EFFECT OF ROTATOR CUFF MUSCLE FATIGUE ON SHOULDER MUSCLE ACTIVATION AND POSTURE DURING DRIVING

BY

WOOJAE KIM

THESIS

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Adviser:

Assistant Professor Mariana Elizabeth Kersh

ABSTRACT

Rotator cuff (RC) muscle dysfunction impacts the ability to perform daily functional tasks, such as driving. It has been suggested that RC muscle fatigue can mimic rotator cuff tears (RCT) during sudden steering in terms of kinematics. It has also been found that two RC muscles (infraspinatus and supraspinatus) are highly active during driving . However, it is unknown whether fatigue of these muscles would change the kinematic strategy during driving. The aim of this research was to analyze changes in joint angle and electromyography (EMG) signals of the upper extremity in simulated driving to identify compensatory mechanism of rotator cuff muscles.

Mean, maximum, standard deviation, and range of motion (ROM) of joint angles for four degrees of freedom (shoulder plane, shoulder elevation, shoulder rotation, and elbow flexion) were examined for four steering patterns (straight, left, right, and complex) and compared between before and after fatigue. Along with kinematic analyses, EMG signals of four muscles (deltoid, supraspinatus, infraspinatus, and biceps) were measured to analyze the relationship between kinematics and muscle usage before and after fatigue .

In straight and left turns, usage of the right deltoid significantly increased ($p \leq 0.05$) in all three measurements (mean,standard deviation, and maximum) whereas in complex turn, the right bicep was used more ($p \leq 0.05$). However, kinematics in corresponding muscles did not show significant change, which indicates change in muscle usage did not impact driver's kinematic strategy. The results suggest that in simple steering, the deltoid compensates for fatigue of RC muscles while in more dynamic steering, the biceps compensate for fatigue of RC muscles. However, the extent of this compensation was minimal as activation level of infraspinatus reached close to its maximum contraction ($\sim$96.5% MVC) while non-RC muscles were generally below 30% MVC in all turns.

ACKNOWLEDGMENTS

I would like to express my special thanks of gratitude to Professor Mariana Kersh. Her guidance helped me overcome academic challenges and grow as a scholar. Her patience and kindness helped me stay motivated and become resilient at hard time. I would also like to thank all TBL lab members who supported me through two years of graduate school. I was grateful to be part of the lab. Lastly, I would like to thank my parents and friends who encouraged and supported me during my time at graduate school

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LIST OF ABBREVIATIONS

ADL	Activities of daily living
EMG	Electromyography
DOF	Degrees of Freedom
MVC	Maximum Voluntary Contraction
ISB	International Standard of Biomechanics
RPE	Rate of Perceived Exertion
UX	Upper Extremities
LX	Lower Extremities
SUR	Scapulothoracic Upward Rotation
MRI	Magnetic Resonance Imaging
RCT	Rotator Cuff Tear
RC	Rotator Cuff
ROM	Range of Motion
SP	Shoulder Plane
SE	Shoulder Elevation
SR	Shoulder Rotation
EF	Elbow Flexion

CHAPTER 1

INTRODUCTION

1.1 Clinical Motivation for Biomechanics of Rotator Cuff Muscles

Rotator cuff tear (RCT) is known to reduce an individual's range of motion in performing functional tasks [1]. Although surgery could restore one's mobility, it causes one to have diminished power and poor control and many surgeries result in re-tear [2]. In addition, the number of RCT impaired individual are more prevalent than reported due to it being asymptomatic [3]. While the change in range of motion due to RCT is well documented [2], the mechanism behind compensatory movement in alleviating muscle weakness or force imbalance is not well known [4]. A recent study suggests that the presence of rotator cuff muscle fatigue may induce adaptive movements comparable to partial or full rotator cuff tear[5]. Thus, studying fatigued rotator cuff muscles may give insight to compensatory responses in kinematics of RCT patients performing functional tasks.

1.2 Effect of Muscle Fatigue in Driving

It is claimed that 30∼50% of all driver fatalities are attributable to some form of fatigue [6]. Although the effect of fatigue in cognitive and sensory function, such as reaction time and attention span, associated with safe driving are well documented [7], the physical function of driving is less studied. Electromyography studies of muscle activation during steering have shown that the large superficial muscles (deltoid, trapezius, biceps, and triceps) are active but are not the prime movers in steering [7]. Infraspinatus and supraspinatus, two of the rotator cuff muscles, have been shown to maintain high activity (30∼50%) during steering [5].

1.3 Aims and Objectives

Disability of the rotator cuff muscles from fatigue could lead to altered muscle usage in steering which may also manifested as a change in kinematic. Little is known about how weakness or injury in these muscles would be compensated in steering and the aim of this study was to investigate if compensation would occur in terms of muscle activation and kinematic strategy after rotator muscle fatigue. More specifically, this research focuses on the relationship between shoulder kinematic and muscle usage with onset of rotator cuff muscle fatigue during steering using motion capture and EMG sensors.

CHAPTER 2

LITERATURE REVIEW

2.1 Kinematics of Daily Activities

With recent development in technologies associated with motion analysis equipment, kinetic and kinematic measurement of human movement can be recorded with accuracy and repeatability. With these technologies, the study of biomechanics of lower extremity locomotion known as "gait analysis" became popular in both clinical and research environment. However, compared to gait analysis, there were fewer studies in biomechanics of upper limb [8]. Numerous past and current studies are still developing a method to accurately measure movement of the shoulder, particularly scapulohumeral rhythm. Although there have been several studies relating the biomechanical properties of the upper extremities to its use in daily functional tasks [8], fundamental biomechanical knowledge of upper extremities is required before establishing relation between its kinematics and biomechanical properties.

However, according to van Andel et al,[9] there are several difficulties in motion analysis of the upper extremities (UX). First, there are more functional activities that can be performed in the upper extremities compare to the lower extremities (LX). Such variety makes standardizing functional activities difficult. Second, UX functional activities show larger variation in performance among the normal population than those of LX. Compared to standardized gait patterns, UX functional movement is less strict to accomplish a given task and has higher degrees of freedom yielding higher deviations. Third, shoulder joints have a larger range of motion. This makes it harder to accurately measure joint angle with soft tissue artifact from skin and cartilage. Finally, 3D movement analysis data can be challenging to communicate successfully as there are lack of standards in measurement. It is important for the data to be both biomechanically sound and informative to the clinician to support their decision making.

Despite these difficulties, there are already studies examining kinematics of the upper extremities [10, 11, 12, 13]. However, previous studies only considered elbow/hand motion and shoulder/elbow motion separately [14, 15]. It is crucial to examine both components simultaneous since the arm and hand are both major determinants in completing functional tasks. For instance, when shoulder range of motion (ROM) is increased after reconstructive surgery, the hand should be able to use the extended ROM to improve one's ability to complete functional tasks. Several studies [10, 11, 12, 13] have analyzed the complete upper extremities, but have neglected scapular motion. Recently, two studies [16, 17]have shown that measuring scapular motion by tracking sensors on the acromion is valid for humerous elevation up to 120 degreees. A follow up study by Van andel et al [18] showed 3D kinematic analysis of complete UX using the assumption and provided clinically feasible data that also follows ISB standard.

2.2 Motor Skills in Driving

Driving is an essential skill in many people's daily life [19]. According to U.S Census, of roughly 323 million people, 222 million have a valid license [20] and spend roughly 300 hours each year driving [21]. Being able to understand how people utilizes their upper body during driving can help us develop helpful intervention to improve safety and mobility of driver. Driving is more complicated activities of daily living (ADL) such as brushing teeth or combing hair.

Driving involves reacting continuously to spatial information from environment and coordinates with upper and lower limb, head and neck [22]. Spatial information would be acquired through vision, which needs to be translated to motion through motor control. Motor control is a process by which nervous system coordinates with muscle to achieve desired motion [23]. Motor control plays critical role in performing sudden steering, recovering from a skid and sudden motion in emergencies [22]. such maneuver may be a deciding factor in preventing fatal injuries.

It is well known that motor control performance decreases with aging. With decreased motor control performance, older adult prepares for movement in a different manner than what younger adults does.Some reports indicate that elderly driver tend to struggle in changing their driving behavior to adapt to complex traffic situation [19]. For example, in a study of collision simulation involving older driver, compared to younger driver, older drivers are more likely to crash in overtaking another vehicle, merging, and intersecting [24, 25, 26]. Furthermore, consensus data indicates that older driver violates more traffic controls such as heeding to stop sign or yielding the right of way [26, 27]. Although several literatures provide an insight in how those error are consequence of high mental workload [19], there are no literature regarding motor controls to our knowledge.

2.3 Driver Fatigue

Driver fatigue is a major cause of road accidents and has implications for road safety [28]. Although effect fatigue seems intuitive for many, it is important to define and analyze what fatigue is. There are various definition of fatigue [29, 30]. In 1979, Grandjean defined fatigue as a state at which a person displays reduced efficiency and reluctance to work [29]. In 1994, Brown defined fatigue as unwillingness to continue performing the task, which also involves impairment of efficiency [30]. In both cases, definition of fatigue comprises of both physical and mental symptom. Physical fatigue is considered identical to muscle fatigue whereas mental fatigue involves psychology [28].

Physical fatigue in its fundamental level is closely related to its biochemical reaction [31]. Muscular contraction yields chemical changes and produces energy. However, after numerous muscular contractions, its energy source, sugar and phosphorous, becomes depleted while lactic acid and carbon dioxide accumulates in the muscle, which makes the muscular tissue acidic. These bio-chemical reaction during muscular contraction is a complex phenomenon with multiple symptoms, such as a decline in mental concentration and motivation, reduced work output, weaker and slower muscular contraction, increased perception of pain, a decrease in EMG frequency, and increase in temperature [32, 33].

Mental fatigue is also a by-product of biochemical reaction [34]. However, exact chemical reaction is not well studied [35]. The primary symptom of mental fatigue includes feelings of inhibition and impaired activity and general sense of weariness [28]. It is believed to be affected by factors such as environment, physical activity [36], nutrition and physical health [37]. Although mental fatigue may attribute to results of our analysis, considering its relatively short experiment duration we disregarded effect of this type of fatigue. There has been analysis of various muscle bicep, forearm flexor, frontalis muscle, of sEMG, on driver fatigue but there was none in rotator cuff muscle [38].

2.4 Senior Drivers in U.S

Individuals over age of 65 are increasingly becoming larger segments of the driving population [39]. By the year 2020, 50 million elderly persons will be eligible for driving and it raises concern regarding their safety [40]. Several studies have shown that the rate of motor vehicle accident is significantly higher among senior drivers [39, 41] and the rate increases exponentially when driver is above the age of 75 [42]. A similar pattern was observed for their fatality rate [27]. In addition to physical injuries related to the accident, it causes significant financial burden to their families and community members [43]. However, it is important to maintain their driving capability as it is essential to their independence. Furthermore, the ability to drive has been closely linked to other activities of daily living. Thus, reducing or ceasing the driving may increase the risk of depression, isolation, and other mental illness [44].

The physical functioning is one of the key factors in driving performance among older individual. Physical weakness and medical condition increase the risk of unsafe driving and probability of injury in the accident. According to study by Lyman et al [41], drivers of age between 70 and 74 and 80 and older are twice and five times as likely to die in an accident compare to driver at age of 30 and 59. Association between physical functioning and driving outcomes was studied by Anstey et al [7]. The poor neck rotation increases the risk of crashing by twice. However, no significant association was found for shoulder abduction, grip strength, trunk rotation, and disability status on crash risk. Although there are logical reasons behind association between physical function and driving performance, there are little data to back this view. Only few studies have examined change in driving behavior to compensate for physical impairment and limited mobility [43].

2.5 Analysis of Upper Extremity: Scapular Rotation

The mobility of the humerus with respect to thorax is determined by the shape of glenohumeral joint and mobility of scapula with respect to thorax [45]. Thus, it is important to understand how geometry and orientation of scapula affects glenohumeral joint angle.

The scapula is a triangular shaped bone that is located behind the rib cages and moves with respect to humerus in a complex three-dimensional motion. Scapular movement is a crucial component in arm elevation because it acts as an adaptive base of support for the humerus [18]. The deviation from this motion pattern is associated with shoulder dysfunction such as rotator cuff disease, and impingement syndrome [46]. However, anatomical location and unique shape of scapula made studying of the bone in vivo difficult [18].

Traditionally, goniometers and x ray was used to study scapular rotation with respect to thorax [47]. However, x-rays are prone to errors as it projects a 3-dimensional object in 2D-space. Thus, only one rotation in scapula can be measured. To resolve this problem, several methods have been developed to measure three-dimensional scapular kinematics [47]. One of the early method involves utilizing biplanar radiograph, both with and without implanted marker. Although this method has direct access to bony landmarks, the protocol is limited due to exposure to harmful radiation. Another method is three-dimensional imaging modalities called open configuration MRI. This method is more accurate, but is expensive and unavailable.

More recent method involves a skin-based approach that digitized discrete bony landmarks that are touchable through the skin. Another skin-based approach uses magnetic tracking devices that directly captures three-dimensional scapular orientation. The static approach involves coupling a magnetic sensor to an alignment jig and dynamic approach involves attaching a magnetic sensor directly to the acromion. Although most of these methods produced satisfactory reliability, only the latest skin approach by Mcquade [47] examined accuracy issue, which may be the most important parameter in scapular measurement.

After it has become standard practice to use skin-based method for lower extremity study, several investigators evaluated accuracy of skin-based system to bone based system that utilizes bone pins or x-rays [9]. The evaluation was made for hip, knee, ankle, and foot. Although results only showed relatively small error for some measurement, average error could exceed 50 percent of the actual motion.

2.6 Analysis of Upper Extremity: Scapulohumeral Rhythm

Scapulohumeral rhythm is defined as a ratio of glenohumeral motion to scapulothoracic motion [48]. It is the kinematic interaction between the scapula and the humerus, first published by Codeman [49]. Scapulohumeral rhythm is essential in estimating scapular rotation during shoulder elevation without using invasive method. Thus, a better understanding of scapulohumeral rhythm is required to improve analysis of upper extremities.

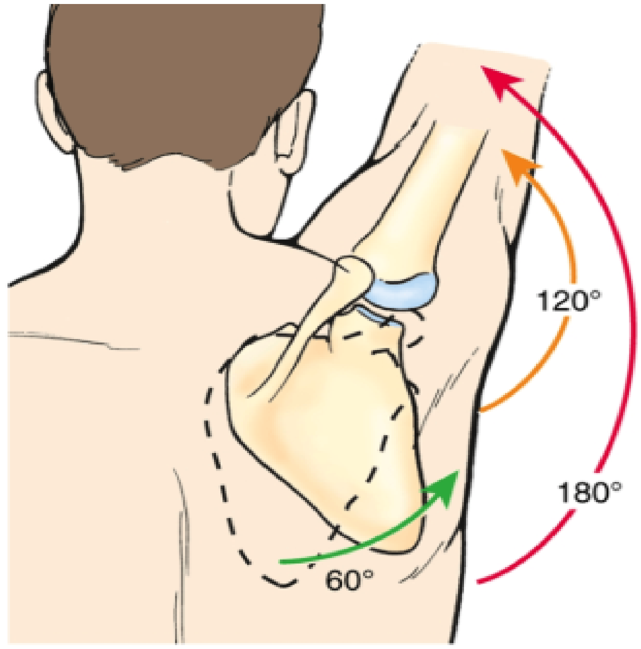


Figure 1: Scapulo-humeral rhythm. The scapula and humerus move in 1:2 ratio. If the arm was abducted by 180 degree, scapula would rotate by 60 degree, and humerus would rotate by 120 degree

The first attempt to measure scapulohumeral rhythm was done by Saunders et al [50]. They used radiography to calculate widely accepted 2:1 ratio between glenohumeral elevation and scapulothoracic upward rotation (SUR). Since then other imaging modalities such as cinematography, gonoiometry, MRI (magnetic resonance imaging), and X ray gained more popularities and were used to capture more accurate kinematics of shoulder [48]. Studies using the new technology suggested that 2:1 ratio is not consistent across the entire arc of shoulder elevation. The studies have concluded that variability in scapulohumeral rhythm exist in subjects with shoulder injuries.

According to a study done by Mcquad et al [48], the variability in scapulohumeral rhythm also occurs when an arm is subjected to a load. In the study, scapulohumeral rhythm was measured during three different state of loading: passive elevation of an arm, active elevation against the weight of the limb, heavy loading against maximum resistance. Their results show that scapulohumeral rhythm changes depending on the phase of elevation and external load on the arm. For passive elevation, the rhythm decreased from 7.9:1 to 2.1:1 as the arm elevated. For active elevation, the rhythm increased from 3.1:1 to 4.3:1 as the arm elevated. For heavy load, the rhythm increases from 3.1:1 to 4.3:1. The result shows the historical assumption of a simple linear 2:1 scapulohumeral rhythm ratio may be overly simplistic and may not accurately describe the motion in different dynamic conditions.

2.7 Biomechanical Model in OpenSim

Biomechanical models of the musculoskeletal system are often used to analyze kinematics and kinetics of a human body. The model used in our study is a dynamics model of the upper limb created by Wu et al [51] in OpenSim software. It has 5 rigid body segments and provides 10 degree of freedom (DOF). It can be used to perform inverse kinematics and inverse dynamics to calculate joint angles and moments from marker trajectory data. The model's graphical representation of the thorax, clavicle, scapula, humerus, ulna, radius, and bones of wrist and hand are shown in Figure 2. The forearm, ulna and radius, and wrist were modeled as one rigid body.

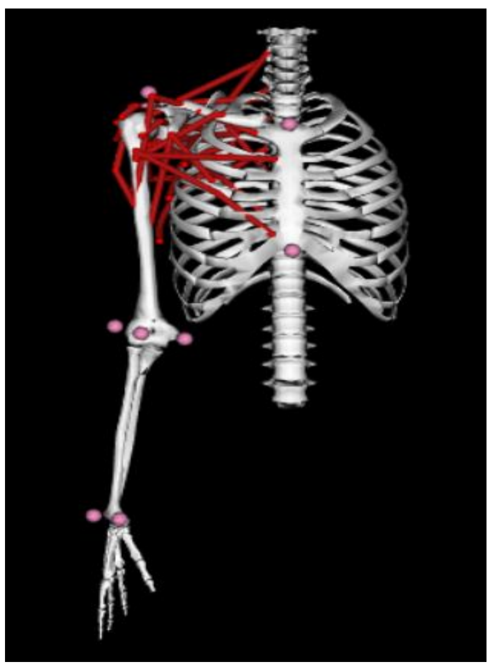


Figure 2: An image of the biomechanical model used in this study. Pink spheres indicate the bony landmark where optical marker will be attached

The model contains 10 degrees of freedom representing the wrist, forearm, elbow, and the shoulder and its coordinate system follows conventions recommended by the International Society of Biomechanics [52]. The forearm and wrist were modeled as 2 DOF universal joint. The acromioclavicular and glenohumeral joint articulate as 3 DOF constrained ball and socket joints. Sternoclavicular joints were modeled as 2 DOF universal joint. The force-generating parameters for each muscle and the kinematics of each joint were derived from the experiments [52]. The model also approximates moment arm and the muscle-tendon lengths for each of the muscle over a wide range of motion. The maximum moment-generating capacity and moment arms of each muscle group were compared to experimental data [52] to evaluate its accuracy. The model represents muscle force-generating characteristics and anthropometry of a 50th percentile adult male.

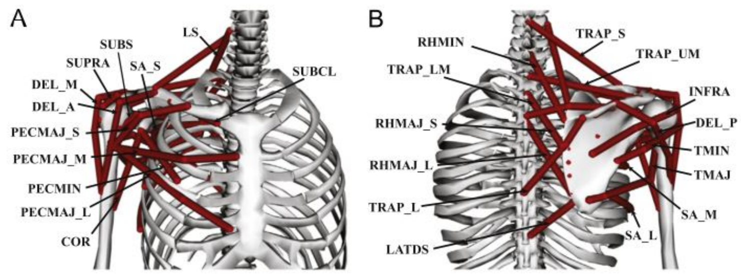


Figure 3: All muscles represented in the model.

2.8 Inverse Dynamics of Upper Extremities and Different Methods of Inverse Dynamics

Inverse dynamics is the procedure where kinematics data are used to calculate torque produced at the joint. It is an essential tool for analysis of functional human movement and an effective method to estimate how muscles are used during a movement. During performing the inverse dynamics, data such as inertial properties and motion of the limbs, and any external forces applied during the movement are required. Out of these measurements, inertial properties of the limb or body segment parameters (BSP) are particularly difficult to measure in vivo. Thus, it is usually preferred to approximate the best possible joint torque from imperfect measurement [53].

The inverse dynamics data for upper extremities are less available than that for lower extremities in literature. Many research/studies have examined the upper extremities kinematics of healthy subject performing daily functional task. However, only few studies performed dynamic analysis on kinematics data. In addition, published Inverse dynamics data has large inconsistency on predicting BSP due to incorrect estimation of these parameters. For instance, values for upper arm mass range from 4.9 to 6.2 percent of total body mass among published literature, which utilizes different techniques [54].

In addition to variation in BSP, there are difference methods in performing inverse dynamics. The conventional method involves iterative solution of the Newton-Euler equation of motion for each body segment [53]. When an angular acceleration is the only input data, the iteration begins at the top segment, successively proceeding to lower segment. In each iteration, joint torque is calculated that satisfy dynamic equilibrium. In such case, result tend to be noisy because two-differentiation on acceleration amplify the noise on input data. However, when a ground reaction force is also introduced as input, providing boundary condition for the bottom-most segment, dynamics equilibrium condition can be solved starting from the bottom segment proceeding upward. Since force plate data are more precise and less noisy, the output joint torques are more precise and less noisy. But adding the new input creates redundant equilibrium solution and to avoid this overdeterminacy, acceleration measurement needs to be discarded for top segment by introducing torque and residual force to top segment.

Another method for the inverse dynamics problem is based on optimization. According to Vaughan et al [55], overdeteriminacy from ground reaction force data can be solved using extra degree of freedom to optimize the body segment parameter. Such method provides better parameter values and enforces boundary conditions. This indirectly improves inverse dynamics problem while not affecting joint torque estimate. The other and the most sophisticated method is to calculate the trajectory of joint torque using dynamics optimization. This is known to reproduce the observed motion the best in a forward simulation. Because this method enforces time variant dynamic equation of motion, it is considered theoretically ideal method for estimating joint torque.

According to Kuo et al [53], each method has its disadvantage. The conventional Newton-Euler method produces inaccurate joint torque estimate when ground reaction force is not included and is highly sensitive to mismatch in origin reference frame of original data and ground reaction force. Dynamic optimization can solve some of this issue but is often difficult for practical uses [53].

Alternative to these three methods is to use static optimization. In this method, Inverse dynamic problem is considered overcomplete system of equation and were optimized at each time frame to calculate joint torques that best agrees with input parameters [54].

Although results from the static optimization are not theoretically precise as in dynamics optimization, it is easier and faster to solve and can resolve some issues with conventional method [54]. For instance, it is better at utilizing available data, while rejecting noise in the measurement. Also, it calculates joint torques that is completely consistent with equation of motion at each point in time with no residual force or torque. Furthermore, the static optimization works without fully matching ground reaction force data. It can be used to fix any mismatch between reference frame for GRF and motion data [54].

2.9 Structure of Glenohumeral Joint and Uses of RC Muscles in Steering

The shoulder joint or also known as glenohumeral joint is a ball and socket joint between scapula and humerus [56]. It is a major joint connecting upper limb and trunk Glenohumeral joint is said to be inherently unstable due to amount of motion it provides [56]. It is the most commonly dislocated major join in the human body [57]. According to Matsen et al [58], glenohumeral instability is defined as a condition in which unwanted translation of the humeral head occur that compromises comfort and function of the shoulder. The amount of translation that causes instability and pain varies depending on individual [58, 59].

Repetitive work or cumulative work have been related to pain and discomfort [60]. This work – discomfort relationship is linked to development of muscle fatigue [61]. Shoulder muscle fatigue has been known to induce physical change in glenohumeral joint mechanics. According to Chopp et al [62], superior humeral head excursion occurred as an effect of fatigue. In addition to excursion, fatigue of these muscle could lead to compromised function of the shoulder joint. Some studies [62] have even suggested that fatigue of these muscle emulate the effect of rotator cuff tear.

2.10 Uses of Muscles in Driving

The function of the shoulder in driving is to shift gears and to steer the wheel. Although there have been studies measuring muscle activation and kinematics of these ac-

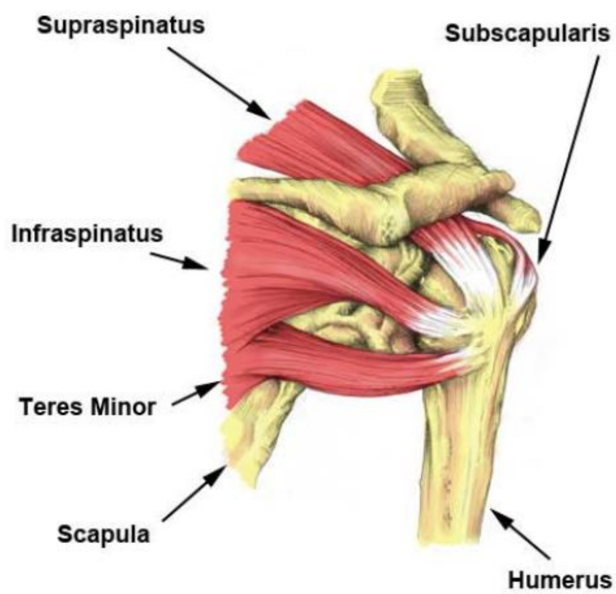
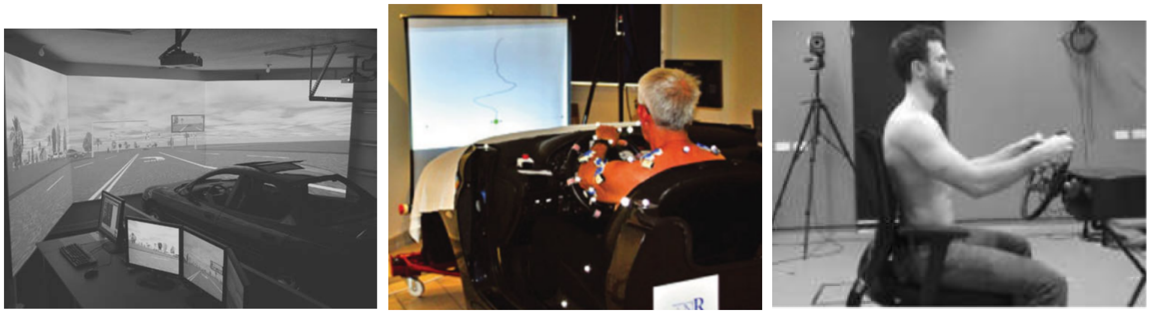


Figure 4: Anatomy of glenohumeral joint with rotator cuff muscles.

tivities, little is known about how impaired condition of shoulder muscles would affect these measurements. Fatigue or injuries to these muscles may alter muscle usage and kinematic strategy of steering the wheel. Few works have been done measuring muscle activation using surface electromyography(EMG). According to Solveig et al [5], both main flexors and extensors of the elbow were not prime movers in turning the wheel. More recent study by Pandis et al [62], concluded that regardless of sitting posture of driver, rotator cuff muscles like supraspinatus and infraspinatus maintained moderate (∼30%) to high (∼50%) activation in steering. In addition, supraspinatus and deltoid muscles had nearly two times higher activation than any other muscles of the upper limb. Thus, injury to one of these rotator cuff muscles could lead to dangerous level of muscle activation of other shoulder muscles to compensate. This may also lead to joint stability leading to different posture and kinematic strategy.

2.11 Driving Simulator and Instrumented Steering Wheel

Driving simulators experiments vary depending on their purpose. All the simulators are designed to measure the parameter of interest while trying to reproduce the most realistic design. The driving simulator in literature can be categorized into three types: (a) full-



(a) (b) (c)

Figure 5: Types of driving simulator

fledged, (b) line tracking, and (c) manually -instructed as shown in Figure 5. The full-fledged design includes simulator created in Shechtman et al [63]. The purpose of this study was to analyze effectiveness of safety protocol proposed by the Federal Highway Administration (FHWA) in a high-fidelity driving simulator. However, it has not been validated with an on-road driving and some subjects experienced simulator sickness, a type of motion sickness. Another type of simulator is a line tracking driving simulator which includes simulator created in previous studies[64, 65, 66]. In the former study, test subjects were asked to track a line scrolling downward on the screen using a steering wheel as shown in Figure 5 (b). The line was produced based on the steering angle recorded on an actual car driving around a campus. Their driving simulator was made from a wrecked car pieces purchased from a scrap yard.

This ensures a realistic static steering torque resistance on a wheel but does not account for the torque feedback (or dynamics steering torque resistance) generated while turning [67]. The latter study projected a light beam, which was synchronized to the movement of the wheel, to a movie screen in front of a 'mock-up' car. The movie screen projected a film showing a rear of a car, which created a marking, driving in different types of road. Subjects were instructed to maintain the light beam on a marking. The other type of simulator is an instrumented steering wheel [5, 68, 69, 70]. These types of simulator usually have a steering wheel mounted on a customized steering column. Depending on parameter of interest, a steering column generates either constant torque resistance on a wheel [5] or dynamic torque resistance generated from a motor on a steering column [68].

CHAPTER 3

METHODOLOGY

3.1 Subject Recruitment and Inclusion/Exclusion Criteria

Following approval from the Institutional Review Board of the University of Illinois at Urbana-Champaign, sixteen women (n=9) and men (n=7) with a mean age of 24.8 $\pm$ 2.5 were recruited from the local community.

The inclusion criteria for the subjects are the following:

- 18$\sim$30 years old
- Can consent and follow command

The exclusion criteria for the subjects are the following:

- History of known musculoskeletal disease
- History of major shoulder injury or surgery
- Recent injury to dominant arm
- Feeling of pain or discomfort during arm extension/flexion movement in dominant arm

All subjects were asked the PAR-Q questionnaire [71] for risk assessment.If the subjects answered yes to any of the questionnaire, they were excluded from the study. All subjects also signed an informed consent after being informed on the procedures and aim of the experiment.

3.2 Maximum Voluntary Contraction Measurements

Once the subjects meet the inclusion/ exclusion criteria, electromyograhy (EMG) surface electrodes (Bagnoli-8TM, Delsys Inc., USA) were attached to the middle deltoid, infraspinatus, supraspinatus and bicep brachii of their right arm according to the literature [72].To obtain the maximum voluntary contraction measurements, the subjects were the asked to exert their maximum effort under four conditions: (1) shoulder abduction at 0 degrees humeral elevation, (2) abduction at 90 degrees, (3) external rotation at 0-degree humeral rotation and (4) bicep flexion for the maximum voluntary contractions (MVCs) of the deltoid, infraspinatus, supraspinatus, and biceps respectively. All tasks were performed with resistance against a fixed wall. A one-minute rest period was given between each maximal effort and three trials per subject were recorded. EMG data were sampled at 2,000Hz. EMG data was filtered using a second-order, low-pass Butterworth filter with a 100 Hz cut-off frequency. The signals were rectified, and a linear envelope was created. EMG signals from the supraspinatus and infraspinatus were calculated using the equation [73] relating indwelling electrode and surface electrode.

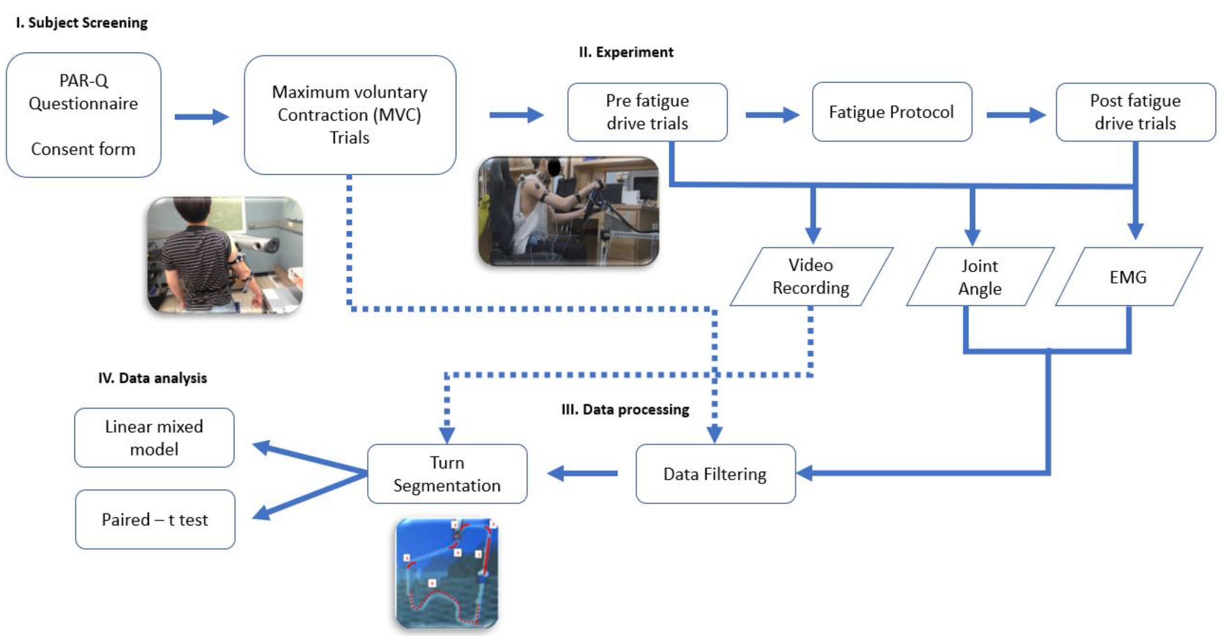
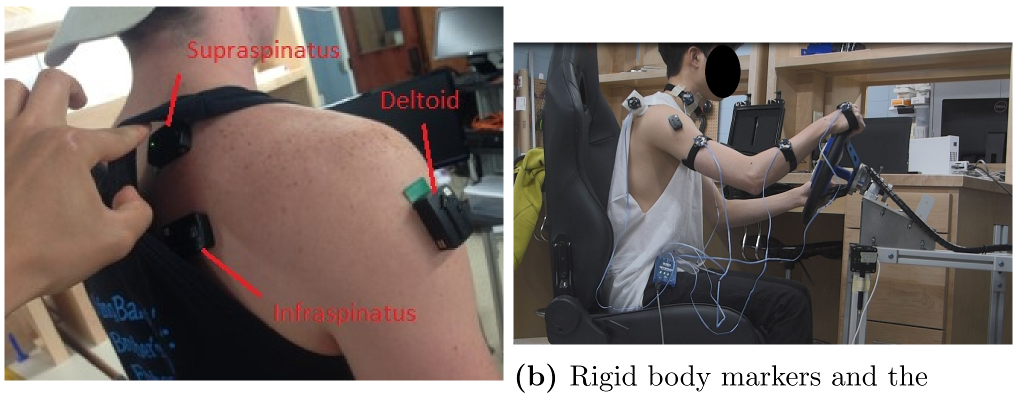


Figure 6: Experimental setup flow chart.

3.3 Digitizing Bony Landmark for Kinematic Measurements

Next, a set of motion capture rigid body markers (iRed, Optotrak, NDI) were attached to the subjects using soft cotton band at the center of the forearm, mid-humerus, and right side of the trunk. Additional custom jig with a rigid body were placed on scapula to collect motion of scapula. Lastly, bony land marks were digitized with respect to the existing rigid body markers. Sixteen virtual markers were digitized to anatomical landmarks using an established marker set [51]. Marker data was sampled at 50Hz, and filtered using a fourth-order, low-pass Butterworth filter with a 4Hz cut-off frequency.



(a) EMG sensors

(b) Rigid body markers and the driving simulator

Figure 7: EMG and Motion capture marker placment for the experiment

3.4 Steering Experiments

The subjects were asked to maneuver a vehicle through a track within a driving video game (GTProseries, Wii, Nintendo, Japan) using a custom-designed driving simulator as shown in Figure 7. The driving simulator was designed following ergonomic measurements from a real car [74]. The subjects were also asked to adjust a seat in their comfortable driving position. Vehicle travel speed was held constant at 80 mph for all trials. The subjects were also asked to have their hands in contact with the marked spot on the steering wheel at all times. Practice sessions of 3 to 6 laps were performed prior to data collection to enable participants to become accustomed to the track. After the practice session, the subjects were asked to play the game for 3 laps while EMG and 3D kinematic data were collected. A video recording of the screen was taken to determine the time frames at which the vehicle travels within pre-defined segments in the driving course.

EMG were collected for the deltoid, infraspinatus, supraspinatus, and biceps at 2000Hz. Kinematic data were collected using an active motion capture system (Optotrak, Northern Digital Inc.,Canada). Rigid body marker clusters were attached to the neck, middle of the humerus, middle of the forearm, and the back of the hand as shown in Figure 7. An additional marker cluster was attached to the flat part of the acromion. Sixteen virtual markers were digitized to anatomical landmarks using an established marker set [51]. Marker data was sampled at 50Hz, and filtered using a fourth-order, low-pass Butterworth filter with a 4Hz cut-off frequency. Data during the driving simulation were collected before and after induction of rotator cuff fatigue.

3.5 Fatigue Protocol

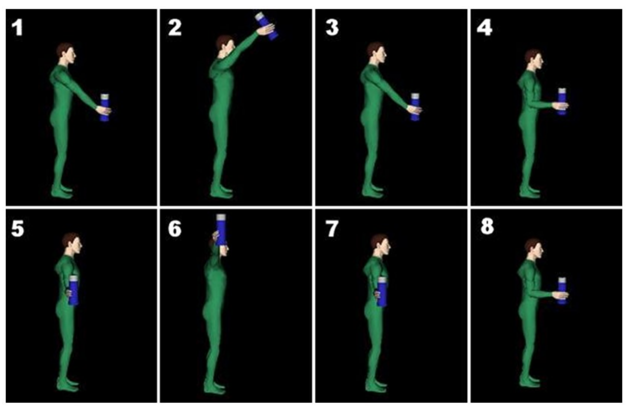


Figure 8: Fatigue protocol in "Superior humeral head migration occurs after a protocol designed to fatigue the rotator cuff"[62]

Following the driving exercise, the subjects performed a set of tasks designed to fatigue the rotator cuff muscles from a combination of overhead work (supraspinatus) and internal (subscapularis) and external rotation (infraspinatus, teres minor) [62]. Specifically, Subjects were then asked to hold 2kg at 0 degrees of horizontal abduction and 45 degrees of vertical shoulder flexion. The shoulder was then flexed to 135 degrees of shoulder flexion and then returned to the starting position. Next, the subjects bent the elbow to 90 degrees, followed by external rotation to 90 degrees of horizontal abduction. The weight was then lifted vertically until the shoulder was again at 135 degrees of shoulder flexion and then reversed to the starting position. These steps were repeated for one minute at a pace of 40 beats/min. The subjects then performed a 5-second static hold at 90 degree shoulder elevation in the scapular plane during which EMG data were collected to evaluate fatigue onset. An increase in EMG signal during these static hold was used as an evidence of fatigue. The subjects were asked to rate their perceived exertion using a Borg scale [75] scale as shown in Table 1.

If the rating was below 10, the subjects completed the 5 second static hold and perform the fatigue protocol for another minute. If the rating was 10 or the subjects had completed the fatigue protocol 5 times, the subjects proceeded to a second series of 3 laps in the driving simulator.

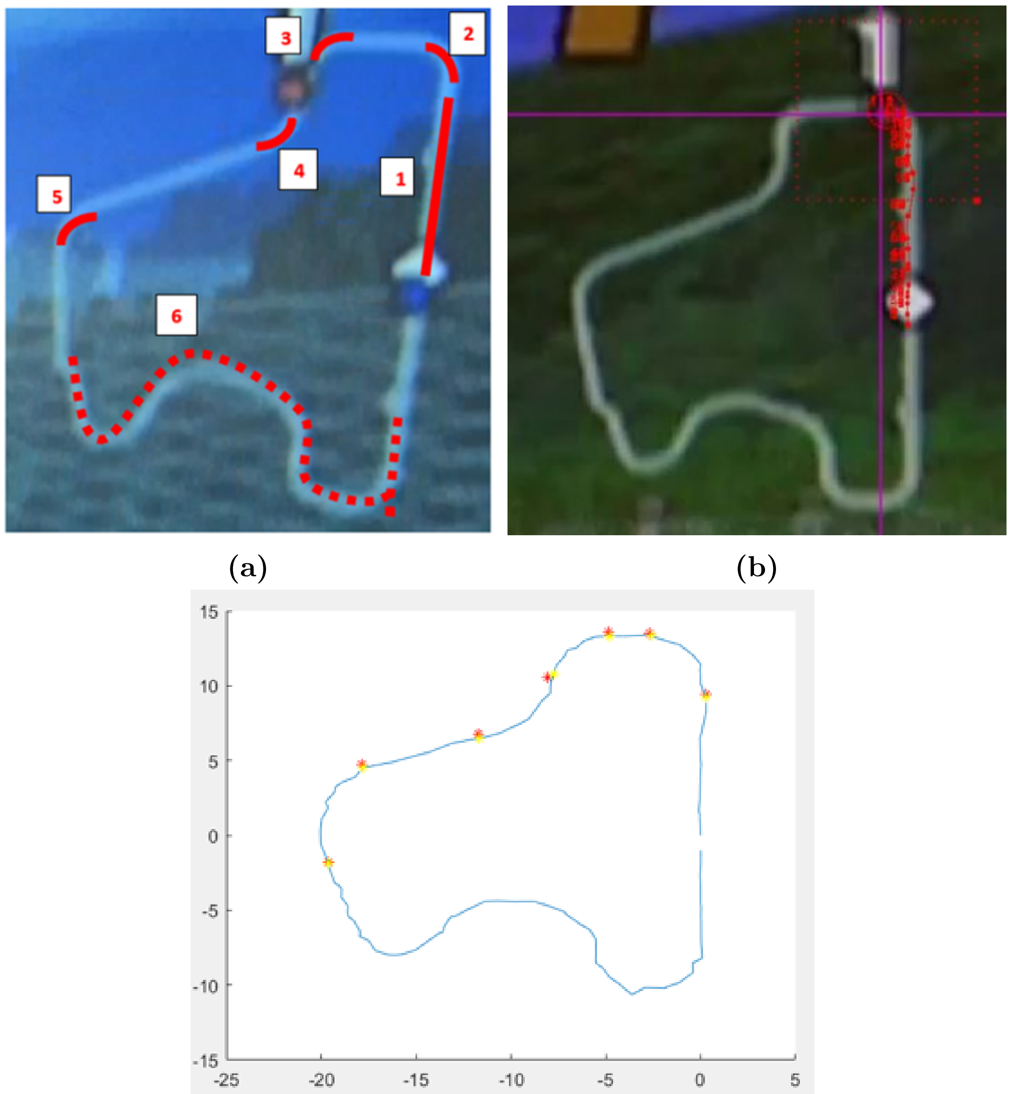
Rating	Descriptor
0	Rest
1	Very, very easy
2	Easy
3	Moderate
4	Somewhat hard
5	Hard
6	
7	Very hard
8	
9	
10	Maximal

Table 1: The Borg RPE scale chart

3.6 Kinematic Calculation

A musculoskeletal model was used to quantify joint angles during the steering experiments [51]. The model contains 10 degrees of freedom (DOF) representing the wrist, forearm, elbow, and the shoulder. The forearm and wrist were modeled as a two-DOF universal joint. The acromioclavicular and glenohumeral joint were articulated as 3-DOF constrained ball and socket joints. Sternoclavicular joints were modeled as 2 DOF universal joint. The model was scaled to each subject using marker locations from a static trial. The inverse kinematic tool was used to calculate joint angle in a coordinate system that follows the International Society of Biomechanics (ISB) standards [52].

3.7 Segmenting Driving Course



(a) (b)

(c)

Figure 9: (a) Segments shown in the game screen. (b) Tracker software tracking the vehicle in the screen. (c) MATLAB results of segmenting. Yellow points shows a closest estimate point to the red points, marked coordinate from representative trial.

3.8 Data Analysis

The kinematic trajectory and EMG signals of each trial was separated into six pre-defined segments based on the start and end of each maneuver type: straight, left turn, right turn, complex turn. An open source video analysis tool (Tracker), was used to determine the time frame for the start and end points of each segment as shown in Figure 9 (a). Using video taken of the game screen during the experiment, a red dot representing the driver's location over the course of the game was tracked to extract the spatial coordinate and associated time frame of the vehicle during each segment. The kinematic motion capture data and electromyography data was then synchronized to the start and end times for each segment.

3.9 Statistical Analysis

All statistical analyses was performed in RStudio (R Core team, Vienna, Austria). Multiple linear mixed effects models were used to test the influence of two fixed predictors (fatigue and gender) and one random predictor (subject) on joint angles in three different measurements (Mean, Maximum, and Range of motion) for the shoulder plane,shoulder elevation, shoulder rotation, and elbow flexion. The Kolmogorov-Smirnov test was used to test for normality of residuals of the models. For each fixed predictor, the null model (as a function of a random predictor Subject) was compared with the full model (as a function of one fixed predictor and the random predictor) to determine whether a fixed predictor was a significant variable in predicting joint angle. A paired-sample t-test was used to compare time-varying joint angles and EMG in pre-fatigue conditions to post-fatigue conditions to determine the time frames at which significant deviation occured due to fatigue.

CHAPTER 4

RESULTS

4.1 Effect of Fatigue on Shoulder Muscle Activation Levels

During the straight section and left turn, significant increase ($p \leq 0.05$) in muscle activation occurred in the deltoid and infraspinatus for all three measurements (Mean,Max,and ROM) as shown in Table 2. Significant decrease ($p \leq 0.05$) in muscle activation only occurred in the deltoid during right turn for all measurements whereas the other muscles during right turn showed minimal($\leq\sim$11%MVC) changes. A significant increase in muscle activation was also observed in the infraspinatus and bicep during complex turn in all three measurements. Significant increase ($p \leq 0.05$) in muscle activation of the supraspinatus only occured during left turn (p=0.001) and complex turn (p=0.001). Overall, the deltoid showed the highest number of significant changes ($p \leq 0.05$) in all turns.

Table 2: Change in EMG amplitude in %MVC for four muscles before and after fatigue at different turns. Significant difference is indicated by *.

Turns	**Measurements**	**Deltoid**		**Infraspinatus**		**Supraspinatus**		**Bicep**	
		Pre	Post	Pre	Post	Pre	Post	Pre	Post
Straight (Segment1)	Mean	7.97	9.31*	26.15	29.85*	17	17.16	2.83	2.92
	SD	3.95	4.39*	11.87	19.33*	6.68	6.55	2.94	3.12
	MAX	28.22	32.96*	64.12	74.57*	45.73	51.75	14.91	30.62*
Left (Segment3)	Mean	9.39	10.20*	30.74	38.29*	17.53	18.15	3.75	3.56
	SD	4.36	6.47*	13.58	19.87*	5.66	6.66	3.62	3.11
	MAX	25.06	28.38*	70.77	75.73*	38.54	47.82*	17.96	17.49
Right (Segment4)	Mean	8.55	7.42*	31.47	31.64	17.91	18.39	3.63	4.01
	SD	4.45	3.00*	15.29	14.79	6.51	7.04	2.98	3.25
	MAX	28.40	24.03*	84.51	83.3	47.36	46.8	19.83	19.76
Complex (Segment6)	Mean	10	10.57	31.31	36.12*	18.27	18.71	4.54	4.75*
	SD	5.78	6.13	14.64	17.36*	6.38	7.22	4.04	4.85*
	MAX	78.81	49.55*	87.94	95.03*	47.84	75.80*	23.36	29.22*

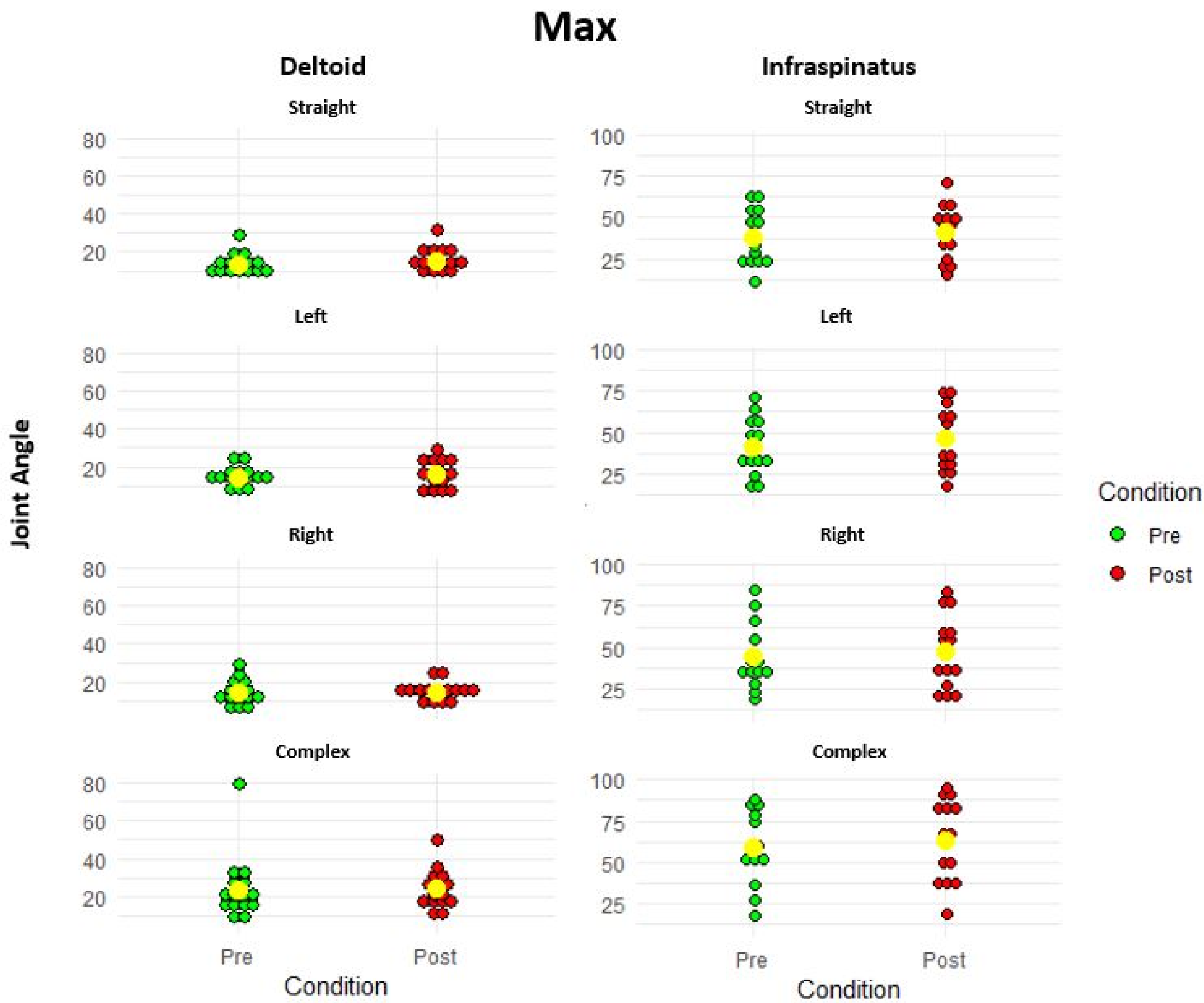


Figure 10: Maximum EMG of the deltoid and infraspinatus in %MVC over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate an mean of max %MVC over 16 subjects.

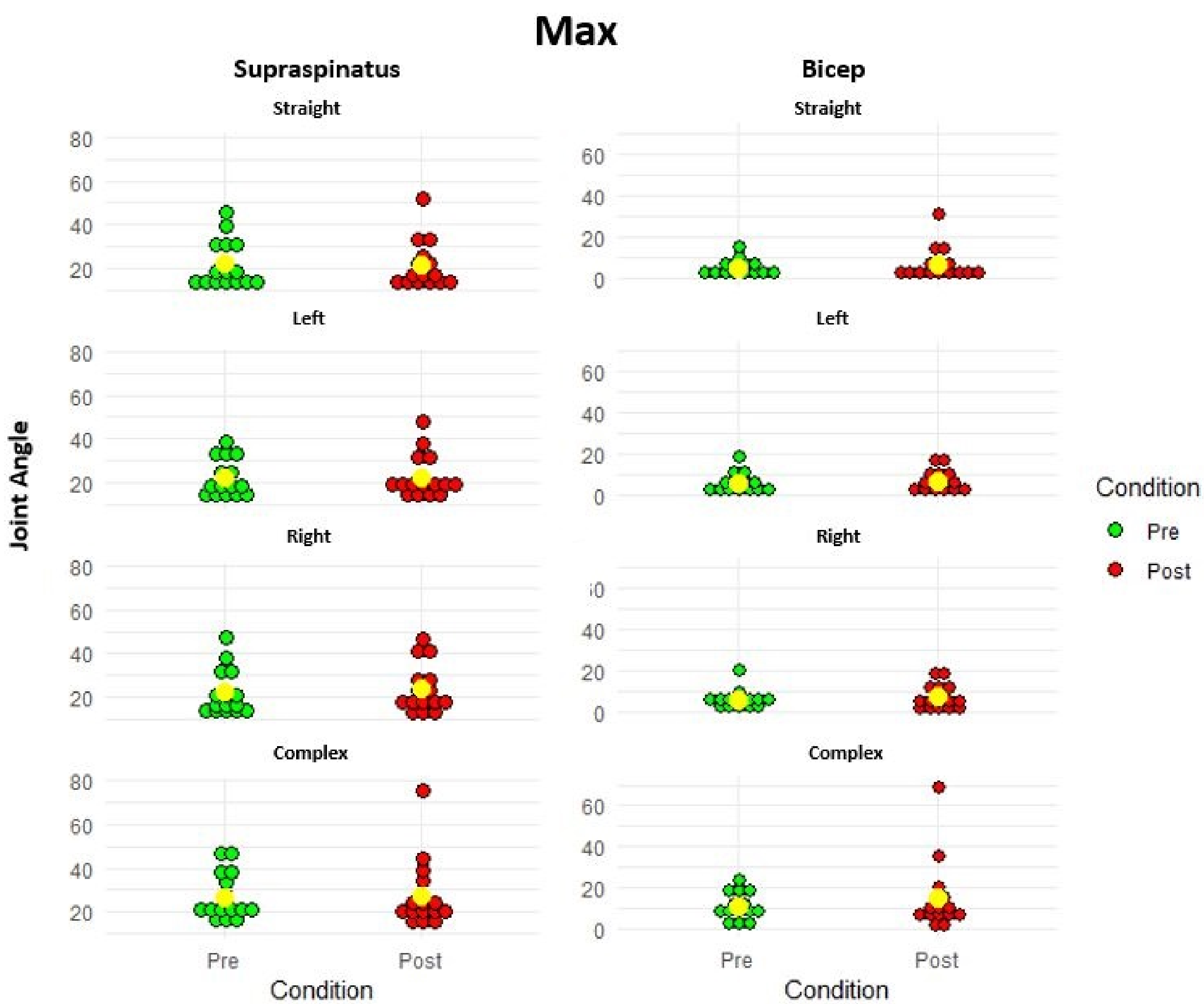


Figure 11: Maximum EMG of the supraspinatus and bicep in %MVC over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate an mean of max %MVC over 16 subjects.

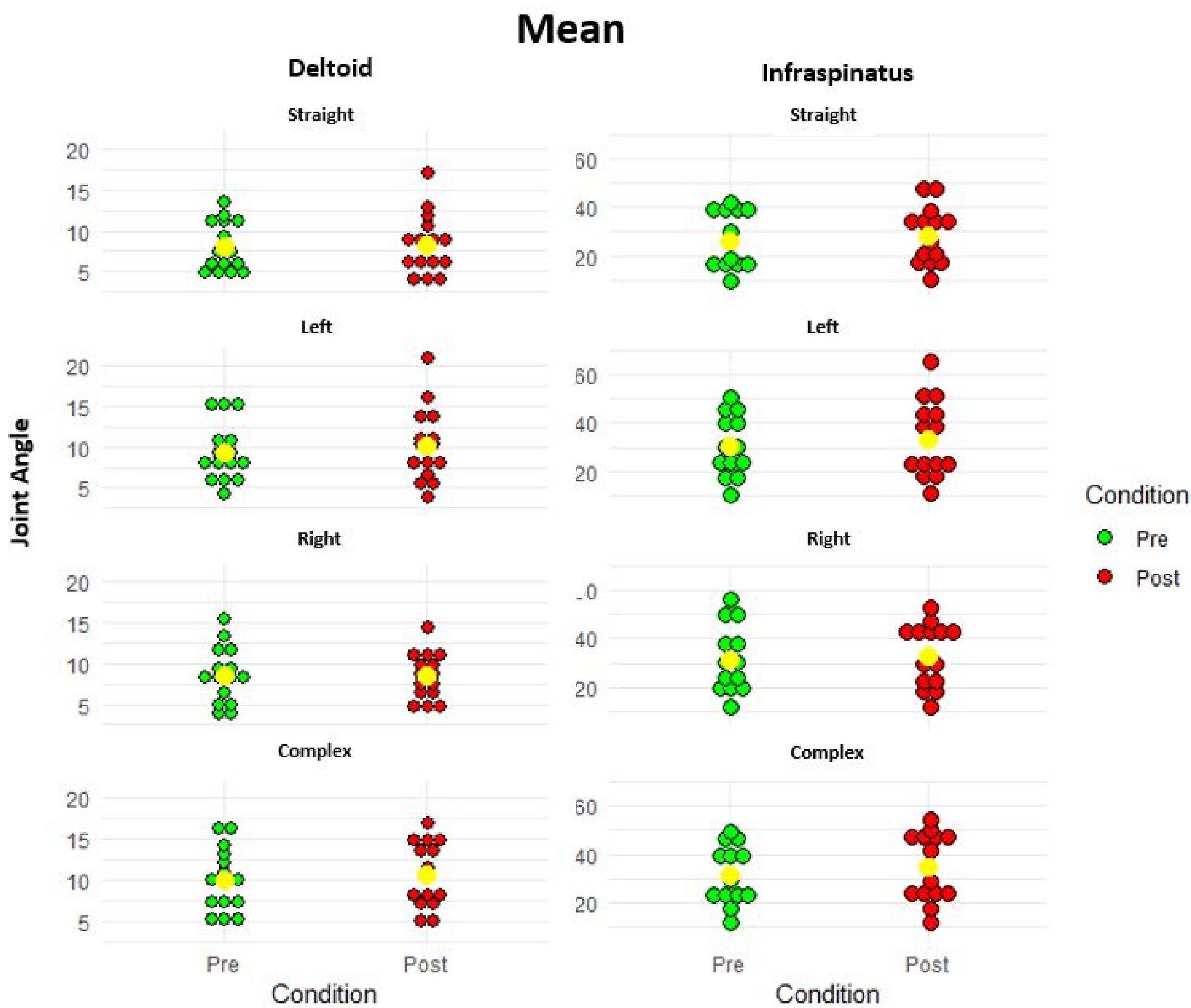


Figure 12: Average EMG of the deltoid and infraspinatus in %MVC over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate a mean of average %MVC over 16 subjects.

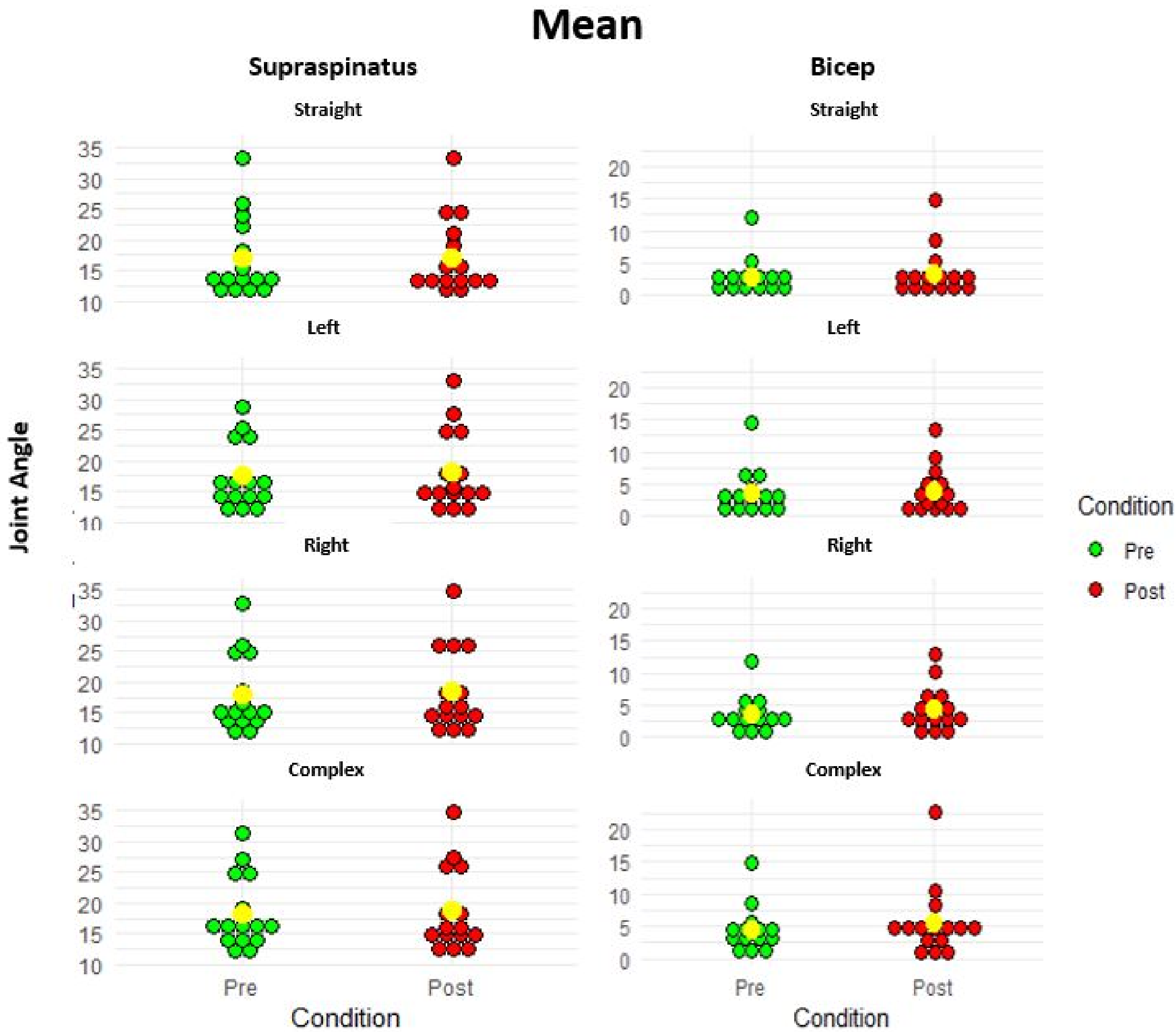


Figure 13: Average EMG of the supraspinatus and bicep in %MVC over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate a mean of average %MVC over 16 subjects.

4.2 Effect of Fatigue on Shoulder Joint Angles

Overall, there was no significant effect of fatigue on joint angles for all degrees of freedom except in shoulder plane as shown in Table 3. However, all of the significant changes ($p \leq 0.05$) in shoulder plane occurred in range of motion (ROM). The values in range of motion were all significantly increased ($p \leq 0.05$) regardless of the turns. In shoulder elevation, maximum joint angle (p=0.001) and range of motion (p=0.003) decreased significantly. In elbow flexion, range of motion (p=0.002) increased significantly only at the straight section.

Table 3: Change in joint angle in degree (°) for four degrees of freedom before and after fatigue at different turns. Significant difference was indicated by *

Turns	**Measurements**	**SP**		**SE**		**SR**		**EF**	
		Pre	Post	Pre	Post	Pre	Post	Pre	Post
Straight (Segment1)	Mean	-10.63	-12.05	50.68	48.93	20.63	22.11	63.17	67.81
	SD	12.1	16.53	12.03	13.04	14.04	14.55	28.4	28.9
	MAX	16.98	17.23	76.59	74.26	64.62	60.28	114.75	121.05
	ROM	53.49	80.26*	48.33	47.47	63.47	59.47	108.96	112.34*
Left (Segment3)	Mean	-12.54	-14.1	57.96	56.65	23.27	25.54	60.53	64.02
	SD	11.59	13.83	12.99	13.53	13.71	14.71	26.03	26.37
	MAX	15.35	12.22	87.74	79.16	68.06	70.25	113	117.92
	ROM	57.36	82.69*	63.95	65.08	66.57	69.57	103.48	103.72
Right (Segment4)	Mean	-7.05	-9.56	45.17	44.35	19.95	22.23	65.46	69.33
	SD	13.48	18.36	12.55	13.09	12.03	14.13	25.71	25.88
	MAX	19.32	18.74	85.14	77.02*	73.11	73.16	110.19	117.02
	ROM	62.08	90.9*	61.7	55.43*	71.93	72.13	99.29	103.52
Complex (Segment6)	Mean	-10.1	-11.8	51.79	49.82	23.05	25.29	62.65	66.42
	SD	12.87	17.37	14.86	15.23	13	15.16	26.17	25.76
	MAX	19.6	19.21	91.76	89.25	75.23	78.63	113.2	121.15
	ROM	60.7	91.13*	73.59	71.58	74.22	78.07	113.29	106.33

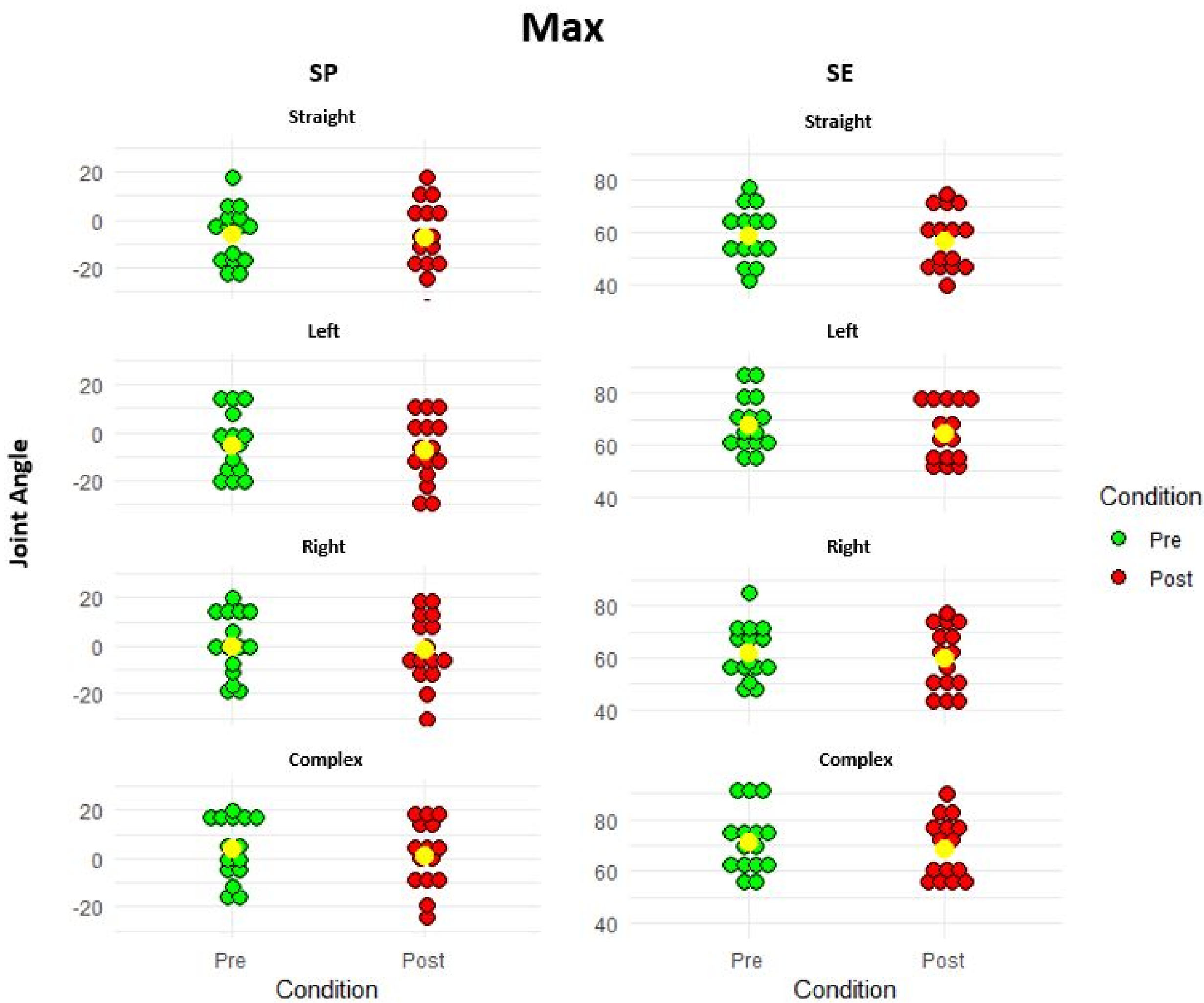


Figure 14: Maximum joint angle in degree (°) over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate an mean of maximum joint angle over 16 subjects.

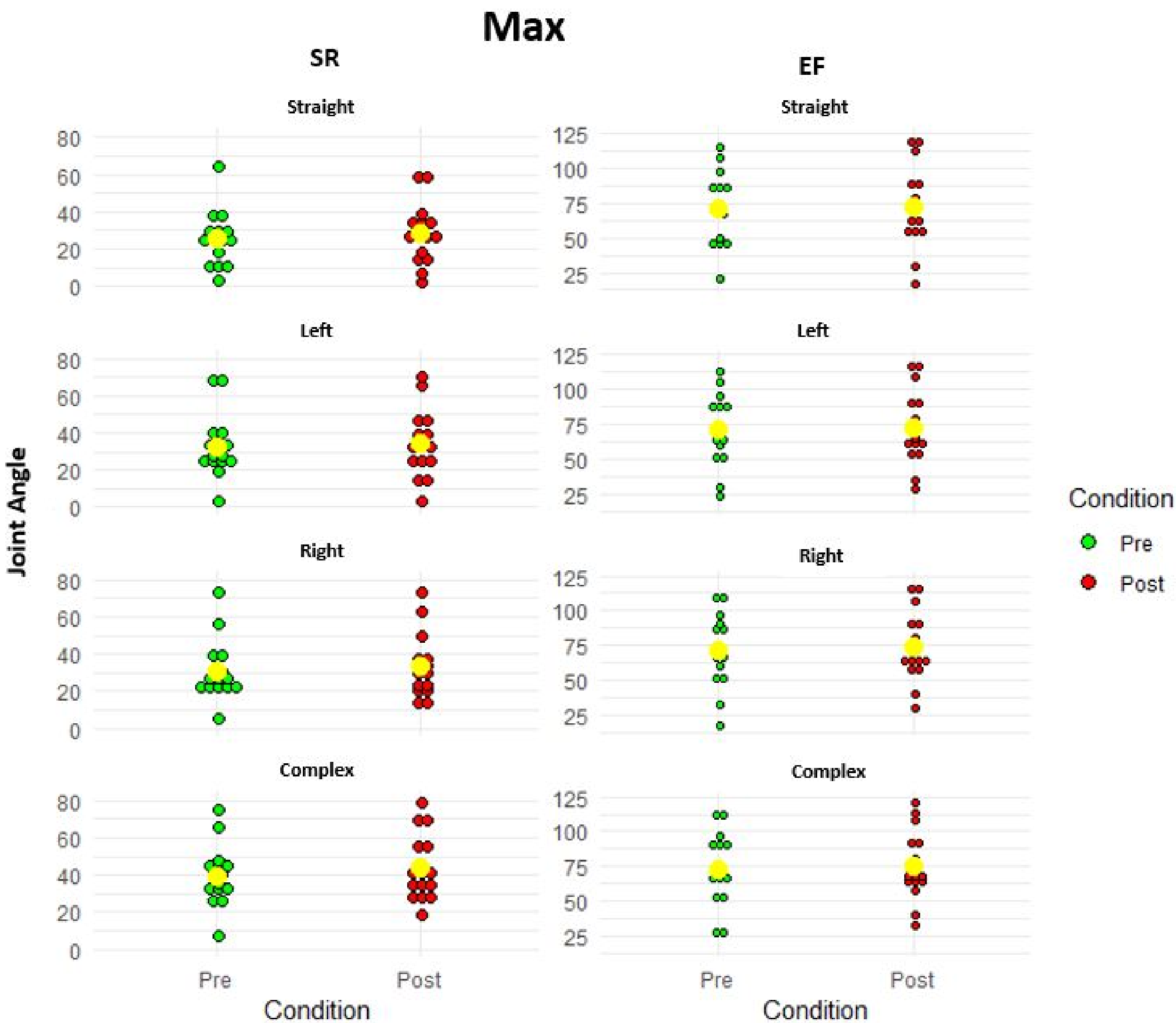


Figure 15: Maximum joint angle in degree (°) over the whole trajectory for each individual before and after the fatigue at different turns. Yellow dot indicate an mean of maximum joint angle over 16 subjects.

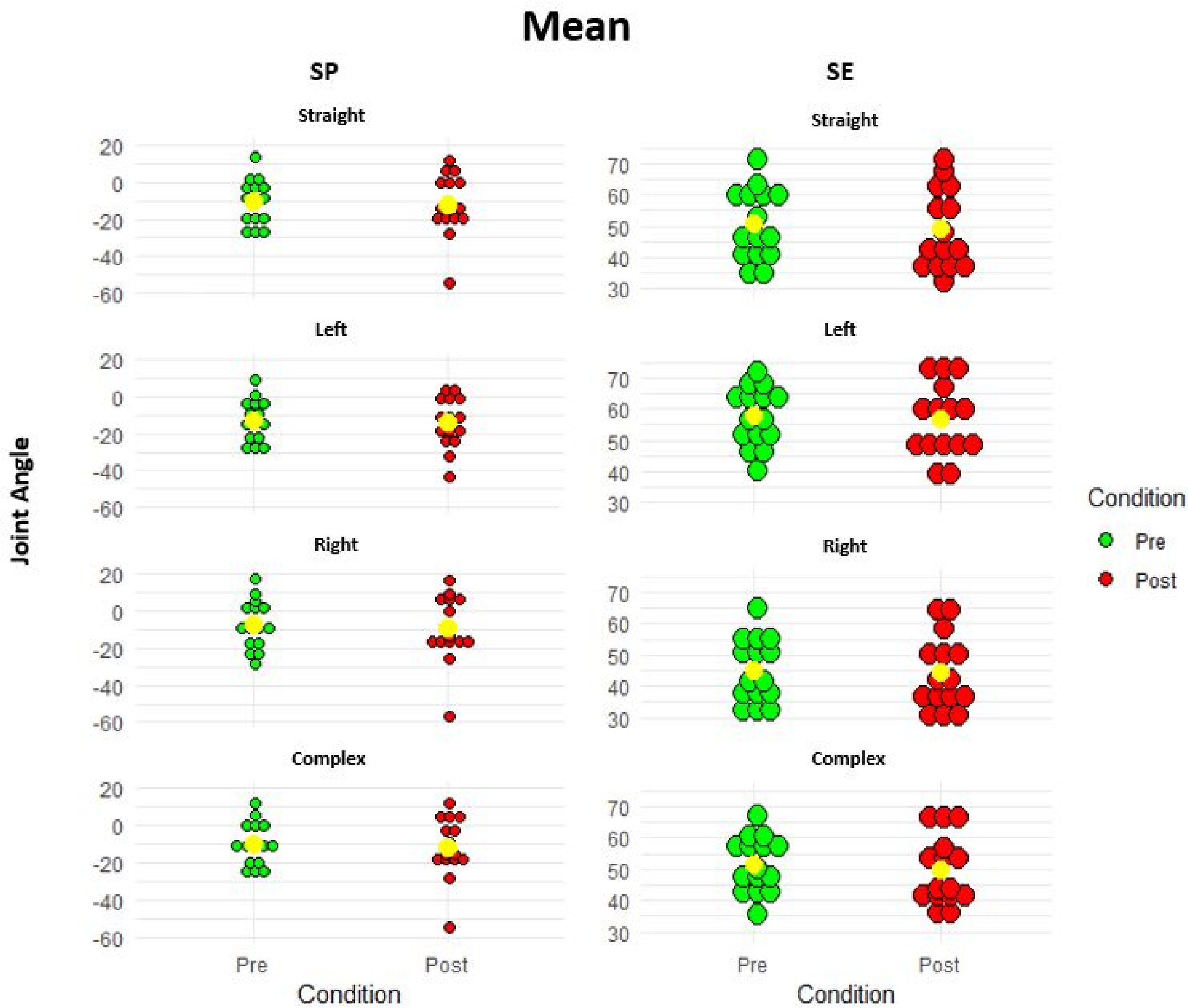


Figure 16: Average joint angle in degree (°) over the whole trajectory for each individual before and after the fatigue at different turns.Yellow dot indicate a mean of average joint angle over 16 subjects.

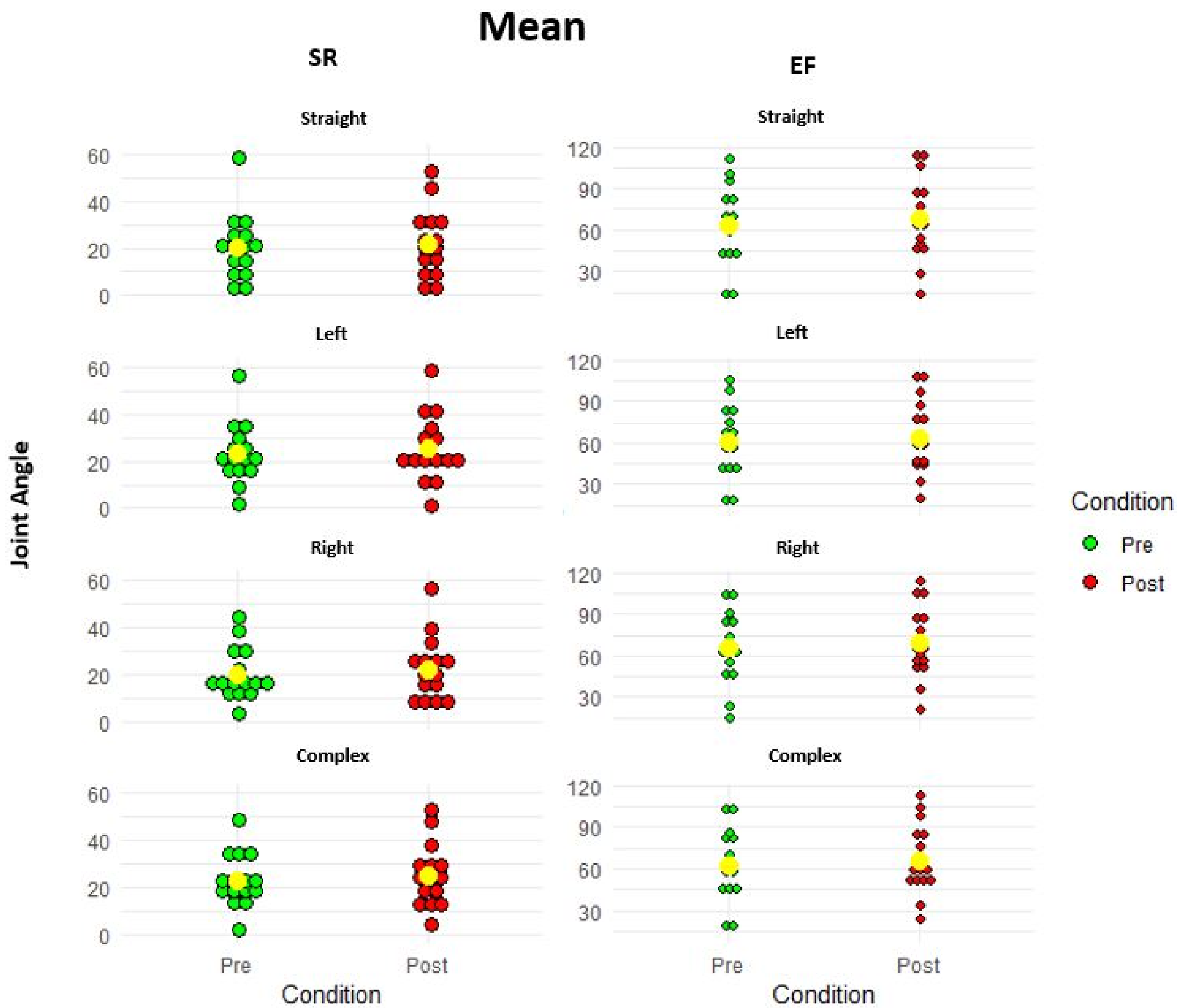


Figure 17: Average joint angle in degree (°) over the whole trajectory for each individual before and after the fatigue at different turns.Yellow dot indicate a mean of average joint angle over 16 subjects.

4.3 Correlation Between Fatigue, Gender, and Kinematics

Table 4: Linear mixed model results. Two variables, fatigue and gender, were evaluated whether they are significant predictor of joint angle in three measurements (Mean,Maximum, and ROM). Values shown in the table are p values and if they were less than 0.05, they were considered significantly different. Significant difference was indicated by *

DOF	Measurements	Segments							
		1 (straight)		3 (Left turn)		4 (Right turn)		6 (Complex turns)	
		Fatigue	Gender	Fatigue	Gender	Fatigue	Gender	Fatigue	Gender
SP	Mean	0.312	0.612	0.251	0.237	0.306	0.071	0.516	0.971
	Max	0.499	0.725	0.254	0.366	0.829	0.644	0.704	0.861
	ROM	0.058	0.578	0.254	0.643	0.068	0.876	0.096	0.847
SE	Mean	0.363	0.610	0.054	0.687	0.692	0.815	0.666	0.981
	Max	0.308	0.190	0.988	0.048*	0.199	0.138	0.007*	0.121
	ROM	0.426	0.571	0.845	0.011*	0.139	0.058	0.037*	0.048*
SR	Mean	0.021*	0.551	0.306	0.297	0.534	0.508	0.461	0.744
	Max	0.160	0.988	0.257	0.898	0.620	0.227	0.026*	0.213
	ROM	0.375	0.935	0.753	0.630	0.330	0.182	0.022*	0.316
EF	Mean	0.180	0.612	0.016*	0.237	0.009*	0.071	0.756	0.971
	Max	0.012*	0.725	0.001*	0.366	0.541	0.644	0.034*	0.861
	ROM	0.021*	0.578	0.001*	0.643	0.088	0.876	0.017*	0.847

The linear mixed model results indicate that fatigue is generally a significant factor in predicting joint angle in elbow flexion ($p \leq 0.05$) for all turns as shown in Table 4. This is contrary to results from the paired t-test as they show the most significant difference in shoulder plane. This may be due to some residuals of the data set deviating from a normal distribution. As we were only examining DOF in the shoulder, it is not clear whether this significant change in elbow flexion comes from flexion of the elbow or tilting of the torso causing elbow flexion, which was observed in some video recordings. Future studies should consider measuring tilt of the thorax with respect to the lower body.

The linear mixed model also suggested 56.7% of joint angle variation can be predicted from subject variation, meaning variation of the subject may be the strong indicator of the variation of the joint angle. The joint angle in the shoulder plane was similar between pre-fatigue and post-fatigue in all three measures. Neither gender nor the presence of fatigue were influential factors in determining the joint angle ($p \leq 0.05$). The shoulder rotation and elbow flexion seem to be subject dependent, meaning the largest variation comes between subjects not from fatigue conditions or maneuver type.

CHAPTER 5

DISCUSSION

5.1 Comparing Change in Muscle Activation Before and After Fatigue

An increase of average EMG amplitude in straight, left, and complex turns is consistent with our hypothesis that the biceps become more active to compensate for fatigue of rotator cuff muscles. However, the change was only statistically significant during complex turns. A significant increase in the bicep EMG during complex turns suggests that during rigorous steering maneuvers the bicep may be exerting more work to compensate for muscle fatigue of the infraspinatus and supraspinatus. A significant decrease in maximum EMG amplitude of the deltoid, a muscle known to be highly active in steering, during complex turn also indicates a transfer of muscle usage during rigorous steering. However, further studies that examine other major muscles need to be conducted to identify which muscles make up for a significant decrease in deltoid activity.

Contrary to our hypothesis, in straight and left turn, the deltoid seems to be compensating for the fatigue of rotator cuff muscle, as suggested by a significant increase in its activity. Combined with analysis of complex turns, it seems to suggest that for simple turns, which requires less dynamic movement, the deltoid is more likely to become more active whereas in complex turns the bicep is more active. Consistent with this concept, in the right turn, the deltoid had significant decrease in EMG for all three measurements as the right deltoid is an extensor during right turns.

Rotator cuff muscles maintained high (~50%) maximum activation at all turns, which is expected after fatigue protocol. However, in complex turns, infraspinatus reached high activation levels (95.03%) after fatigue with activation of the bicep increasing significantly, yet the deltoid decreased significantly. This suggests that in complex turns the infraspinatus is still at work while the deltoid seems to reduce its usage. This indicates that the infraspinatus has a higher threshold in maintaining high activation while the deltoid has a lower threshold. This suggests that the infraspinatus may be more vulnerable to tear in demanding steering maneuver compared to the supraspinatus and deltoid. In real-life driving, activation may be higher than the values from this study since real driving wheels exert torque resistance. As this torque resistance is usually higher in turns at a high speed, reported values can be higher in real driving.

Maximum EMG amplitude on the infraspinatus showed activation close to its maximum voluntary contraction amplitude in simple turns. This suggests that the bicep may still be relatively unused despite fatigue of rotator cuff muscles. Although a portion of the stress is distributed to the bicep, the rotator cuff muscles seem to be used more instead of being compensated. This implies more dangerous outcome for steering with rotator cuff muscle fatigue as there is no reliable compensatory mechanics to alleviate muscle weakness in steering.

The results also suggest that an increase in mean activation level is correlated to an increase in the standard deviation and maximum muscle activation level as well. Therefore, fatigue of rotator cuff muscles may lead to more erratic use of muscles in steering. The maximum EMG amplitude were approximately two to four times higher than mean. This large difference between maximum and mean EMG amplitude indicates a large fluctuation in EMG amplitude during steering. This may suggest that fatigue may have led the subjects to perform more erratic driving maneuvers. Such maneuvers may have caused a spike in EMG measurements, leading to significant increase of maximum EMG amplitude for all muscles during the complex turn. Change in data processing such as an increase in cut-off frequency and fine segmenting technique may improve the analysis.

The mean EMG amplitude for the deltoid, infraspinatus, and bicep are comparable to the literature. However, the mean EMG amplitude for supraspinatus in our study was noticeably low (17∼18%) compare to established value (30∼50%), which is more comparable to the maximum value (45∼50%) from our study. This may be due to the literature only looking at sudden steering, whereas this study has a combination of slow steering and sudden steering.

5.2 Comparing Change in Joint Angle Before and After Fatigue

The second aim of this study was to analyze how change in muscle usage after rotator cuff muscle fatigue would manifest in kinematics and its implications for clinical practice regarding RCT. Contrary to our hypothesis, significant change in muscle activation generally did not translate to significant change in the joint angle of the muscles studied muscles as shown in Table 3. This suggests that regardless of reaching fatigue threshold in muscle activation, subjects will perform steering in similar way. However, our driving trials only lasted a minute for each trial and may not be a perfect representation of real-life driving. Changes in kinematics may manifest the longer one drives.

Although mean shoulder elevation, shoulder rotation, and elbow flexion did not significantly change after fatigue, a significant increase occurred in ROM of the joint angle in the shoulder plane for all turns. However, the major muscles that are engaged in the shoulder plane such as pectoralis major and trapezius were not monitored in this study. Thus, fatigue of rotator cuff muscles may have been compensated through either pectoralis major or trapezius.

As mentioned previously, EMG measurements for deep muscles, the infraspinatus and supraspinatus, were filtered using an equation relating surface electrode measurements to indwelling electrode measurements. The equation may only be accurate during a movement for which the deep muscle of interest is a prime mover. Although steering is known to highly activate the infraspinatus and supraspinatus, they may not be a prime mover for the whole duration of steering. Future studies should consider using indwelling electrode for more accurate data acquisition.

CHAPTER 6

CONCLUSION

6.1 Summary of Findings

Although our study concludes that drivers only minimally compensated for fatigue of rotator cuff muscle through the deltoid or bicep, it is still not known whether fatigue is compensated through other prime movers of the elbow such as the triceps or other muscles. Our study also concluded that the two rotator cuff muscles, infraspinatus and supraspinatus, remain highly active even after on-set of fatigue during sudden steering. These suggest that the rotator cuff muscles may be susceptible to tear if one drives for prolonged hours with the presence of fatigue. However, in real life driving, sudden steering maneuvers are much less common, and frequency of steering is likely lower than we tested. The fatigue protocol conducted in this study is not representative of prolonged driving but other physical activity that requires numerous overhead movements. Thus, caution must be taken when driving for a prolonged period if one engages in rigorous overhead exercises prior to driving.

6.2 Future Work

1. The use of indwelling electrodes in acquiring EMG signals of the rotator cuff muslces. In the current study, EMG measurements for deep muscles, the infraspinatus and supraspinatus, were filtered using an equation relating surface electrode measurements to indwelling electrode measurements. The equation may only be accurate during a movement for which the deep muscle of interest is a prime mover. Although steering is known to highly activate the infraspinatus and supraspinatus, they may not be a prime mover for the whole duration of steering. Future studies should consider using indwelling electrode for more accurate data acquisition.

2. Analysis of forces and moments on the steering wheel. Force and moment measurements from a load cell mounted behind the steering wheel were never utilized in this study. Future study should consider analyzing forces and moments on the wheel to

understand how force exertion on the wheel changes with respect to muscle usage and kinematics after fatigue. This will provide better explanation for tilting of the torso that were observed after fatigue and how it plays a role in the compensatory mechanism of the rotator cuff muscle fatigue during steering.

6.3 Limitations of the Study

1. Limited number of EMG sensors. Due to limited number of EMG sensors, only four muscles were monitored in our study. Future study should consider monitoring more muscles to better understand relationship between kinematics and muscle usages during steering. Particularly, the pectoralis major should be monitored since joint angle measurement in shoulder plane showed significant increase in range of motion at all turns. Measuring activation of pectoralis major will provide more insight to the significant change occured in range of motion in shoulder plane.

2. The use of research-grade driving simulator. The custom driving simualtor in this study lacks in many respects. It does not provide any torque resistance on its wheel and requires abrupt turning in certain segments of the driving course. Using a research-grade driving simulator will better imitate real-life driving.

CHAPTER 7

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APPENDIX A

KINEMATICS AND MUSCLE ACTIVATION DATA

A.1 Average Joint Angle Trajectory

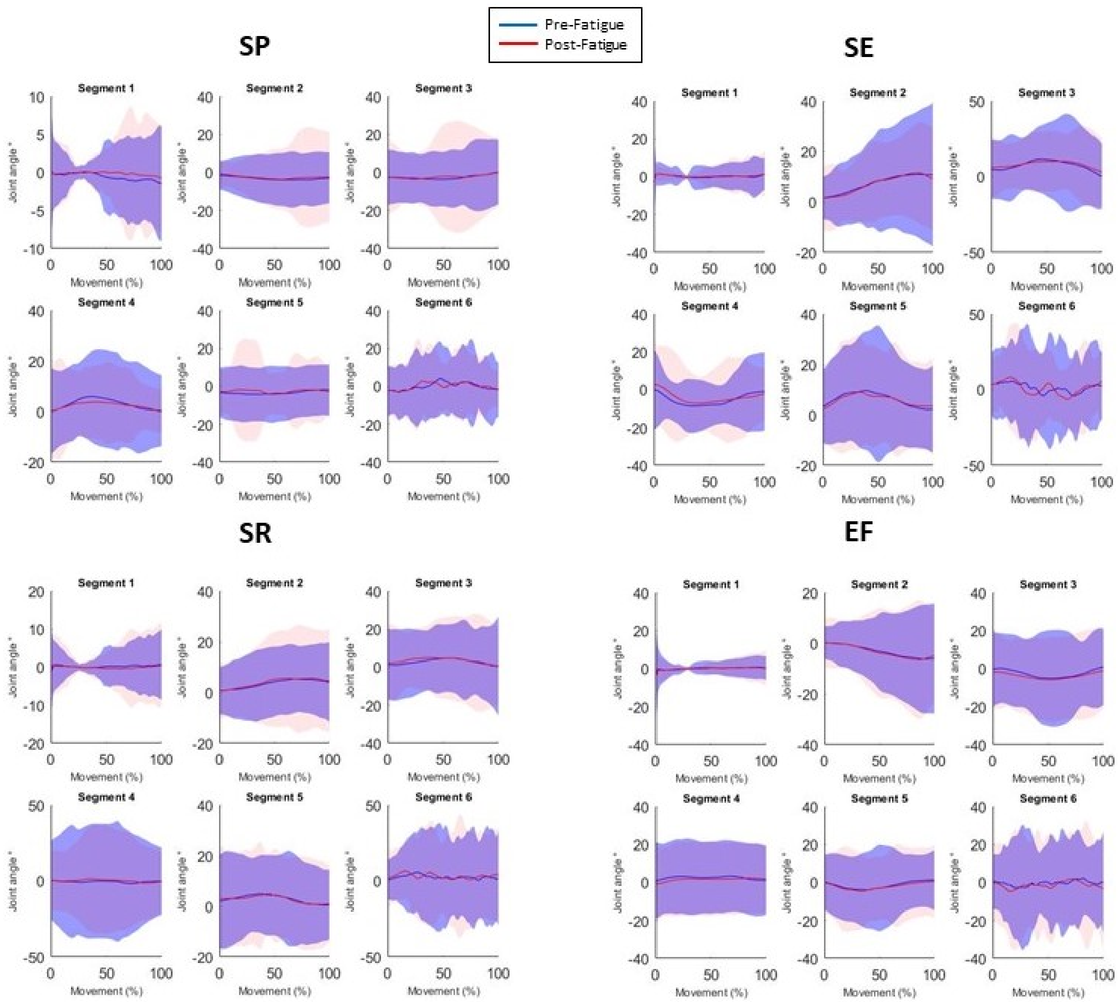


Figure 18: Average joint angle (line) and its standard deviation (shaded bound) for pre-fatigue(blue) and post-fatigue(red).

During the data analysis, joint angle trajectory was plotted to determine patterns in joint angle at each segment. However due to large subject variance in joint angle, identifying a pattern for each turn was difficult. Figure 18 shows that standard deviation is ranges from 20° to 50° depending on degree of freedom. This suggests that steering is subjective upper body activity and varies notably between subjects.

A.2 Individual EMG Trajectory

Figure 19 shows trajectory of EMG amplitude at different segments. Same with joint angle analysis, EMG amplitude vary noticeably between subjects.

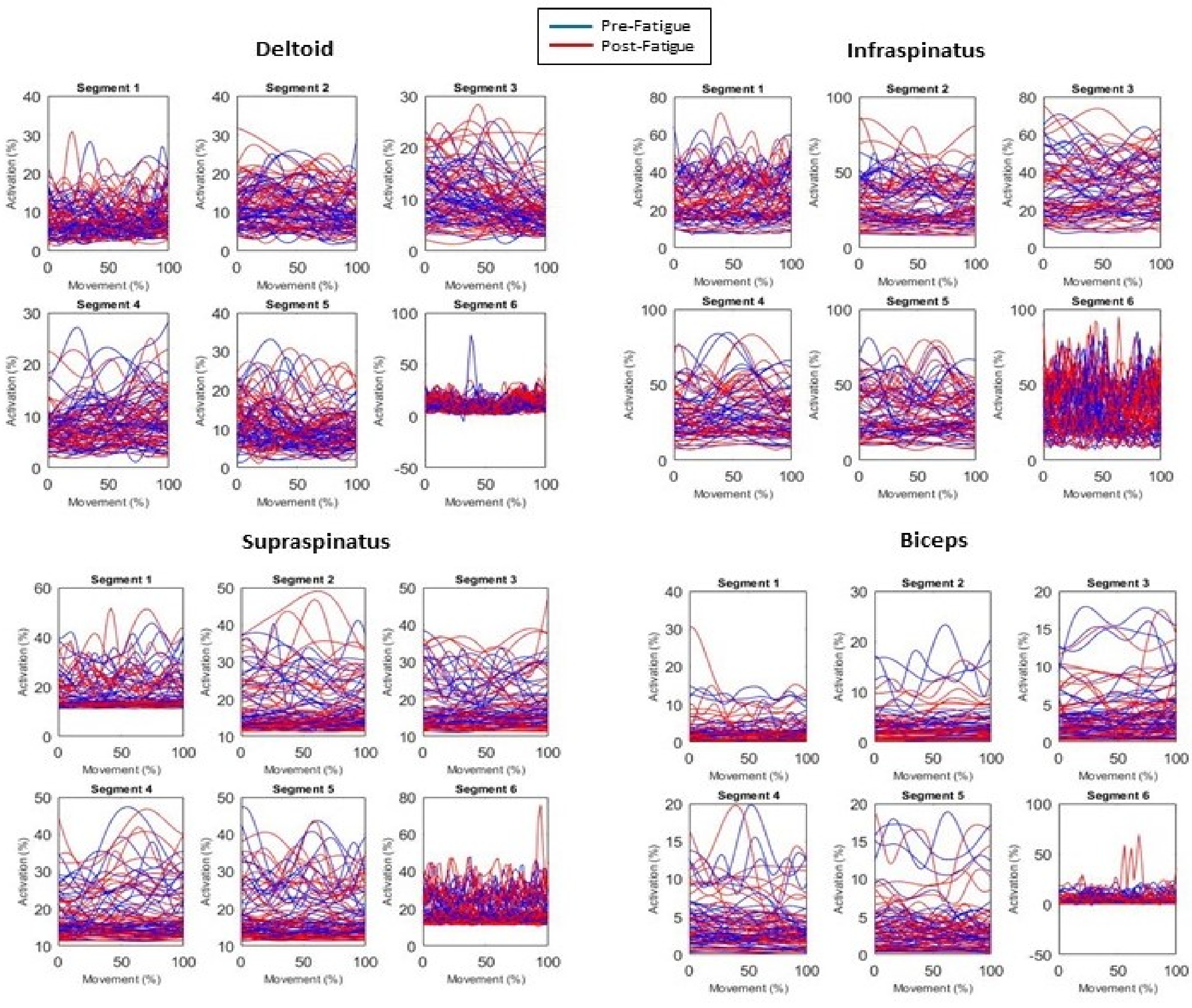


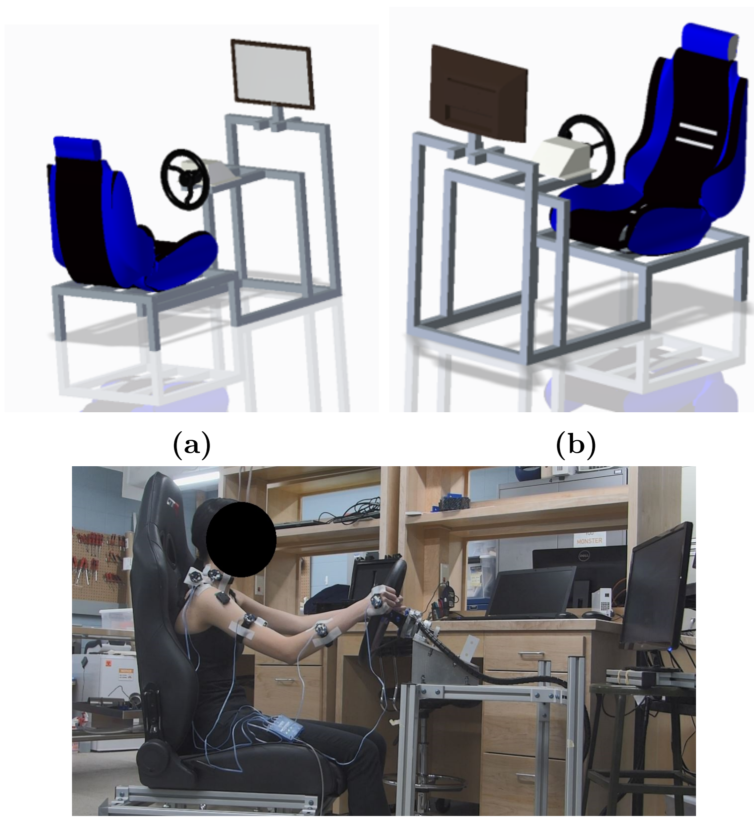
Figure 19: Trajectory of EMG amplitiude in %MVC for all muscles at different segments.

APPENDIX B

EXPERIMENTAL SETUP

B.1 Custom Driving Simulator

Figure 20 shows CAD drawing and photo of custom driving simulator. 40mm X 40mm aluminum column were used to build the structure of the simulator (purchased from 8020.net). Figure 21 shows custom designed steering wheel assembly for force and moment data acquisition during experiment. However, these were never utilized in this study.



(a) **(b)** **(c)**

Figure 20: (a) Side view of custom designed driving simulator. (b)&(c) CAD drawing of the driving simulator.

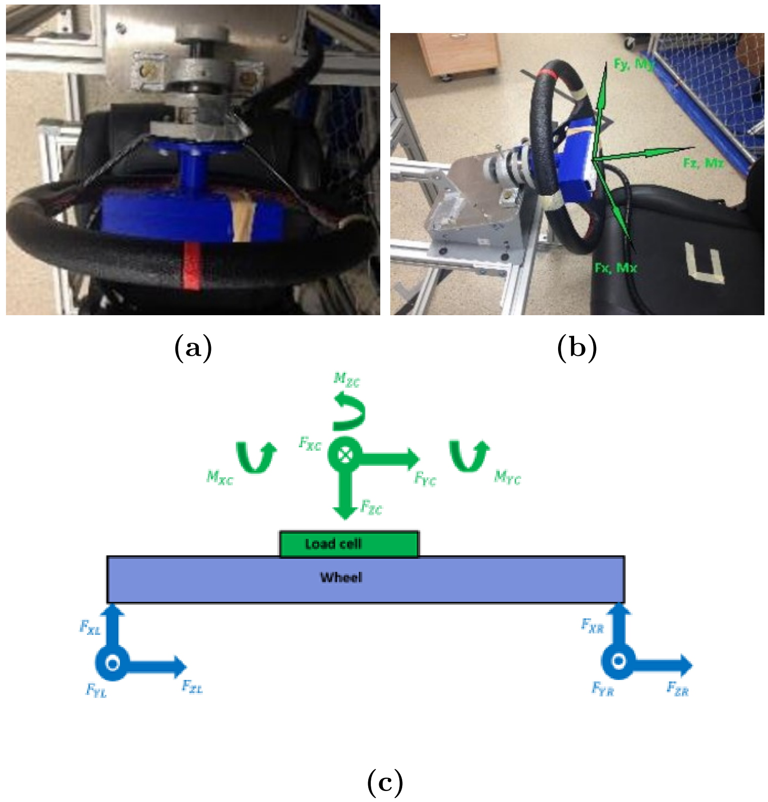


Figure 21: (a)&(b) Top view of the steering wheel assembly. (c) Diagram of the force and moment direction respect to the steering wheel.

B.2 PAR-Q Questionaire

Upon arrival for the test session, the participant will be informed of the experimental procedure and asked to sign a consent form. PAR-Q questionnaire will be asked to participants for risk stratification and the questionnaires are below.

(1) Do you have a history of any heart condition?

(2) Do you feels pain in your chest during physical activity?

(3) Have you felt any chest pain in the past month from non-physical activity?

(4) Have you lost your balance due to dizziness or lost your consciousness.

(5) Do you have a history of bone or joint injury that prohibits physical activities?

(6) Are you taking drugs for blood pressure or heart condition?

(7) Are you under advisement to not perform physical activity?

If participant answers yes for any of these questions, the participant would be excluded from the study