contributions

Conception and design: ET, GM, MSH

Data collection: ET

Writing the article: ET

Statistical analysis: ET

Interpretation: ET, GM, MSH

Critical revision of the article: ET, GM, MSH

Final approval of the article: ET, GM, MSH

Overall responsibility: ET

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Appendix: IRB Approval Memo

IOWA STATE UNIVERSITY
OF SCIENCE AND TECHNOLOGY

Institutional Review Board
Office of Research Ethics
Vice President for Research
2420 Lincoln Way, Suite 202
Ames, Iowa 50014
515 294-4566

Date: 12/15/2020

To: Gary Mirka

From: Office of Research Ethics

Title: **Exploring the benefit of exoskeleton technologies on muscle fatigue development in the operating room**

IRB ID: **20-288**

Submission Type: Modification **Review Type:** Expedited

Approval Date: 12/15/2020 **Approval Expiration Date: N/A**

The project referenced above has received approval from the Institutional Review Board (IRB) at Iowa State University according to the dates shown above. Please refer to the IRB ID number shown above in all correspondence regarding this study.

To ensure compliance with federal regulations (45 CFR 46 & 21 CFR 56), please be sure to:

- **Use only the approved study materials** in your research, including the **recruitment materials and informed consent documents that have the IRB approval stamp.**
- **Retain signed informed consent documents for 3 years after the close of the study,** when documented consent is required.
- **Obtain IRB approval prior to implementing any changes** to the study or study materials.
- **Promptly inform the IRB of any addition of or change in federal funding for this study.** Approval of the protocol referenced above applies only to funding sources that are specifically identified in the corresponding IRB application.

- **Inform the IRB if the Principal Investigator and/or Supervising Investigator end their role or involvement with the project** with sufficient time to allow an alternate PI/Supervising Investigator to assume oversight responsibility. Projects must have an eligible PI to remain open.

- **Immediately inform the IRB of** (1) **all serious and/or unexpected adverse experiences** involving risks to subjects or others; and (2) **any other unanticipated problems involving risks** to subjects or others.

- IRB approval means that you have met the requirements of federal regulations and ISU policies governing human subjects research. **Approval from other entities may also be needed.** For example, access to data from private records (e.g., student, medical, or employment records, etc.) that are

protected by FERPA, HIPAA, or other confidentiality policies requires permission from the holders of those records. Similarly, for research conducted in institutions other than ISU (e.g., schools, other colleges or universities, medical facilities, companies, etc.), investigators must obtain permission from the institution(s) as required by their policies. **IRB approval in no way implies or guarantees that permission from these other entities will be granted.**

- Your research study may be subject to post-approval monitoring **by Iowa State University's Office for Responsible Research.** In some cases, it may also be subject to formal audit or inspection by federal agencies and study sponsors.

- Upon completion of the project, transfer of IRB oversight to another IRB, or departure of the PI and/or Supervising Investigator, please initiate a Project Closure to officially close the project. For information on instances when a study may be closed, please refer to the IRB Study Closure Policy.

If your study requires continuing review, indicated by a specific Approval Expiration Date above, you should:

- **Stop all human subjects research activity if IRB approval lapses**, unless continuation is necessary to prevent harm to research participants. Human subjects research activity can resume once IRB approval is re-established.

- **Submit an application for Continuing Review** at least three to four weeks prior to the **Approval Expiration Date** as noted above to provide sufficient time for the IRB to review and approve continuation of the study. We will send a courtesy reminder as this date approaches.

Please don't hesitate to contact us if you have questions or concerns at 515-294-4566 or IRB@iastate.edu.

CHAPTER 6. GENERAL CONCLUSION

Exploring the potential use and benefits of exoskeletons in healthcare environments, notably, the operating room has garnered enormous interest in healthcare ergonomics due to the detrimental effects of musculoskeletal disorders on healthcare and the promising results that exoskeletons have shown in industries other than healthcare. Creating a standardized technique to determine if deploying exoskeletons for optimum benefit, backed with empirical data on benefits is the first step to introducing exoskeletons into the operating room and the healthcare environment. This dissertation presents three individual studies that address the need raised by the previous sentence.

The first study (Chapter 3) was a proof-of-concept study that quantified intraoperative posture and workload in vascular surgery, using a combination of postural data and subjective surveys. The results from the analyses of segmental posture showed that the neck and low back were in unfavorable postures (37.1º ±12.7º and 18.1º±6.7º flexion, respectively) across the surgeries studied, leading to considerable pain and discomfort (2.8±0.4 and 2.1±0.4 out of 10) in those segments. Furthermore, adjunctive equipment such as surgical loupes exacerbated the risk exposure of the neck. This indicates that exposure to risks of musculoskeletal disorders varies across body segments based on the particular surgery under consideration. Hence, for optimal benefits, exoskeleton interventions must be targeted at body segments with significantly high exposure to risk factors.

The study in Chapter 4 focused on developing a standardized technique to identify intraoperative surgical postures that would benefit from exoskeletons based on segmental kinematics. First, the results from that study showed that a cluster of segmental kinematics data could be used to recommend exoskeleton intervention for surgical procedures. Second, this study showed that the postural fatigue accumulated while performing surgical tasks regardless of body segments is one factor in recommending exoskeleton intervention.

Chapter 5 was a laboratory study exploring the ability of segmental exoskeletons to reduce the demand of neck, shoulder, and trunk muscles in a series of static postures and while performing a simulated vascular surgery task. This was to explore the potential short and extended-duration benefits of different exoskeletons (back, arm and neck) for awkward segmental postures. The results showed that for even brief periods of awkward posture, the exoskeleton effect on muscular activity varied as a function of body segment and postural angle. For example, the exoskeleton significantly reduced muscle activity ($p < 0.05$) for non-dominant neck extensor muscles for neck flexion angles ranging from 30°-55°. Also, trunk exoskeleton significantly reduced ($p< 0.05$) low back muscle activity for trunk flexion angles 25° and greater. For extended period of trunk flexion (as in a vascular surgery simulation task) while wearing a neck and trunk exoskeleton, the trunk exoskeleton significantly reduced the rise in lumbar EMG (+1.365%MVC/min vs. +0.769%MVC/min for NDLES, $p = 0.0108$: +1.377%MVC/min vs. +0.770%MVC/min for DLES, $p = 0.0196$). This significant reduction in the rise in lumbar EMG was reflected in the significant reduction in lumbar fatigue while performing the simulated vascular surgery task (4.31 vs. 7.33). These results indicate that for a given surgery, any exoskeleton intervention, if recommended based on the model in Chapter 3, needs to be targeted at specific body segments that are highly exposed to the causative factors of musculoskeletal disorders and the type and level of support should also be targeted. For instance, laparoscopic surgery is known to induce awkward upper arm deviation (abduction and flexion) angles (30°-75°), leading to shoulder discomfort and fatigue but with almost neutral trunk postures. Thus, recommending an upper exoskeleton intervention for laparoscopic surgery would be a recommended and feasible intervention compared to a trunk exoskeleton as our results indicate that upper arm exoskeletons provided significant reduction in shoulder sEMG in the range of upper arm angles like that of laparoscopic surgery.

The findings from this study provide a framework for deploying exoskeletons in the operating and offers empirical evidence to back the benefits of exoskeletons. However, this is the first step towards adopting these devices into the operating room, and subsequently, the healthcare delivery. Beyond this study, several future study directions will help to shape the scope of exoskeleton intervention. Below are a handful of such directions.

1. Expanding muscle activity testing of exoskeletons on real surgeons in the operating room while performing live surgeries. The advent of compact wireless electromyography systems makes this endeavor possible.

2. Incorporating usability studies into biomechanical demand studies on exoskeletons. This will enable researchers to understand the user's perspective on their interaction with the device and provide vital data to improve current exoskeleton designs.

3. The predictive model in Chapter 4 includes segmental kinematic data from 30 surgeries from four surgical subspecialties. This model's predictive capability can be enhanced by including kinematic data from a broader range of surgical subspecialties.

Overall, the findings of this dissertation, taken as a whole are that the exposure to musculoskeletal disorder risk factors during surgery vary according to surgical task and body segment, thus, exoskeleton interventions need to be targeted based on exposure to body segments. Furthermore, the benefit of derived from exoskeleton intervention vary according to body segment and segmental postures.