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## Validation of Shoulder Strain Index

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# **Validation of Shoulder Strain Index**

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Thesis submitted  
to the Benjamin M. Statler College of Engineering and Mineral Resources  
at West Virginia University  
in partial fulfillment of the requirement for the degree of

**Master of Science  
in  
Industrial Engineering**

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**Keywords:** Shoulder Biomechanics; Strain Index; Forceful Arm Exertion; SEMG

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# **Abstract**

## **Validation of Shoulder Strain Index**

**Fabiha Islam**

Musculoskeletal disorders (MSDs) of shoulder are one of the most frequent and major causes of morbidity and pain among full-time workers in the USA. In 2018, shoulder disorders were the most severe type of MSDs in terms of lost workdays. Recent epidemiological studies have identified several biomechanical, organizational, and psychological factors that increase the risk of shoulder MSDs. Though multiple biomechanical studies have evaluated physical causal factors of shoulder MSDs with a goal of risk mitigation, literature lacks a standardized risk assessment tool. The main objective of this study was to validate a previously developed shoulder strain index. The strain index is based on the concept of shoulder stabilizing concavity compression mechanism. A two-fold process was used to conduct the validation: (1) Forty different forceful arm exertions characterized by two shoulder planes, two shoulder angles, two force levels and five orthogonal directions of force exertion were modelled using a full body biomechanical modeling software (AnyBody Modeling System™). The magnitude and direction of internal joint reaction forces acting at glenohumeral joint were used to estimate the shoulder strain index. (2) The surface electromyography data recorded from seven shoulder stabilizing muscles were analyzed to estimate the total physiological demand on the shoulder complex. Statistical analysis was performed to understand the effect of forceful arm exertions on the strain index and to quantify its relationship with the shoulder physiological demand. The strain index was significantly affected by work-related factors such as shoulder plane, shoulder angle, force exertion level and direction. The correlation coefficients of 0.7 ( $p < 0.001$ ) was observed between total physiological demand and the strain index indicating that shoulder strain index tested in this study is a good estimator of physiological workload and can serve as an assessment tool to predict risk of shoulder MSD due to forceful manual material handling tasks.

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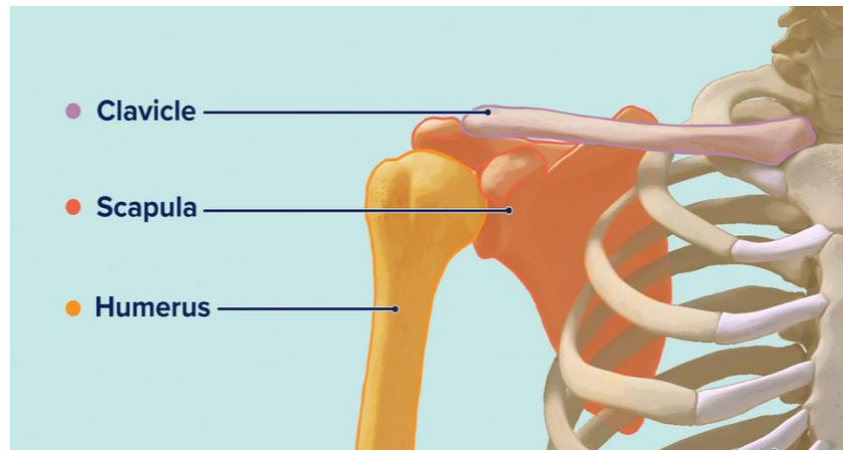
## **Chapter 1: Introduction**

Musculoskeletal disorders (MSDs) are reported as the most frequent and significant work-related injuries among full-time workers in different occupational sectors. The Bureau of Labor Statistics indicates that in 2018, MSDs accounted for 50% of all illness cases in industries and 40% of all illness cases in occupational sectors. Particularly, shoulder disorders experienced by the workers are the most common MSDs in occupational settings. According to the Bureau of Labor Statistics (BLS), shoulder disorders were the most severe type of MSDs in terms of lost workdays in 2015, which resulted in 23 median lost workdays whereas all other MSDs resulted in 9 median lost workdays (Chowdhury et al., 2018). Moreover, shoulder MSDs bring about huge direct and medical expenses. The compensation allocation of the direct cost for work-related shoulder disorders from 1997 to 2005 was \$16,092 per claim in the state of Washington. Additionally, it has been observed that indirect costs such as reduced health, impaired task ability, decreased productivity, and low quality is ideally higher in the industrial sectors compared to other occupational and service sectors (Alasim et al., 2019).

Recent studies have found that workers of certain industries such as retail trade, manufacturing, healthcare and social assistance, transportation and warehousing, construction, agriculture, forestry, fishing and hunting suffer the most from shoulder MSDs. However, operators in other occupational and service sectors also suffer from shoulder fatigue as they are exposed to prolonged static or repetitive loads for a major part of the working day. These occupations include freight, stock and material movement, nursing and health care assistance, heavy tractor-trailer truck and delivery service driving, maintenance and repair work (Bureau of Labor Statistics, 2018). In general, the prevalence rate of MSDs is higher among male workers (37.5 per 10,000) compared to female workers (29.7 per 10,000) (Charles et al., 2018). Suffering from shoulder discomfort and loss of function are common complications among the patients (van der Windt et al., 1995). Another primary shoulder complication is rotator cuff disorder, which is found in people who perform repetitive exertion tasks. Moreover, the anatomical conflict between the intrinsic and extrinsic factors results in partial or complete tears of the rotator cuff tendons, frozen shoulder, subacromial bursitis, joint impingement syndrome, and pain in the glenohumeral joint (Bramanet al., 2014; Cudlipet al., 2018).



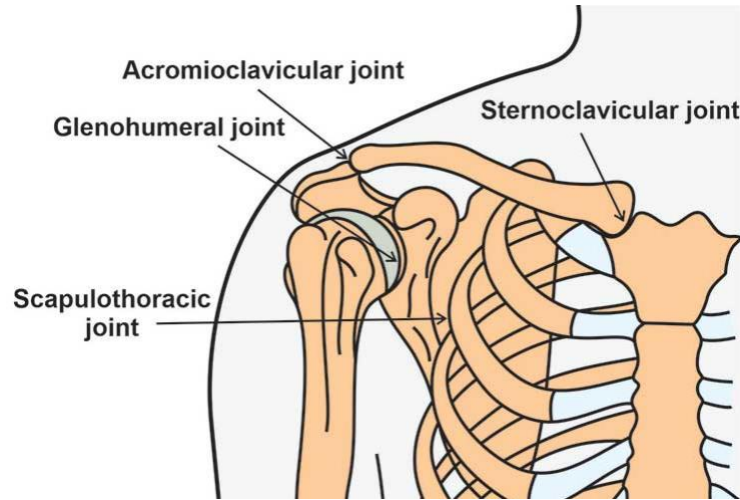
Multiple studies have identified several biomechanical, organizational, and psychological factors that increase the risk of shoulder MSDs at the workplace. Biomechanical hazards such as forceful arm exertions, awkward postures, repetitive motions of upper extremities, hand, arm, and full-body vibration, unbalanced workstations as well as insufficient recovery time are significant risk factors of subacromial diseases (Bodin et al., 2020; Chowdhury et al., 2018). Additionally, there is ample evidence that mechanical impingement and biomechanical overloading also lead to the development of shoulder disorders among industrial workers (Braman et al., 2014; Alasim et al., 2019). In addition to the biomechanical influences, epidemiological investigations have documented different organizational factors that are associated with shoulder disabilities. These exposures include but are not limited to an uncomfortable sitting posture, limited rest or breaks, and unavailability of adjustable office desks. Furthermore, several psychosocial conditions of the work environment such as high job demands but lower levels of control, poor support from supervisors and colleagues disrupt the muscle activation, leads to a higher level of muscle stress resulting in muscle fatigue (Jiskani et al., 2020; Mehrdad et al., 2010; Roquelaure, 2018)



**Figure 1.1: Shoulder bones: clavicle, scapula, and humerus<sup>i</sup>**

The human shoulder is a complex combination of numerous muscles, ligaments, and tendons that enable flexible movements of the shoulder across several degrees of freedom and multiple planes. The shoulder complex is comprised of three bones: clavicle, scapula, and humerus (Figure 1.1). The clavicle, also known as the collarbone, extends from the shoulder blade to the

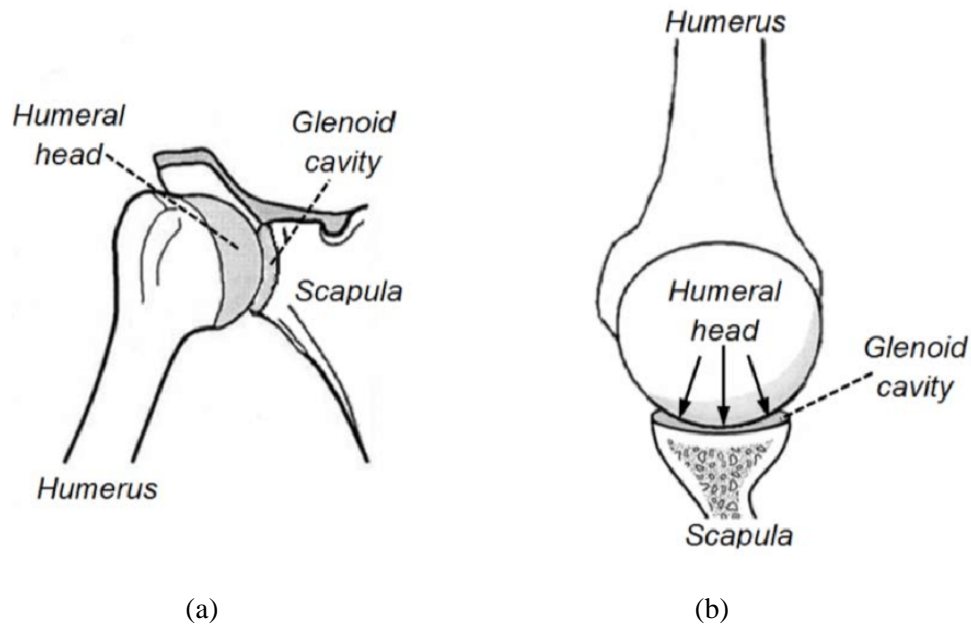
sternum and meets the acromion in the acromioclavicular joint. The scapula is a flat triangular-shaped bone that is located at the poster lateral side of the thoracic cage and links the humerus with the collar bone. The humerus is a long bone that connects the scapula with the radius and ulna. In order to translate an extensive range of motion, these three bones meet at an angle of 90° and create four basic shoulder joints.



**Figure 1.2: Shoulder joints: glenohumeral joint, scapulothoracic joint, acromioclavicular joint, and sternoclavicular joint<sup>ii</sup>**

The fundamental shoulder joints are the glenohumeral joint, scapulothoracic joint, acromioclavicular joint, and sternoclavicular joint (Nicolozakes, 2021) (Figure 1.2). The glenohumeral joint is the main joint of the shoulder which is a ball and socket type joint made by the humerus head and scapula (Figure 1.3a). As the humerus head is larger than the glenoid cavity, it fits slightly free in its movement, allowing extension, flexion, abduction, and rotation at the cost of underlying instability. The shoulder joint instability causes the humeral head to translate away from the glenohumeral cavity and compresses the soft tissues which result in subacromial impingement (Mulla et al., 2020). In order to support the humeral movement and make the shoulder stable, the joint is surrounded by a group of muscles and tendons, known as the rotator cuff. Supraspinatus, infraspinatus, teres minor, and subscapular are four muscles of the rotator cuff originating from the scapula. Anterior deltoid, lateral deltoid, and posterior deltoid are three muscles that facilitate glenohumeral elevation and rotation. When an external force is applied to

the shoulder, these muscles produce a compressive force in order to counteract that external force. For stabilizing the shoulder joint, this compressive force must be larger than the individual translational forces. The mechanism through which this coordinated action is maintained is called the concavity compression mechanism (Lazarus et al., 1996; Lippitt et al., 1993) (Figure 1.3b).



**Figure 1.3: (a) Ball and socket joint of glenohumeral joint (b) concavity compression Mechanism<sup>iii</sup>**

Numerous studies have proposed several clinical and research approaches to detect the sources of shoulder disorder, quantify the amount of discomfort and reduce the risk of exposure. These include the use of self-reported checklists, Shoulder Pain and Instability Index, Quick DASH, ultrasound imaging, and multiple readily available devices (Curtis et al., 1995; da Silva et al., 2018; Minoughan et al., 2018; Sudarshan et al., 2019; O’Sullivan et al., 2012). In a recent study, Chowdhury et al., (2018) proposed the use of the shoulder strain index which is developed based on the functionality of the concavity compression process. This is relatively new tool and sufficient data is not available regarding its reliability and validity. In order to prevent the work related MSDs (WMSDs) of shoulder, it is critically important that accurate and fully validated workplace assessment tools are available to the employers. Toward this goal, in this study, the

strain index developed by Chowdhury et al., (2016) will be further examined and evaluated to test its efficacy in predicting risk of shoulder WMSDs during work-related forceful arm exertions.

## **Chapter 2: Literature Review**

Previous studies investigating the relationship between musculoskeletal strain and external force, the direction of arm exertion, and posture have utilized several different methodologies. These methodologies can be broadly categorized into three main groups: observational studies, physiological studies, and biomechanical modeling. Each group has its own methodology with each having its own set of advantages and drawbacks. Below, a brief review of these studies is presented.

### **2.1 Observational Studies**

The goal of an observational study is to analyze the consequence of a treatment that is not randomized and affected by any designed procedure. In general, during the experiments, either the subjects select their own way of conducting the experiment, or their process, as well as output, is affected by the environment (Rosenbaum 2005). Several studies have attempted to develop observational methods to measure the intensity, frequency, and duration of exposure in a systematic and comprehensive manner (Andreas & Johanssons, 2018). In recent years, Health and Safety executives have suggested a number of survey techniques, self-reported questionnaires, video recordings, and other observational aids to acknowledge the etiology, chronicity, and severity of shoulder disorder (Linaker & Walker-Bone, 2015). The most commonly used tools are Rapid Upper Body Assessment tool (RULA), Rapid Entire Body Assessment (REBA), Quick Exposure Check, The American Conference of Governmental Industrial Hygienists Threshold Limit Value (ACGIH TLV), Video Exposure Monitoring (VEM), Nordic Musculoskeletal Questionnaire (NMQ) (Beese et al., 2021; Chu et al., 2021; Hignett & McAtamney, 2000; Nadri et al., 2013; Thomas et al., 2021). One important aspect of observational studies is that they provide insightful information that is often concealed in randomized experiments. The advantages of observational studies are that they are simple and consider a wide range of socio and psychological factors (Suissa & Garbe, 2017).

Questionnaires and checklists allow researchers to understand the personal factors, organizational orientation, psychological and biomechanical stresses among a larger group of individuals. In one such study conducted by Thomas et al., (2021), the investigators compared the

general risk assessment tool used by the local health and safety executives with Body Mapping and MSD ill-health absence rate during domestic waste collection. The participants reported that they experienced the highest fatigue in moving 100l sacks containing garden waste and recycling boxes while the lowest level of pain was associated with serving the 240l wheeled bins. In order to reduce the MSDs among the workers, they suggested redesigning the waste collection process and proposed to use a body mapping strategy for the risk assessment (Thomas et al., 2021). Chu et al., (2021) used Nordic Musculoskeletal Checklist incorporated with a multivariable logistic regression model to find out the personal as well as work-related risk factors associated with shoulder disorder. They hired 931 electronics workers to gather necessary information about movement frequency, posture, load, and time of exposure. One of their major findings was that multiple components including age, repetitive exertions, working posture, and external loads had the most significant influence on developing shoulder symptoms among male workers. In contrast, shoulder fatigue among female workers was influenced mostly by the posture at the workplace (Chu et al., 2021). Kee (2021) compared Ovako Working Posture Analysis System (OWAS) against Rapid Upper Limb Assessment (RULA) and Rapid Entire Body Assessment (REBA) using 209 medical samples associated with MSDs. The author found that RULA and REBA had better action levels than OWAS and based on the grand score, RULA serves the best for estimating the postural load (Kee, 2021).

Apart from the observational surveys, several video monitoring assessment techniques have been employed for investigating the exposure and the working environment in a more objective manner. These video-based assessment tools combined with motion tracking devices and other sensors can provide a more accurate and documented picture of the category of workload, exposure angles, duration, and frequency at each posture. Moreover, they provide more critical and corrective measures compared to traditional physical load computation techniques (Heberger et al., 2012; Hernandez et al., 2019; McKinnon et al., 2019; Pehkonen et al., 2009). In a current study, Polat et al., (2016) combined video recordings with Ovako Working Posture Analysis System (OWAS) to analyze the physical workload and risk factors during loading and unloading the machinery in a medium-sized wood processing factory. From this combined assessment the authors found that five machines among twelve had the most hazardous effect on the musculoskeletal system of the workers and suggested ergonomic improvements of the operational settings (Polat et al., 2016). Forsman et al., (2006) proposed an integrated method of VIDAR and

Quick Exposure Checklist, where operators were asked to identify the physically demanding situations, the amount and zone of fatigue, and the external load weight from pre-recorded video clips. Once the hazardous situations were identified by the operators, they were printed out and the checklist was implemented to assess the workload in the back, shoulder, wrist, and neck region (Forsman et al., 2006). Syuaib (2015) applied video recording system to analyze the work motion of 141 harvesting workers near the shoulder and arm area. The results showed that the current working posture are not safe for the harvesters and they require immediate ergonomic interventions (Syuaib, 2015).

Observational experiments are simple and economical tools that have served as a great method of measuring exposure in the work environment and suggesting improvements considering musculoskeletal disorders (Chu et al., 2021). However, several occupational studies have addressed some major drawbacks of the application of these tools. Sometimes, the self-assessment questionnaires which are used to gather the participant's physiological information are not directed by specialists. Additionally, because of their immense subjectivity and substantial chances of missing important information, these observations may generate inappropriate results (Beese et al., 2021). Another major fact is that although these surveys and checklists are very effective for running up among larger groups, they are often inconsistent with the recommended guidelines (Smythe et al., 2021). In addition, it is tedious for human observers to process a huge amount of subjective information and use it for a holistic judgment and pain management (Pinto et al., 2011; Posada-Quintero et al., 2021).

## **2.2 Physiological Study**

Physiological studies can be defined as a controlled and observational study of integrated functional changes of the human body (Lim et al., 2013). Electromyography (EMG), Electrocardiograms (EKGs), Electroencephalogram (EEG), Heart Rate (HR), Event-Related Brain Potentials (ERPs), and Blood Pressure (BP) are some of the diverse measuring tools available that facilitates the researcher to better understand the body's physiology. The purposes of recent psychological studies while workload assessment is to evaluate the physiological loads, effects of certain tasks on the strength of muscle, and examine if the load falls within the allowable limit or

not. Physiological studies have been widely used because of their certain advantages over other traditional workload evaluation processes. The advantages of these studies are they provide more specific and unambiguous information indicating the problematic design and they are vastly reliable tools for dynamic risk assessment (Kramer 2020; Tran et al., 2007).

The combination of neurophysiological patterns such as EEG frequency and EMG signals can be used as good indicators of musculoskeletal pain in the shoulder complex for repetitive exertions. Segning et al., (2021) used EEG signals to estimate the cortical activity by analyzing the beta EEG frequency band ( $\beta$ .EEG) of the task-related power increase/decrease. For this experiment, they recruited ten healthy male and two female participants for performing tasks with high and low risk of muscle pain development for 30 minutes. EEG signals were collected by placing EMG electrodes on the anterior and deltoid muscle, the Power Spectral Density (PSD) and Coefficient of Variation (CoV) were used to quantify the pain scores. One of their major findings was that a  $\beta \geq 50\%$  and  $\text{CoV} \leq 18\%$  provides a reliable index for assessing shoulder pain (Segning et al., 2021). In another study conducted by Grauhan et al., (2021), a convolutional neural network was trained using 2700 shoulder radiographs, and the model was used as an assistive device for identifying the most common causes of shoulder pain. The model found the factors that extensively explained the fatigue such as the proximal humeral fractures, joint dislocation, periarticular calcification, osteoarthritis, osteosynthesis, and joint endoprosthesis. Posada-Quintero et al., (2021) applied machine learning approach to electrodermal activity (EDA) in order to estimate the pain agitation in twenty-three subjects during the exposure to thermal grills. The regression model achieved the  $R^2$  values of 0.357 and 0.47 for stimulation intensity and pain sensation respectively. The authors concluded that the EDA are closely correlated to stimuli's intensity and pain sensation reported by the subject (Quintero et al., 2021).

The surface EMG signals are one of the quickest and noninvasive ways of exhibiting human motions executions. Their use has been recognized as an important control signal as it can produce very reliable and promising results and are being widely used by many researchers in recent history (Liu et al., 2019). Goubault et al., (2021) conducted an experiment where EMG data from ten shoulder muscles of thirty healthy men were collected during the loading and unloading of the pallet in a grocery warehouse for 45 minutes. After processing the data, it was found that among all the EMG indicators, spectral entropy, median frequency, and mobility are the best to assess the modified Rate of Perceived Exertion (RPE). Additionally, they suggested that the



combination of these three indicators can improve the assessment result of the myoelectric manifestation of fatigue. Nimbarde et al., (2010) collected EMG signals from sternocleidomastoid and upper trapezius muscles during lifting tasks at neutral, maximally flexed, and maximally extended neck postures to identify the risk factors among construction workers. The EMG signals indicated that the upper trapezius muscle and the sternocleidomastoid muscle were most active at the flexed neck and extended neck posture respectively. Their result suggested that these tasks contribute significantly to neck disorders among construction workers. Wang et al., (2000) evaluated jaw muscle pain by collecting EMG signals from the masseter and anterior temporalis muscles of twelve participants. Jiang et al., (2020) employed surface EMG for controlling the motion of rehabilitative robotic assistive devices and testing the validity machine learning algorithm used to control the shoulder motion pattern.

The physiological tools are very informative while studying the activation patterns of different muscles, which in turn can help to identify the risk factors and estimating the musculoskeletal loading. However, they have their own set of constraints and limitations. Though the physiological signals represent the neuromuscular components of fatigue; they vary from subject to subject and cannot be used alone for an integrated fatigue assessment (Goubault et al., 2021). To produce more accurate results, they should be validated using other assessment tools. Additionally, the placement of the markers is very challenging as the muscle fibers are surrounded by layers of muscles, fats, and tissues. These biological structures can affect the precise placement of the markers, modify the impedance of the electrodes thus leave a remarkable effect on the recorded signals (Artemiadis, 2012).

### **2.3 Biomechanical Modeling**

Biomechanics can be defined as the study of the movement mechanism of living things by analyzing the methods of mechanics, motion, and forces (Hatze, 1974). It is the detailed study of biological structures which help physical and occupational therapists to understand the functions of the musculoskeletal system and joints. Forces acting on living things can produce motion, act as a stimulus for growth and development, or overload tissues, which in turn causes serious neurological injuries. Biomechanics provides conceptual and mathematical tools and models that

help health and safety professionals to evaluate the forces, motions, stress, and strain working on the subjects during biomechanical loading (Lu & Chang, 2012). Hence, miscellaneous biomechanical models have been developed and employed to prevent occupational injuries and provide a safer working environment by illustrating the capacity of the musculoskeletal system. One important aspect of these models is that they consider the force transfer mechanisms and amount of internal joint stability while evaluating the external and internal loadings. The main advantages of biomechanical models are that they can simulate motions that need not to be physically performed in the laboratory or industrial settings and provide a better computational structure of the intricate interaction among the soft tissues (Högfors et al., 1991; Zheng et al., 2020).

There exist several open and closed-loop biomechanical shoulder models which are used by experts where, rigid links are connected through joints that allow movements in three-dimensional planes (Lenarčič & Klopčar, 2006; Tondu, 2005; Yang et al., 2003). These models consider the coupling of the shoulder joints during abduction-adduction, flexion-extension, and rotation and apply various shoulder girdle mechanisms, techniques, and coordination systems to analyze the shoulder complex. Though these models are good estimators of forces acting on the segments, they do not consider the muscles and sometimes they result in mathematical singularities (Yang et al., 2010). The actual position and movement of bones vary significantly from the conceptual models (Högfors et al., 1991) and these models do not consider the effects of muscle fibers. On the other hand, musculoskeletal models provide more accurate results by considering the internal structural loadings and muscle activation and thus can be used for accurately analyzing the motion.

The conceptualization of morphological computation has inspired the development of musculoskeletal models which are able to simulate a wide range of human movements at a convenient price. The applications of these models have served as a key component for imitating infrequent and unusual performances that are very difficult to regulate in practical situations. One such example is found in the study conducted by Do et al., (2019) where, they used OpenSim to simulate dislocated acromioclavicular joints and evaluate the movement of upper limbs. Using data of ten men they verified that the model can be used for abnormal movement and motion analysis. Mansouri & Reinbolt (2012) proposed a new interface by integrating OpenSim and

Simulink to offer their combined strengths, numerical and dynamic simulations and to use by neuroscientists, physiologists, biomechanists, and physical therapists. Odle et al., (2011) used a shoulder model in OpenSim to understand the relationship between impulsion and shoulder injury among manual wheelchair users. One of their findings was that the shoulder model can be used as a reliable and accurate means for investigating dynamic movements. Panariello et al., (2019) used OpenSim for replicating overhead leveraging and drilling tasks performed at industrial settings. They captured the kinematic data using optical motion capture system, evaluated the joint angles through the kinematic analysis and concluded that OpenSim can be used to evaluate the overhead tasks appropriately.

Another commercially available modeling system is 3DSSPP, which allows users to predict the static strength data for manual material handling tasks through the three-dimensional human model form. This software allows its users to identify and quantify the risk factors quickly without having to do any manual calculations. Feyen et al. (2000) used the combined interface of AutoCAD and 3DSSPP to evaluate the design of a workplace. In this study, the authors evaluated biomechanical risk factors of a manual material handling task and computed the reach distance, clearance, and visual requirements of a workplace. The outcome was compared with a previously established result and both indicated that the biomechanical stress experienced by the workers exceed the NIOSH limit. Thus, the authors proposed that this combination is a valid tool for assessing workplace designs. Merryweather et al. (2009) used 3DSSPP to revise and reduce the assumption error of a basic hand-calculation back compressive force (HCBCF) estimation model. For this purpose, the back compressive force of L5/S1 intervertebral disk, torso flexion angle, and upper body mass center location was calculated in 3DSSPP with 6000 observational data for male and female subjects. The result showed that though the HCBCF model exaggerated the back compressive force for L5/S1 disk while comparing with 3DSSPP, the predictive capability and accuracy of the model were relatively high after revision. However, the HCBCF models do not allow asymmetric movement during material handling and do not consider the body mass distribution. Thus, it was concluded that the HCBCF model is a simple alternative to a computer-based model and it can be used to sort out the data for analysis in a more complex model. Matebu et al. (2014) calculated the forces exerted by the workers and created the 3D posture using the anthropometry data of both male and female workers who had been working in the industry for over 20 years. The researchers compared the resulted postures used by the workers with NIOSH

standard and found that the current posture is unacceptable. Therefore, they proposed and applied some modifications to the workplace, analyzed the data in 3DSSPP and the result exhibited that the posture has been improved and the force required to pull or push has reduced to 194.23 N from 931.77N. Asadi et al. (2015) conducted a study on 120 workers from 7 industrial settings in the city of Shiraz in order to determine the prevalence of musculoskeletal disorders and forces exerted on the lower back of the workers during manual load lifting. The reaction forces were determined using 3DSSPP and were found that in 17.5% of the compressive forces, and 10.8% cases of shearing forces exerted on the L5-S1 disc exceeded the permissible levels.

Virtual prototyping affiliated with multibody dynamics, mathematics, artificial intelligence, physiology, and software engineering is extensively being used for evaluating the biomechanical loading and calculating the strain placed on the musculoskeletal systems (Rasmussen et al., 2003). AnyBody Technology is such a software that is capable of analyzing the biomechanics of the human body as well as rigid structures using forward and inverse kinematics (Damsgaard et al., 2006). Several previous studies have confirmed the use of AnyBody for investigated biomechanical loading. For instance, Bassani et al., (2017) focused on validating the usability of the AnyBody full-body musculoskeletal model while calculating the loads placed on the lumbar spine at the L4L5 level. For this purpose, 12 specific exertion tasks were replicated into AnyBody by importing the motion capture data and the intradiscal pressure around the L4L5 region was calculated in vivo. While comparing the lumbar spine loads generated by AnyBody with the pressure in vivo, the authors observed a positive agreement between them, thus exhibiting the suitability of the AnyBody modeling system. Additionally, Ji et al (2016) used surface EMG signals to check the rationality of AnyBody while computing the stress of upper extremity during bench press tasks. The statistics of EMG data collected from bow deltoids, right bicep brachii, right triceps brachii, right flexor carpi ulnaris, and trunk pectoralis muscle highly validated the results obtained from AnyBody.

Recently, the use of several biomechanical software has gained popularity for successfully modeling and simulating diverse musculoskeletal movements including sitting, walking, cycling, running, and forceful exertions (Hamner et al., 2010). Though this software represents very advanced and dynamic methods for solving biomechanical problems, some limitations are needed to be noted. Sometimes, robust design and control components are unavailable in them, which

complicates the accurate representation of muscle (Mansouri & Reinbolt, 2012). Furthermore, some of these models have been build based on a time zero case scenario, as a result, the effect of muscles healing from fatigue and relaxation of tissues over time is not considered. Consequently, small changes in tissue tractions over time may affect the results (Seth et al., 2016).

## **2.4 Other Studies**

In addition to the studies mentioned above, other studies have investigated several other ways of identifying the major causes of shoulder disorders. A study by Alavi et al., (2016) comprising 1488 Iranian office workers identified uncomfortable sitting posture, limited rest breaks, and unavailability of adjustable office desks as the predictors of shoulder MSDs. Antony, et al., (2010) conducted research on 16 participants who performed isometric and dynamic shoulder exertions in flexion, mid-abduction and abduction positions at the angles of 30°, 60°, 90° and 120° with no load, 0.5 kg load, and 30% grip condition. As gripping forces have consequential effects on muscle fiber activation and internal joint loadings, they considered that hand gripping might play a significant aspect in shoulder damage. It was observed that when the shoulder was at an angle of greater than 90°, the contracted muscle and moment arm diminished the moment generating potential which requires greater muscle activity. Palmerud et al., (2000) examined intramuscular pressures of the infraspinatus and supraspinatus with inflated flexion and abduction and found that blood flow rate above 40 mmHg has vulnerable effects on muscle which might result in muscle injury.

## **Chapter 3: Study Rationale**

### **3.1 Problem Statement**

In the United States, MSDs are the most common category of occupational hazard, and causing a large amount of cost in terms of lost workdays and other direct and indirect costs. Among all the work-related injuries, upper body injuries especially shoulder disorders are the third most common complaint that requires musculoskeletal-related primary care consultation. Therefore, it is very important to have shoulder risk assessment tools. Although there are several ergonomics workplace assessment tools available, only one tool – shoulder strain index by Chowdhury et al. (2016) – is a dedicated shoulder assessment tool. This is a relatively new tool and sufficient data is not available regarding its reliability and validity. In order to prevent the WMSDs of shoulder, it is critically important that accurate and fully validated workplace assessment tools are available to the employers. Toward this goal, in this study, the strain index developed by Chowdhury et al. (2016) were further examined and evaluated to test its efficacy in predicting risk of shoulder WMSDs during work-related forceful arm exertions.

### **3.2 Objective and Hypothesis**

In this study, physiological muscle activity data were used to validate the biomechanical model-based strain index. The muscle activity data from a previous study were used. The data were collected from seven shoulder muscles during forceful arm exertion performed at various postures involving different direction of force exertion, anatomical planes, and exertion heights. The postures were simulated in AnyBody Modeling System to compute the reactions forces at the glenohumeral joint. The reaction force data were used to estimate the strain index. The muscle activity data were compared with the strain index data using correlation analysis. It was hypothesized that the strain index data would be highly correlated with the physiological data.

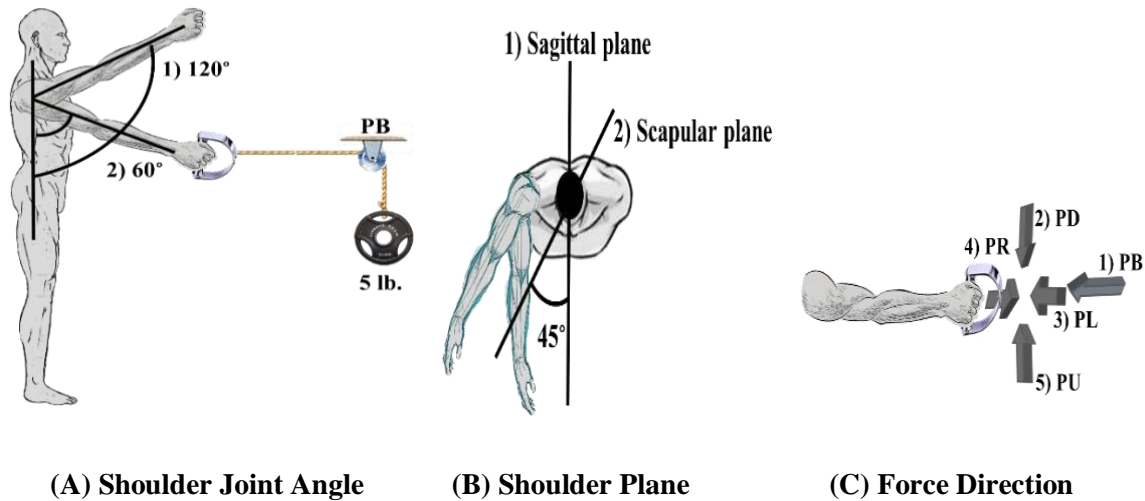
## **Chapter 4: Methodology**

### **4.1 Approach**

A total of forty forceful arm exertions were simulated in AnyBody Technology biomechanical modelling software. The exertions were performed in five orthogonal directions, at two shoulder planes, and at two shoulder angles using two different external force levels. After successfully loading and running the model, glenohumeral joint reaction forces in medial-lateral, inferior-superior, and anterior-posterior directions were obtained. For each exertion, the strain index, which indicates the strain experienced by the shoulder, was calculated based on a previously developed model (Chowdhury et al., 2018). The real physiological strain experienced by the shoulder muscle during the exertions was computed using the electromyography (EMG) data. The EMG data were processed to estimate the root mean square values for each exertion.

### **4.2 Shoulder Muscles EMG Data**

The shoulder muscle EMG data were obtained from a recently completed study in the Industrial Ergonomics lab in WVU. The data were recorded from a sample total of 12 right-hand dominant participants, of which 4 were female participants and the remaining 8 were male participants. The average height, weight, and age of the participants were  $169.33 \pm 9.28$  cm,  $69.92 \pm 13.94$  kg, and  $28.5 \pm 3.71$  years, respectively. Each participant performed a total of 60 forceful arm exertions ( $2$  shoulder joint angles  $\times$   $2$  shoulder planes  $\times$   $3$  force levels  $\times$   $5$  force direction) (Figure 4.1). Two shoulder joint angles used were:  $60^\circ$  and  $120^\circ$ . Two shoulder planes used were: sagittal plane and scapular plane. Three force levels used were: 0 lb., 2.5 lb., and 5 lb. Five force levels used were: pull back (PB), pull down (PD), pull left (PL), pull right (PR), and pull up (PU). Each task was performed for 10 seconds, and a 1-minute resting time was provided between the tasks.



**Figure 4.1: Forceful arm exertions performed at various angles, planes, and force direction conditions**

The EMG data were recorded from the shoulder muscles accessible by SEMG: supraspinatus, infraspinatus, deltoid, biceps, and triceps (long head). SEMG data for the deltoid muscle were recorded from three separate locations (middle, anterior and posterior), as these compartments of the deltoid muscle are sensitive to the direction of force exertion. Bipolar, Ag/AgCl, pre-gelled surface electrodes (Noraxon Inc., AZ, USA), with a 1 cm diameter and an inter-electrode distance of 2 cm, were used for data collection. The electrodes were attached to pre-amplified lead wires with a band-pass of 10-1000 Hz (gain of 500), CMRR >100 dB, and input impedance >100 MΩ. A wireless EMG system (Telemyo 2400 T G2, Noraxon Inc., AZ, USA) was used for SEMG data collection at a sampling rate of 1500 Hz. The root mean square values of the EMG data of the seven shoulder muscles were calculated and the average values of 12 participants were determined for each muscle.

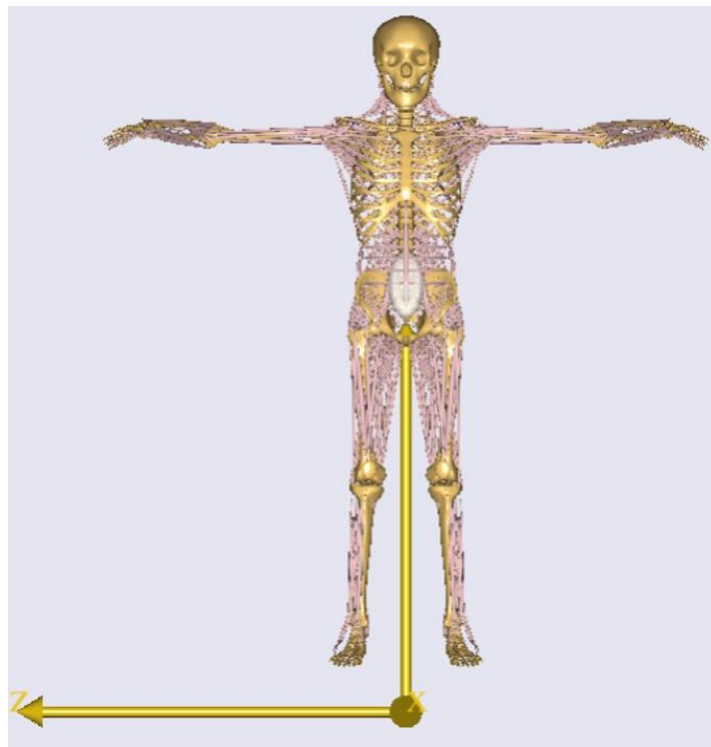
### 4.3 Biomechanical Modeling

#### 4.3.1 Description of the Model

The reaction forces for the glenohumeral joint were estimated by using a standing model from the repository of AnyBody Modeling System™ (version 5.0, AnyBody Technology, Aalborg, Denmark) (Figure 4.2). The shoulder complex of the model consists of 118 muscle fascicles on



each side of the body (left and right), bones, joints, and muscle-tendon units based on real physiological properties of human body. It defines the three main shoulder joints: the glenohumeral joint, the acromioclavicular joint, and the sternoclavicular joint. These joints could be driven by experimentally obtained kinematic and kinetic data. Muscle and joint forces were computed by inverse dynamics analysis and polynomial muscle recruitment pattern were employed for muscle force analysis. The standing model was modified according to the weight and height of individual participant and subsequently the model was ran using the working postures described in the previous section.



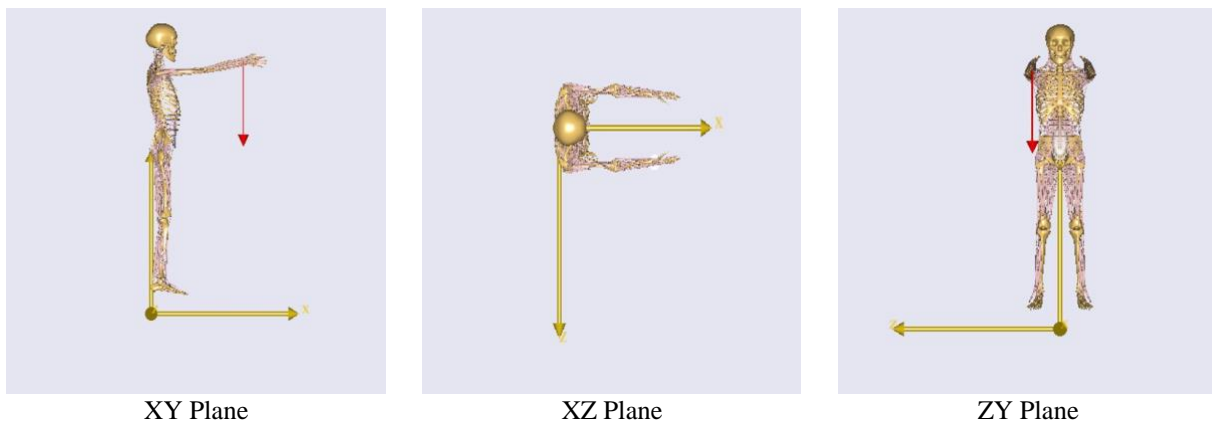
**Figure 4.2: Standing model obtained from AnyBody Modeling System™**

#### **4.3.2 Model Modification Procedure**

The standing model was modified according to the anthropometric data and posture of individual participants. The file AnyManUniform.any within the model was used to make the

appropriate participant-specific changes such as the standing height, body mass and gender. This feature of AnyBody allows the users to develop participant specific biomechanical models.

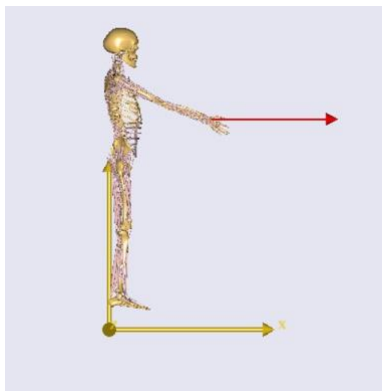
In order to complete the normalization process in the strain index computation process, the maximum isometric exertion was modelled and ran for 10 seconds for each participant. A task shown in Figure 4.3 was used to model the maximum isometric exertion. The task consisted of pulling up an external load of 5lb with shoulder joint flexed at  $90^\circ$ , forearm pronated at  $45^\circ$ , no shoulder abduction and no elbow flexion (Figure 4.3).



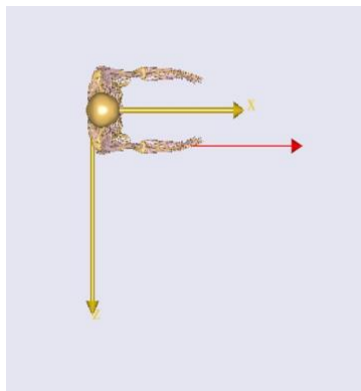
**Figure 4.3 Maximum isometric exertion**

Forty forceful arm exertions were simulated in AnyBody software for each participant. In the main file, the duration of each exertion was selected to be 10 seconds. The postures were obtained by modifying the Posture folder which is available in the Mannequin folder. The magnitudes of external forces were calculated in Newtons and were adjusted in the Load folder. For 2.5lb and 5lb external force, the selected values were 11.121N and 22.242N. The five orthogonal directions of exertions were obtained by taking the positive and negative values of the external forces. For example, the positive magnitudes of the external loads in x, y and z directions resulted in pulling back, pushing down and pulling left tasks respectively, while the negative magnitudes of the external loads in y and z direction resulted in pulling up and pulling right tasks. Figure 4.4, 4.5, 4.6 and 4.7 represents these postures in XY plane, XZ plane, and ZY plane.

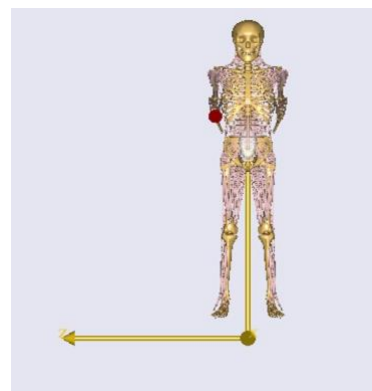
The first ten postures were modelled by adjusting the glenohumeral flexion at  $90^\circ$  (Sagittal plane) and glenohumeral abduction at  $60^\circ$  (Figure 4.4). Two external force levels (11.121N and 22.242N) were used (applied to the right hand of the model). In figure 4.4, (a) represents the exertion in backward direction, (b) represents the exertions in downward direction, (c) represents the exertions in left direction, (d) represents in right direction and (e) represents the exertions in upward directions.



XY Plane

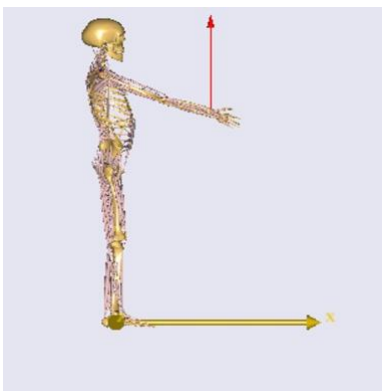


XZ Plane

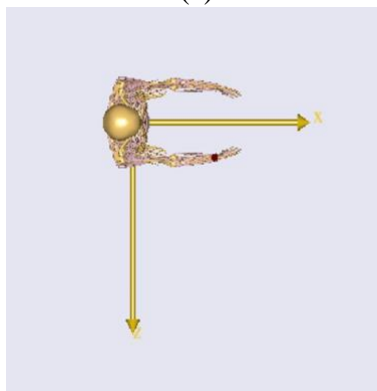


ZY Plane

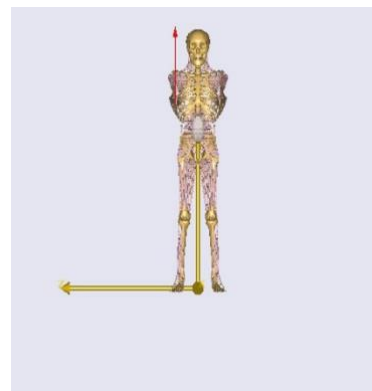
(a)



XY Plane

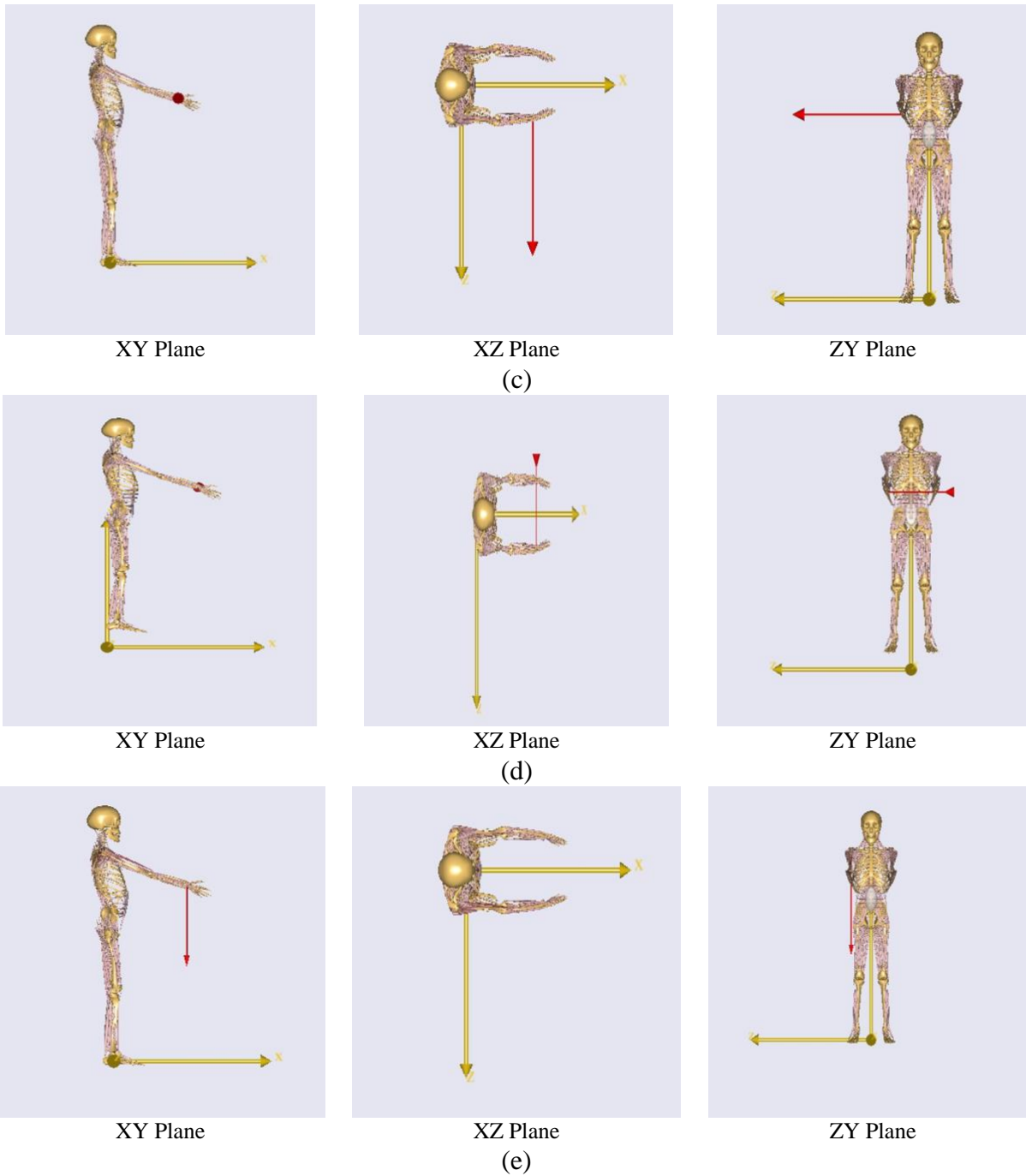


XZ Plane



ZY Plane

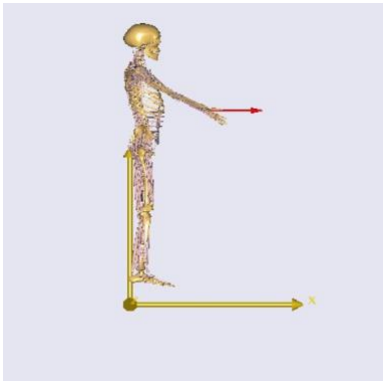
(b)



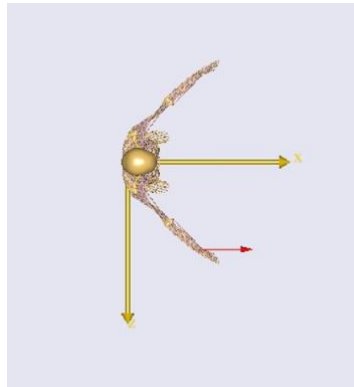
**Figure 4.4 Forceful exertions in sagittal plane with 60° shoulder angle**

The next ten postures were modelled by adjusting the glenohumeral flexion at 45° (Scapular plane) and glenohumeral abduction at 60° (Figure 4.5). This posture was used while applying two external force levels (11.121N and 22.242N) to the right hand of the model. In figure

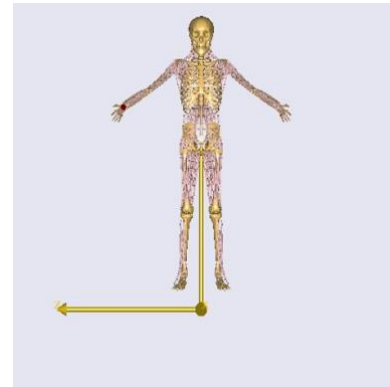
4.5, (a) represents the exertion in backward direction, (b) represents the exertions in downward direction, (c) represents the exertions in left direction, (d) represents in right direction and (e) represents the exertions in upward directions.



XY Plane

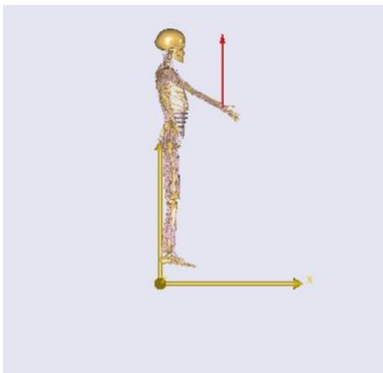


XZ Plane

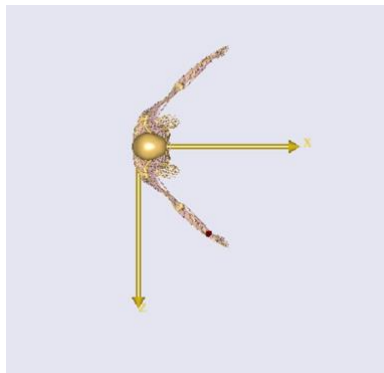


ZY Plane

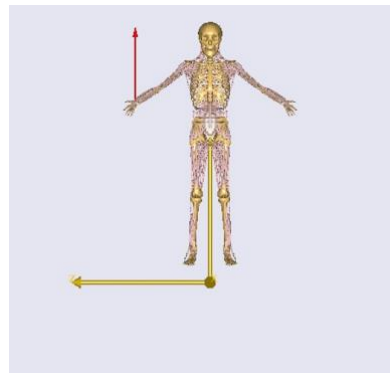
(a)



XY Plane

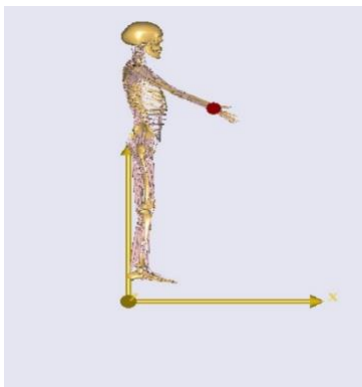


XZ Plane

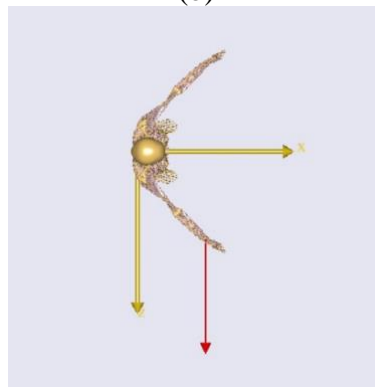


ZY Plane

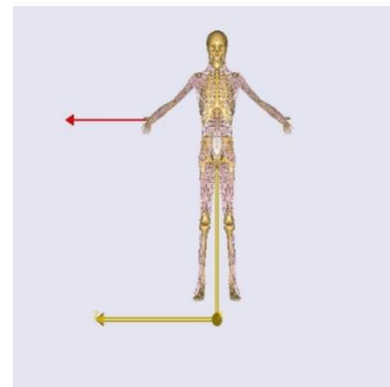
(b)



XY Plane

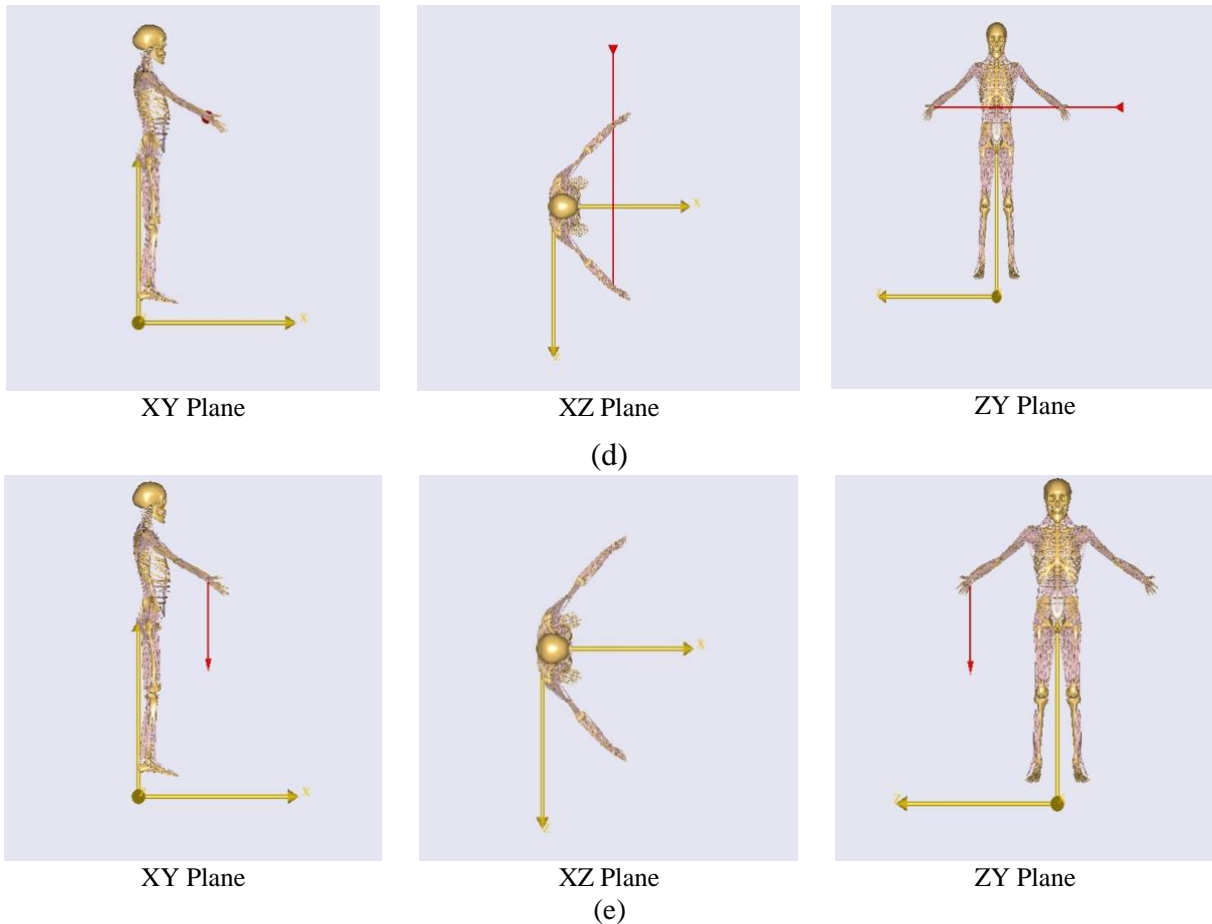


XZ Plane



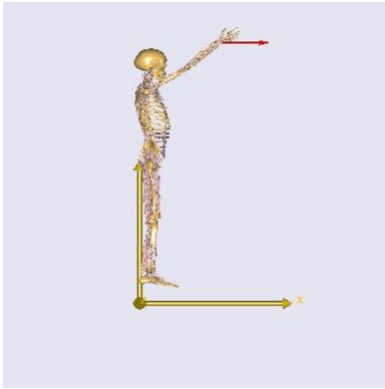
ZY Plane

(c)

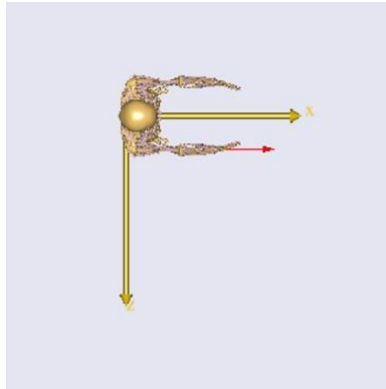


**Figure 4.5 Forceful exertions in scapular plane with 60° shoulder angle**

The next ten postures were modelled by adjusting the glenohumeral flexion at 90° (Sagittal) and glenohumeral abduction at 120° (Figure 4.6). This posture was used while applying two external force levels (11.121N and 22.242N) to the right hand of the model. In figure 4.6, (a) represents the exertion in backward direction, (b) represents the exertions in downward direction, (c) represents the exertions in left direction, (d) represents in right direction and (e) represents the exertions in upward directions.



XY Plane



XZ Plane



ZY Plane

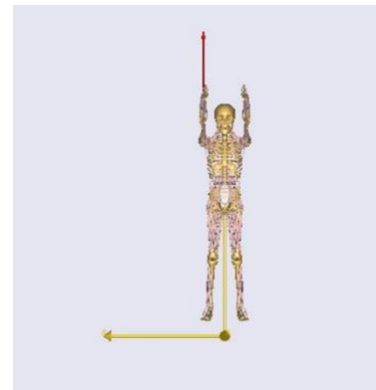
(a)



XY Plane

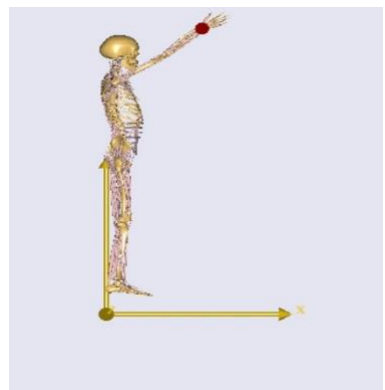


XZ Plane

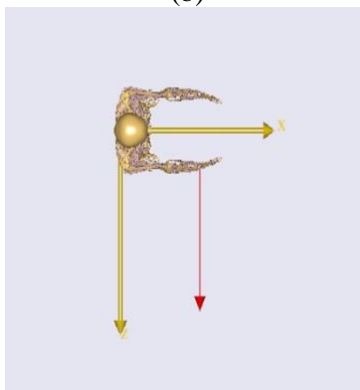


ZY Plane

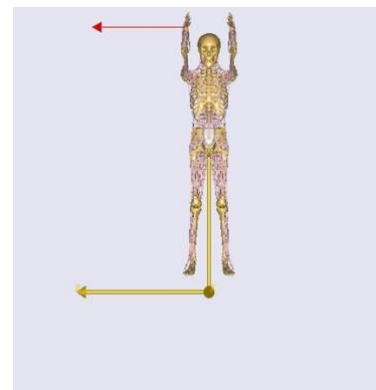
(b)



XY Plane

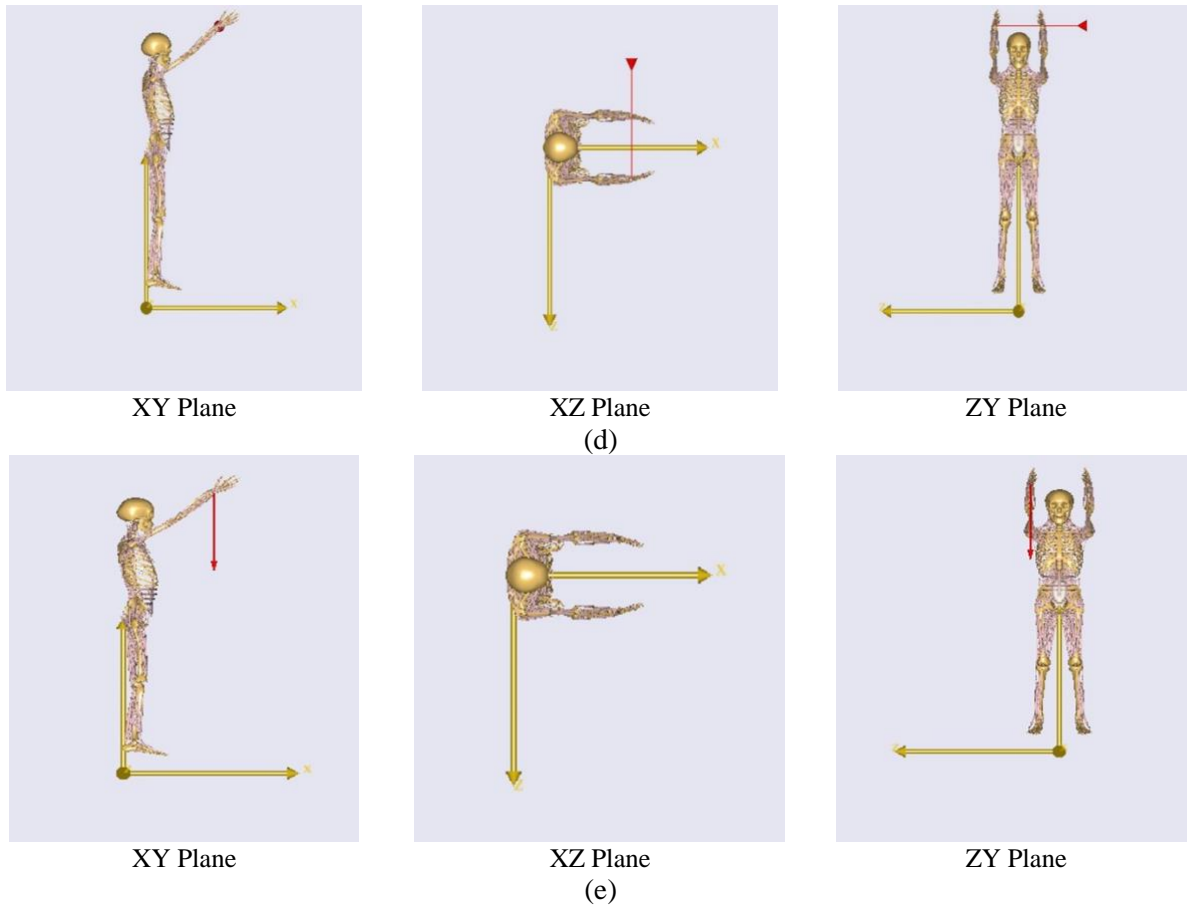


XZ Plane



ZY Plane

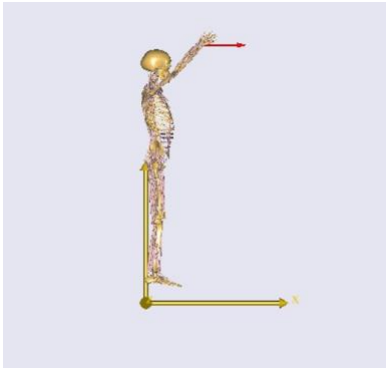
(c)



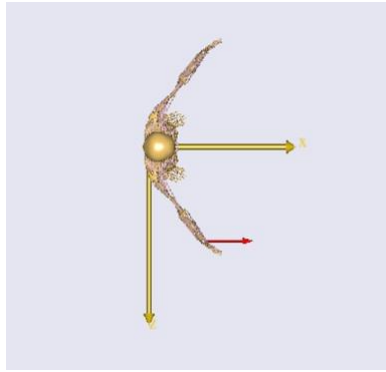
**Figure 4.6 Forceful exertions in sagittal plane with 120° shoulder angle**

The succeeding ten postures were modelled by adjusting the glenohumeral flexion at 45° (Scapular plane) and glenohumeral abduction at 120° (Figure 4.7). This posture was used while applying two external force levels (11.121N and 22.242N) to the right hand of the model. In figure 4.7, (a) represents the exertion in backward direction, (b) represents the exertions in downward direction, (c) represents the exertions in left direction, (d) represents in right direction and (e) represents the exertions in upward directions.

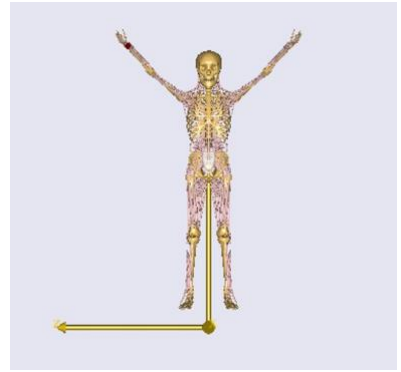




XY Plane

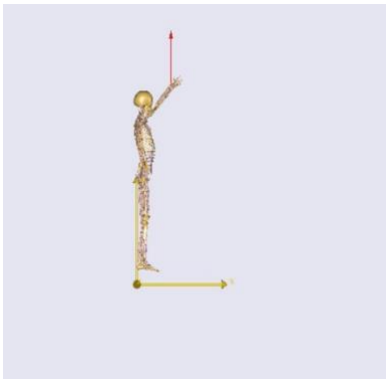


XZ Plane



ZY Plane

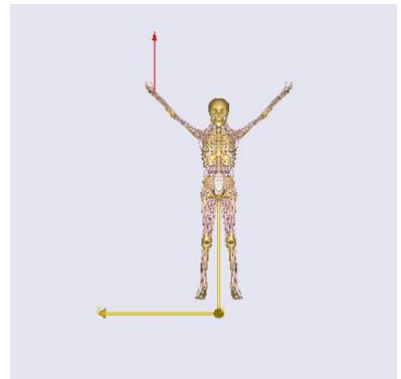
(a)



XY Plane

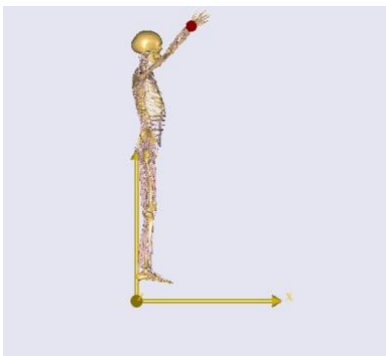


XZ Plane

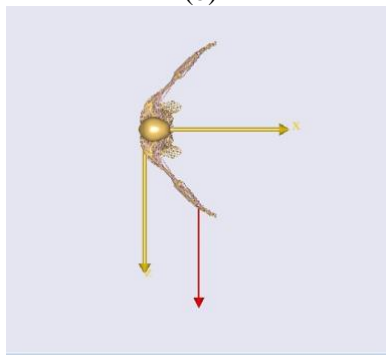


ZY Plane

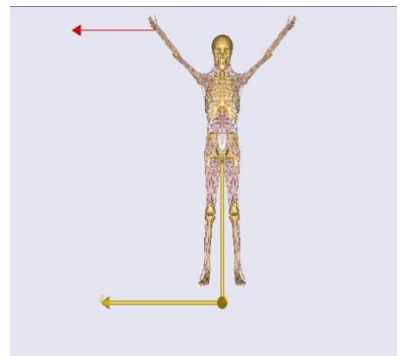
(b)



XY Plane

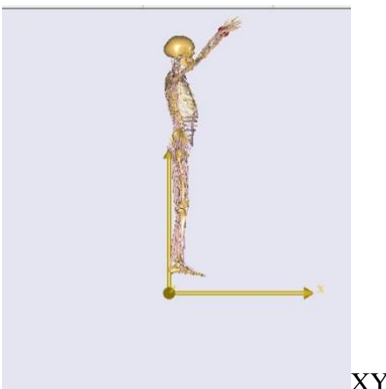


XZ Plane



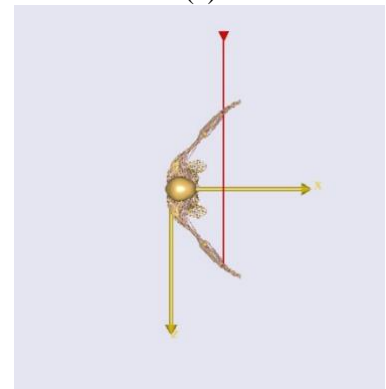
ZY Plane

(c)

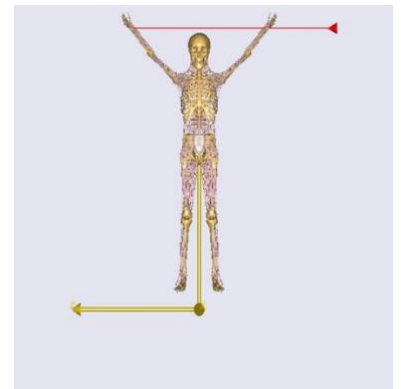


Plane

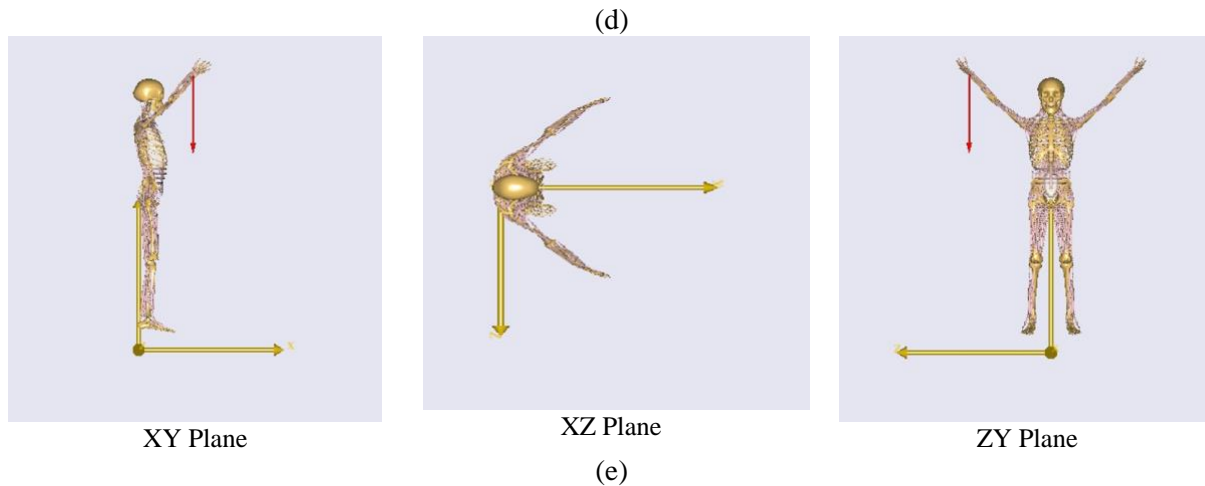
XY



XZ Plane



ZY Plane



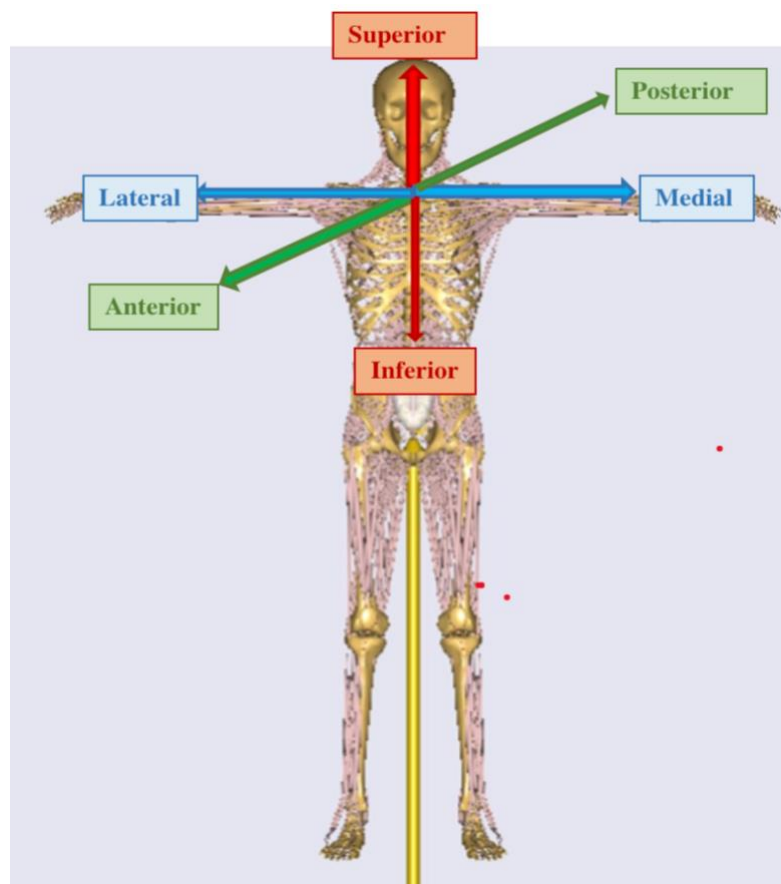
**Figure 4.7 Forceful exertions in scapular plane with 120° shoulder angle**

#### 4.4 Data Processing

The modelling analysis was performed to estimate internal reaction forces for the glenohumeral joint. The reaction and the muscle forces required to generate motion or sustain body posture were computed using inverse-dynamic methods by solving a multi-body dynamics problem. The muscle recruitment in the inverse dynamics process were solved using polynomial optimization within which the objective function was to get more synergy and distribute the load evenly between the muscles. The polynomial recruitment pattern allows the organism to carry larger loads without overloading any individual muscle. Based on the anthropometric data of the participants, the standing models were scaled and subsequently forty conditions (2 external load  $\times$  2 shoulder angle  $\times$  2 shoulder plane  $\times$  5 direction) were loaded and simulated in AnyBody Modeling System™ for 10 seconds. Additionally, maximum isometric exertions (MIE) were performed for each model. After completing the inverse kinematics analysis, the internal joint reaction forces of the right shoulder complex in the three anatomical directions (distraction, inferior-superior, and anterior-posterior) were obtained. The mean of the forces during the force exertion period were used for further calculation of the shoulder strain index.

#### 4.4.1 Shoulder Strain Index Calculation

The shoulder strain index estimates the strain placed on the glenohumeral joint during forceful arm exertions. It was developed based on the relationship between the compressive and translational forces in the shoulder joint. When an external force is applied to the hand, the joint reaction forces in the inferior-superior and anterior-posterior direction moves the humeral head away from the glenoid cavity and destabilizes the shoulder joint. On the other hand, the reaction forces in the medial-lateral direction produce a compressive force in order to counteract the external force and thus stabilizes the glenohumeral joint. The reaction forces obtained from the model configuration were positive in the lateral, superior, and posterior directions. As a result, the resultant of lateral-medial and inferior-superior reaction forces was found in the frontal plane (Rf). Therefore, the resultant of medial and anterior-posterior reaction forces was found in the transverse planes (Rt). The resultant forces were calculated using equations 4.1 and 4.2.

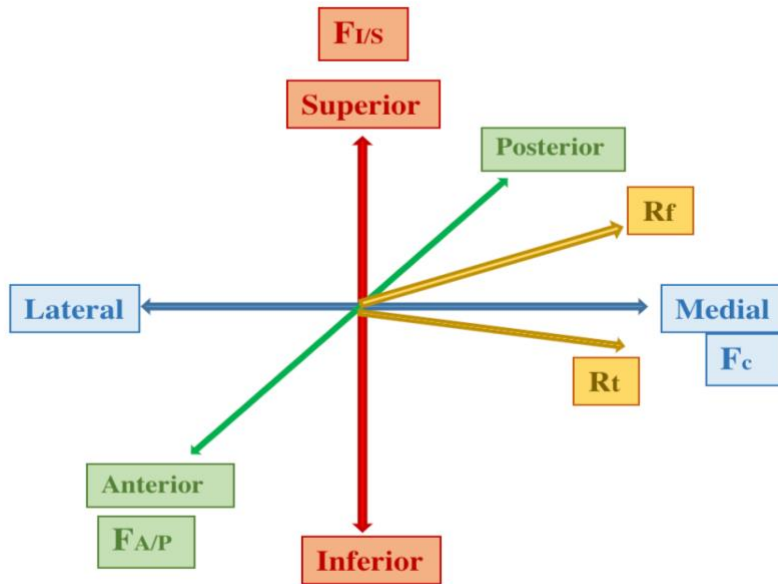


**Figure 4.8: Shoulder joint force directions**

$$R_f = \sqrt{F_c^2 + F_{I/S}^2} \quad (4.1)$$

$$R_t = \sqrt{F_c^2 + F_{A/P}^2} \quad (4.2)$$

Here,  $R_f$  is the resultant force acting in the frontal plane,  $R_t$  is the resultant force acting in the transverse plane,  $F_{I/S}$  is the joint reaction force in the interior-superior direction,  $F_{A/P}$  is the joint reaction force in the anterior- superior direction,  $F_c$  is the joint reaction force acting in the lateral direction (Figure 4.9). For calculating the angular displacement of the resultant forces in the frontal and transverse planes, the directions of the resultant forces were used. The angular displacements were calculated using equations 4.3 and 4.4.



**Figure 4.9: Shoulder joint reaction force directions**

$$\beta = \tan^{-1} \frac{F_{I/S}}{F_c} \quad (4.3)$$

$$\gamma = \tan^{-1} \frac{F_{A/P}}{F_c} \quad (4.4)$$

Here,  $\beta$  is the angular displacement of resultant force in the frontal plane and  $\gamma$  is the angular displacement of resultant force in transverse plane.

To reduce the variability between the resultant forces, they were normalized with respect to the resultant reaction force of MIE exertion using equations 4.5 and 4.6 for individual participants. Similarly, the resultant angular displacements of the joint reaction forces were normalized with respect to that maximum angular displacement using equations 4.7 and 4.8. After calculating the normalized values, the shoulder indices for frontal and transverse planes were determined by equations 4.9 and 4.10. The values for frontal and transverse shoulder indices

should be within a range of 0-50. As these indices might have a positive or negative value, only their absolute values were used for estimating the final shoulder index using equation 4.11.

$$N_f = \frac{R_f}{R_{f-MIE}} \quad (4.5)$$

$$N_t = \frac{R_t}{R_{t-MIE}} \quad (4.6)$$

$$A_f = \frac{\beta}{45} \quad (4.7)$$

$$A_t = \frac{\gamma}{45} \quad (4.8)$$

$$I_f = N_f \times A_f \times 100. \quad (4.9)$$

$$I_t = N_t \times A_t \times 100. \quad (4.10)$$

$$I = |I_f| + |I_t| \quad (4.11)$$

Here,  $N_f$  is the normalized resultant reaction force in the frontal plane,  $N_t$  is the normalized resultant reaction force in the transverse plane,  $R_{f-MIE}$  is the resultant reaction force of maximum isometric exertion in the frontal plane,  $R_{t-MIE}$  is the resultant reaction force of maximum isometric exertion in the transverse plane,  $I_f$  is the strain index in the frontal plane,  $I_t$  is the strain index in the transverse plane and  $I$  is the strain index.

The shoulder strain indices for 12 subjects and the average strain indices are shown in Appendix I.

#### 4.4.2 Statistical Analysis

General Linear Analysis of Variance (ANOVA) model was used to understand the effect of forceful arm exertions on the strain index. For running the ANOVA model, shoulder angles, shoulder plane, external load and exertion directions were treated as the independent (fixed) factors

and strain index was treated as a dependent factor. A total of 480 strain index values were used to run the model and two factors interactions were considered. The statistical significance level of  $\alpha = 0.05$  was used. After finding the significant main and interaction effects, the pairwise comparisons of means were conducted using student's t-test. To quantify the strength of relationship between model-based strain index and the physiological strain (estimated using the muscle activity), correlation analysis was performed. The correlation analysis was performed between (i) strain index data and the activity of individual muscle (local muscle demand); (ii) strain index data and the total shoulder activity (global shoulder joint demand).

## Chapter 5: Results

### 5.1 Effect of Forceful Arm Exertions on Strain Index (SI)

The effect of shoulder angle ( $p=0.0015$ ), shoulder plane ( $p<0.0001$ ), external force ( $p<0.0001$ ) and exertion direction ( $p<0.0001$ ) on strain index was statistically significant. The F-ratio and p-values for the main and interaction effects of the five factors on SI are shown in Table 5.1. The interaction effects between shoulder angle and shoulder plane ( $p<0.0001$ ), shoulder angle and external force ( $p=0.0198$ ), shoulder angle and exertion direction ( $p<0.0001$ ), shoulder plane and external force ( $p=0.0307$ ), shoulder plane and exertion direction ( $p<0.0001$ ) and external force and exertion direction ( $p<0.0001$ ) were found to be statistically significant.

**Table 5.1 ANOVA test results**

<b>Main effect</b>		
<b>Source</b>	<b>F Ratio</b>	<b>Prob &gt; F</b>
Shoulder Angle	10.1497	0.0015*
Shoulder Plane	367.3435	<.0001*
External Force (lb)	431.7706	<.0001*
Exertion direction	1141.309	<.0001*
<b>Interaction effect</b>		
<b>Source</b>	<b>F Ratio</b>	<b>Prob &gt; F</b>
Shoulder Angle*Shoulder Plane	90.2889	<.0001*
Shoulder Angle*External Force (lb)	5.4675	0.0198*
Shoulder Angle*Exertion direction	46.9828	<.0001*
Shoulder Plane*External Force (lb)	4.6989	0.0307*
Shoulder Plane*Exertion direction	27.3401	<.0001*
External Force (lb)*Exertion direction	49.0608	<.0001*

\* Means statistically significant

Statistically significant main and interaction effects were subjected to post hoc comparison using student's t test. The student's t test results along with the means for the levels of different



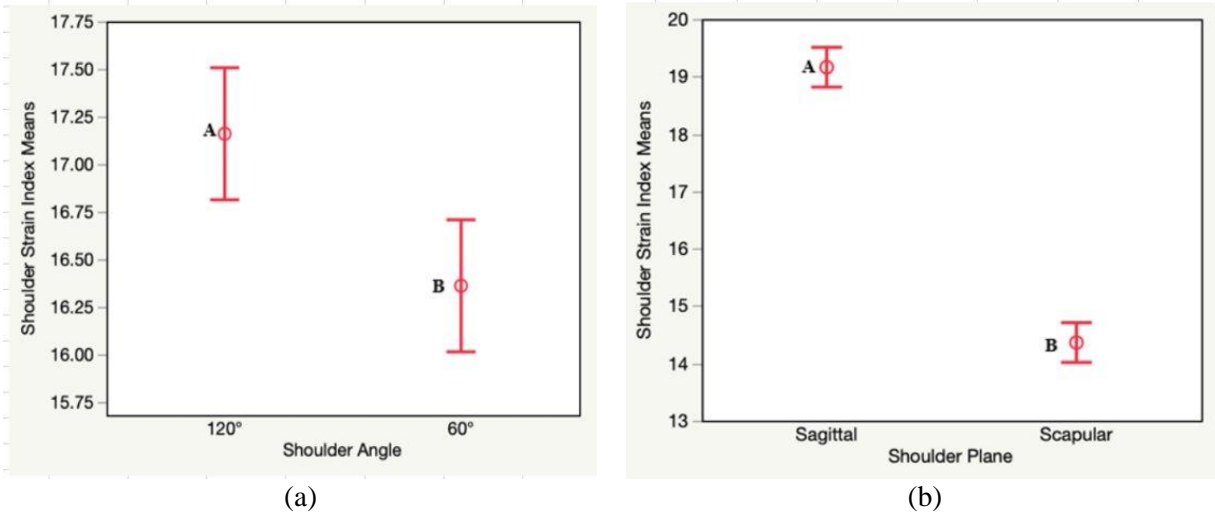
factors are shown in Table 5.2 and 5.3. The connecting letters next to level within a factor indicate that the levels that do not share the same letter are statistically significantly different from each other at  $\alpha=0.05$ . Figure 5.1 and 5.2 represents the mean and 95% Confidence Interval (CI) of strain indices at all levels and combinations of the levels of the treatment factors, respectively.

**Table 5.2 Pairwise comparison of student's t test (factor levels)**

<b>Factor (Independent variable)</b>	<b>Level</b>	<b>Connecting Letters</b>	<b>Mean</b>	<b>t ratio</b>	<b>Prob&gt; t</b>
Shoulder Angle	120°	A	17.16	3.20	0.0015
	60°	B	16.36	-3.20	0.0015
Shoulder Plane	Sagittal	A	19.16	19.25	<0.0001
	Scapula	B	14.36	-19.25	<0.0001
External Force (lb)	5	A	19.36	20.87	<0.0001
	2.5	B	14.16	-20.87	<0.0001
Exertion Direction	PU	A	30.86	56.55	<0.0001
	PL	B	17.17	1.62	0.1054
	PR	B	16.82	0.23	0.8166
	PB	C	14.62	-8.59	<0.0001
	PD	D	4.34	-49.81	<0.0001

From the student's t test, it was found that the mean shoulder strain index at 120° shoulder angle was significantly higher than the mean shoulder strain index at 60° shoulder angle. The mean values of the SI were 17.16 and 16.36 respectively. These values with 95% CI are shown on figure 5.1 (a).

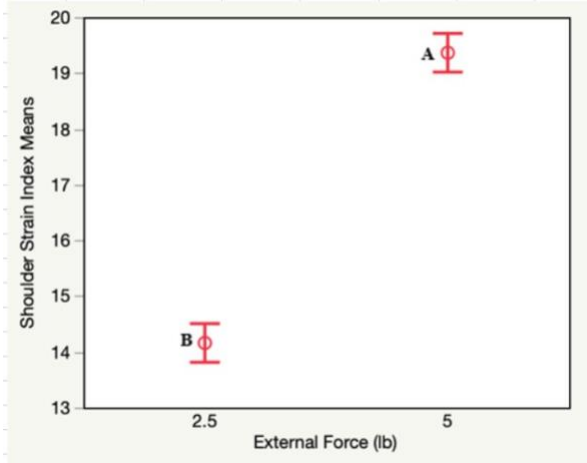
The student's t test also revealed that the sagittal plane generated a significantly higher mean strain index than the scapular plane. The mean strain indices of sagittal plane and scapular plane are 19.16 and 14.36. Figure 5.1 (b) represents these values with 95% CI.



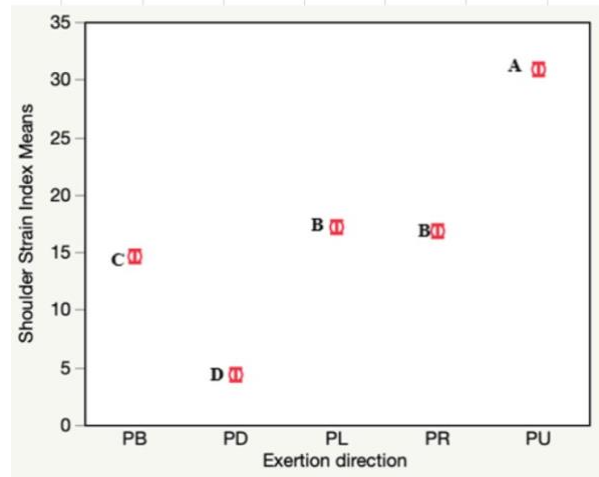
**Figure 5.1: Mean and 95% CI for all combinations of main factors**

Student's t test for external force levels showed that the mean strain index at 5lb was significantly higher than at 2.5 lb load. The mean strain index for 5lb load was 19.36 and the mean strain index for 2.5lb was 14.16. Figure 5.1 (c) represents these values with 95% CI.

From the student's t-test it was found that the mean strain index for pulling up task was significantly higher compared to pulling left, pushing right, pulling back and pulling down tasks. The magnitude of the mean strain index of pulling up was 30.86. Additionally, the strain indices for pulling left and pushing right directions were larger than pulling back and pulling down tasks. However, the difference between pulling left and right was not significant. The mean strain index for pulling left and right was 17.17 and 16.82. It was also noticed that pulling back and pulling down tasks were significantly different from each other with mean strain index values of 14.62 and 4.34, respectively. Figure 5.1 (d) shows these values with 95% CI.



(c)



(d)

**Figure 5.2: Mean and 95% CI for all combinations of main factors (Continued)**

The results of pairwise comparisons based on the student's t test for the interaction effect are summarized in Table 5.3.

**Table 5.3 Pairwise comparison of student's t test (combination of factor levels)**

Factor (Combination of independent variables)	Level	Connecting Letters	Mean	t ratio	Prob> t
Shoulder Angle*Shoulder Plane	120°, Sagittal	A	20.75	9.54	<0.0001
	60°, Sagittal	B	17.57	-9.54	<0.0001
	60°, Scapular	C	15.15	9.54	<0.0001
	120°, Scapular	D	13.57	-9.54	<0.0001
Shoulder Angle*External Force Level	120°, 5	A	20.06	2.35	0.0193
	60°, 5	B	18.67	-2.35	0.0193
	120°, 2.5	C	14.27	-2.35	0.0193
	60°, 2.5	C	14.05	2.35	0.0193
Shoulder Angle*Exertion direction	60°, PU	A	32.93	9.90	<0.0001
	120°, PU	B	28.79	-9.90	<0.0001
	120°, PR	C	18.88	6.64	<0.0001
	60°, PL	D	17.71	3.80	0.0002

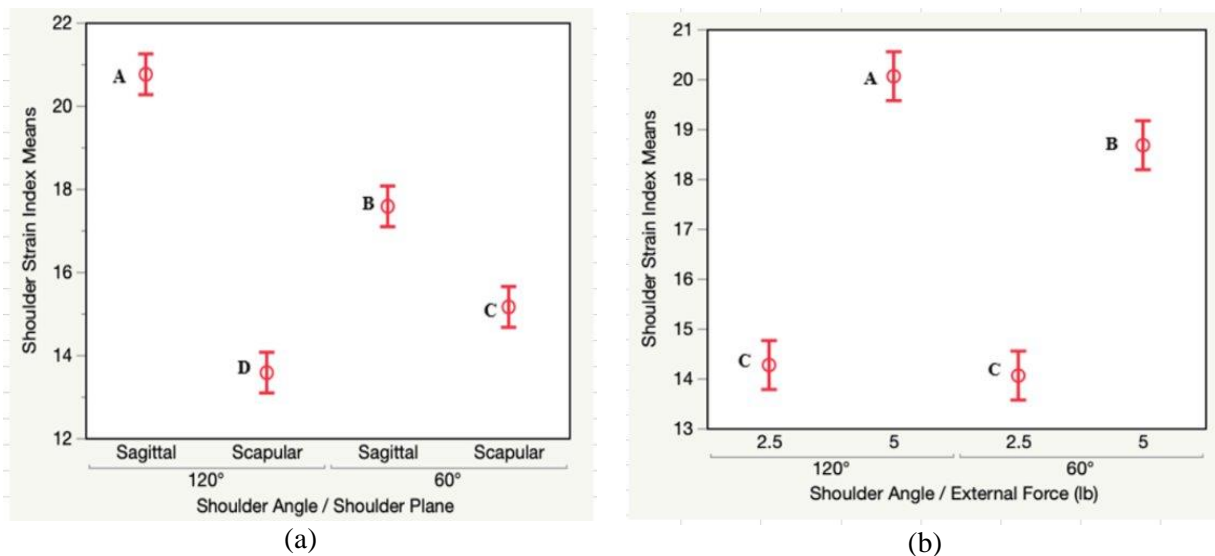
**Table 5.3 Pairwise comparison of student's t test (combination of factor levels)**

Source	Level	Connecting Letters Report	Mean	t ratio	Prob> t
Shoulder Angle*Exertion direction	120°, PB	D	17.21	8.80	<0.0001
	120°, PL	D	16.62	-3.80	0.0002
	60°, PR	E	14.76	-6.64	<0.0001
	60°, PB	F	12.03	-8.80	<0.0001
	60°, PD	G	4.38	1.75	0.0816
	120°, PD	G	4.30	-1.75	0.0816
Shoulder Plane* External Force	Sagittal,5	A	22.04	2.18	0.0300
	Scapular,5	B	16.69	-2.18	0.0300
	Sagittal,2.5	B	16.29	-2.18	0.0300
	Scapular,2.5	C	12.03	2.18	0.0300
Shoulder Plane* Exertion Direction	Sagittal, PU	A	35.45	8.75	<0.0001
	Scapular, PU	B	26.28	-8.75	<0.0001
	Sagittal, PL	C	20.07	2.00	0.0461
	Sagittal, PR	C	19.16	-0.23	0.8216
	Sagittal, PB	D	15.95	-4.30	<0.0001
	Scapular, PR	E	14.48	0.23	0.8216
	Scapular, PL	EF	14.27	-2.00	0.0461
	Scapular, PB	F	13.29	4.30	0.0001
	Sagittal, PD	G	5.19	-6.22	<0.0001
	Scapular, PD	H	3.49	6.22	<0.0001
External Force (lb)*Exertion Direction	5, PU	A	36.89	13.77	<0.0001
	2.5, PU	B	24.83	-13.77	<0.0001
	5, PL	C	19.23	-2.14	0.0327
	5, PR	C	18.84	-2.33	0.0202
	5, PB	D	15.69	-6.16	<0.0001
	2.5, PL	D	15.10	2.14	0.03217
	2.5, PR	D	14.80	2.33	0.0202
	2.5, PB	E	13.56	6.16	<0.0001
	5, PD	F	6.16	-3.13	0.0018
	2.5, PD	G	2.52	3.13	0.0018

The two levels of shoulder angle and shoulder plane resulted in four combinations: 120° shoulder angle in sagittal plane, 60° shoulder angle in sagittal plane, 60° shoulder angle in scapular plane and 120° shoulder angle in scapular plane. The mean strain indices of these combinations

are shown in figure 5.2 (a) with 95% CI. The largest and smallest means were 20.74971 and 13.57037 which are found at the combination of 120° shoulder angle in sagittal plane and 120° shoulder angle in scapular plane respectively. The student's t test indicated that the tasks performed in the sagittal plane were more strenuous than in the scapular plane. At 120° shoulder angle, the increase in the stress between sagittal and scapular plane was much higher than that at 60° shoulder angle.

The four combinations of shoulder angle and external force were found at 120° shoulder angle with 5lb force, 60° shoulder angle with 5lb force, 120° shoulder angle with 2.5lb force and 60° shoulder angle with 2.5lb force. From figure 5.2 (b), it can be clearly said that the combination of 120° shoulder plane and 5lb force generates higher shoulder strain and the combination of 2.5lb force with both 120° and 60° shoulder angle generates lower shoulder strain compared to the combinations of 60° shoulder angle and 5lb external load. The student's t test indicated that the tasks performed using 5lb external load were more strenuous than those tasks performed using 2.5lb load. At 120° shoulder angle, the increase in the stress between 2.5lb and 5lb external load was much higher than that at 60° shoulder angle.

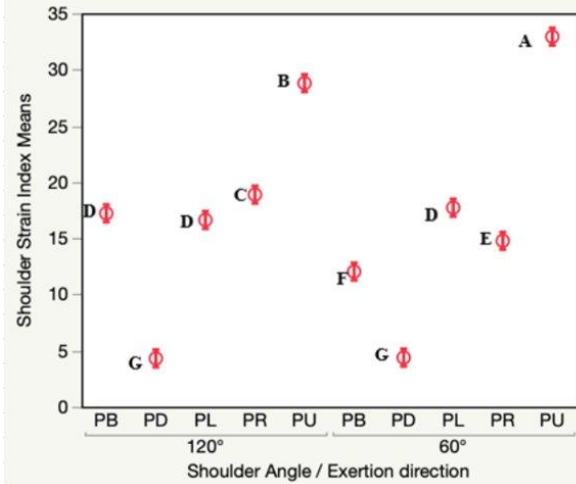


**Figure 5.3 Mean and 95% CI for all combinations of main factor levels**

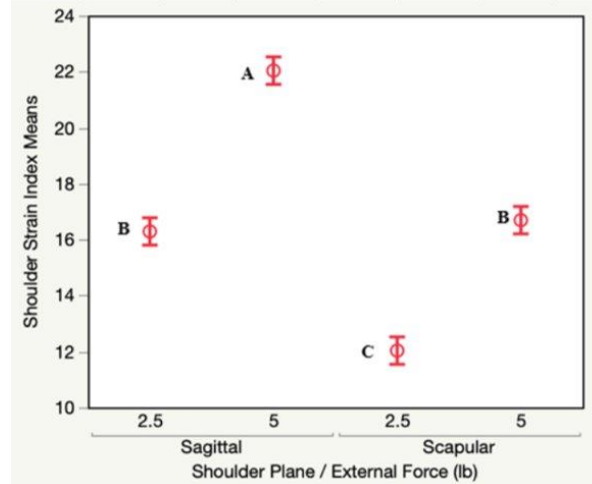
The two shoulder angles and five exertion directions resulted in a total of ten combinations: 60° shoulder angle and pulling up, 120° shoulder angle and pulling up, 120° shoulder angle and

pushing right, 60° shoulder angle and pulling left, 120° shoulder angle and pulling back, 120° shoulder angle and pulling left, 60° shoulder angle and pushing right, 60° shoulder angle and pulling back, 60° shoulder angle and pulling down, and 120° shoulder angle and pulling down. The means and CIs of these combination are shown in figure 5.2 (c). The student's t test informed that pulling up at 60° shoulder angle produces significantly higher shoulder strain compared to the other combination. Additionally, there is no significant difference between pulling left and pulling back at 60° shoulder angle as well as pulling left at 120° shoulder angle. However, the mean strain indices of these three combinations are significantly different from the mean strain indices of other combinations. The student's t test indicated that the tasks performed in pull up direction were more strenuous than those tasks performed in all other directions. At 60° shoulder angle, the strain index of pull up direction was much higher than that at 120° shoulder angle.

The two levels of shoulder plane and external force resulted in four combinations: 5lb at sagittal plane, 5lb at scapular plane, 2.5 lb at sagittal plane and 2.5 lb at scapular plane. From figure 5.2 (d) it can be said that 5lb external force at sagittal plane and 2.5 lb external force at scapular plane has a higher and lower strain index respectively than other combinations and the student's t test indicates that the difference is significant. Conversely, the mean strain indices of 5lb at scapular plane and 2.5 lb at sagittal plane are not different from each other but, they vary significantly from other two combinations. The student's t test indicated that the tasks performed using 5lb external load were more strenuous than those tasks performed using 2.5lb load. At sagittal plane, the increase in the stress between 2.5lb and 5lb external load was much higher than that at scapular plane.



(c)



(d)

**Figure 5.4 Mean and 95% CI for all combinations of main factor levels (Continued)**

The two shoulder plane and five exertion directions lead to ten combinations: pulling up at sagittal plane, pulling up at scapular plane, pulling left at sagittal plane, pulling right at sagittal plane, pulling back at sagittal plane, pulling right at scapular plane, pulling left at scapular plane, pulling back at scapular plane, pulling down at sagittal plane and pulling down at scapular plane. The mean strain indices of these combinations are shown in figure 5.2 (e). The student's t test reported that pulling up at sagittal plane and pulling down at scapular plane had the higher and lower strain indices compared to the other combinations and the differences were significant. Additionally, it was found that pulling left and right at sagittal plane had no significant difference among themselves but they were different from other combinations. Similarly, pulling left and right at scapular plane as well as pulling left and back at scapular plane were not different from each other but, they were significantly different from other combinations. The student's t test indicated that the tasks performed in pull up direction were more strenuous than those tasks performed in other directions. At sagittal plane, the strain index of pull up direction was much higher than the strain index pull up task at scapular plane.

The combinations of two external force levels at five exertion directions were found at pulling up 5lb, pulling up 2.5lb, pulling left 5lb, pushing right 5lb, pulling back 5lb, pulling left 2.5lb, pulling right 2.5 lb, pulling back 2.5lb, pulling down 5lb and pulling down 2.5lb. These combinations along with their mean strain indices and 95% CI are shown in figure 5.2 (f). The

student's t test of these combinations discloses that pulling up 5lb induces significantly higher pain compared to all other combinations whereas pulling down 2.5lb causes significantly less pain. It also specifies that though pulling 5lb external load in the left and right directions has no significant difference, they are different from other exertions with force levels and direction. Similarly, pulling 5lb in the backward direction, pushing 2.5lb in the right and left directions did not exhibit any significant differences among themselves. The student's t test indicated that the tasks performed using 5lb external load were more strenuous than those tasks performed using 2.5 lb load. At pull up direction, the strain index of 5lb task was much higher than the strain index of 2.5lb task.

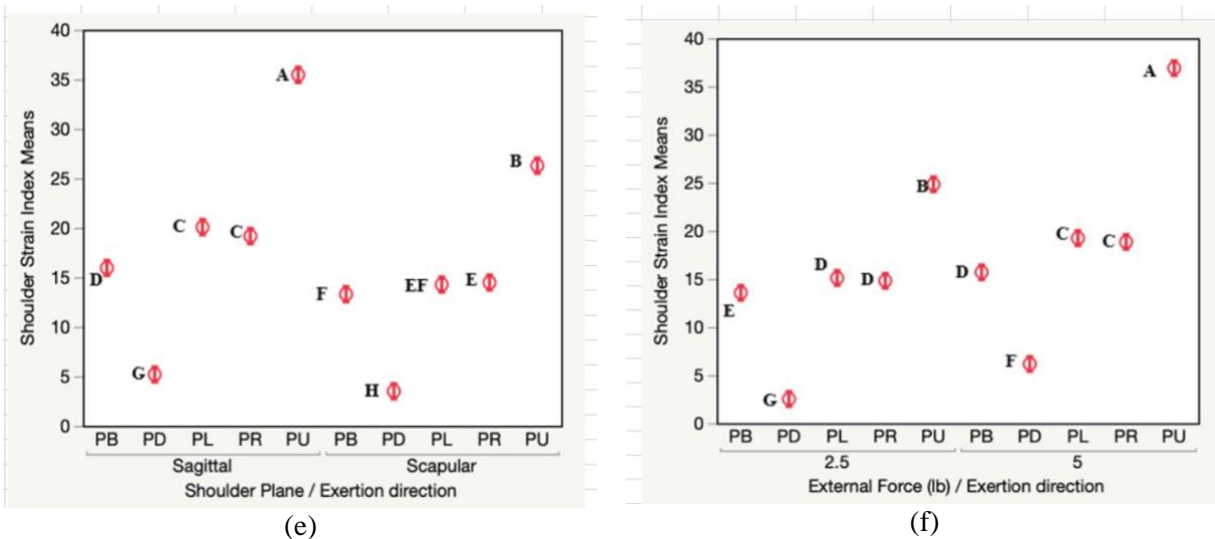


Figure 5.5 Mean and 95% CI for all combinations of main factor levels (Continued)

## 5.2 Correlation between Muscle Activation and Shoulder Strain Index

Correlation analysis was conducted to test the relationship between the SI and individual muscle demand as well as the average muscle demand. Table 5.4 shows the correlation between the average shoulder strain indices and seven shoulder muscles: Supraspinatus, Infraspinatus, Middle Deltoid, Anterior Deltoid, Posterior Deltoid, Biceps and Triceps. For the correlation analysis, the Root Mean Square (RMS) values of the individual shoulder muscles EMG data and the average strain indices of 12 participants for the forty arm exertion tasks were calculated. A

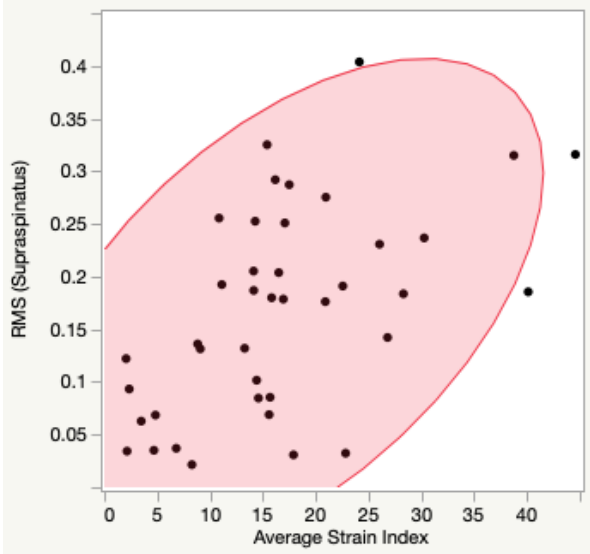


multivariate correlation analysis was performed between the RMS data and the average strain indices.

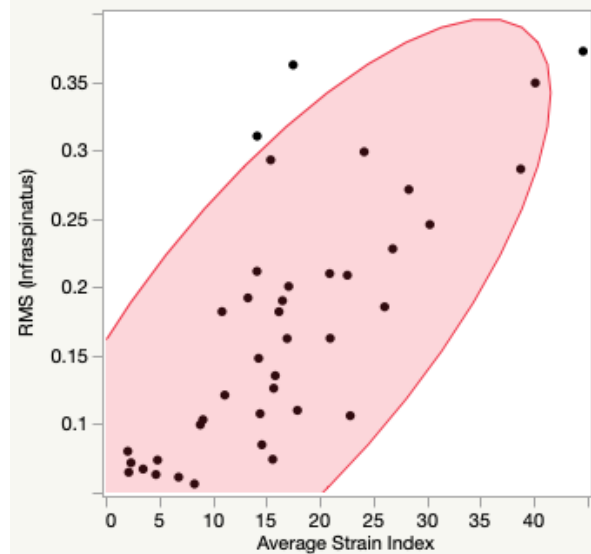
**Table 5.4 Correlation between individual muscles and average strain index**

Correlations	Supraspinatus	Infraspinatus	Middle deltoid	Anterior deltoid	Posterior deltoid	Biceps	Triceps
Average Strain Index	0.546	0.760	0.556	0.672	0.399	0.637	0.116

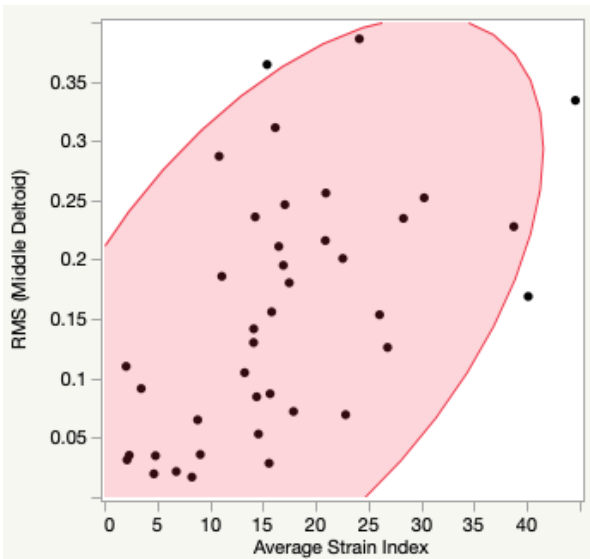
The correlation values ( $r$ ) between strain index and the individual muscle activity ranged from 0.116 to 0.760. Figure 5.3 shows the relationship between the individual shoulder muscle activity and the strain index. The strongest relationship was found for infraspinatus ( $r=0.760$ ), followed by anterior deltoid ( $r=0.672$ ) and biceps ( $r=0.637$ ) (Figure 5.6 b, d and f). The middle deltoid ( $r=0.556$ ) and supraspinatus ( $r=0.546$ ) also exhibited good relationship (Figure 5.6 c and a). However, moderate and weak correlations were found for posterior deltoid ( $r=0.399$ ) and triceps ( $r=0.116$ ) (Figure 5.6 e and g)



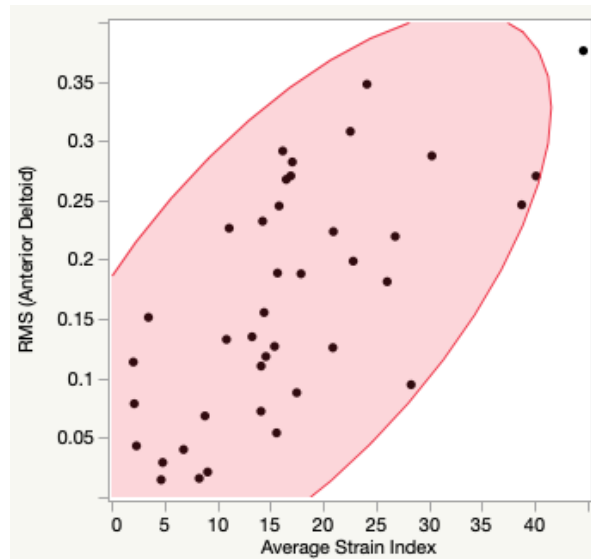
(a)



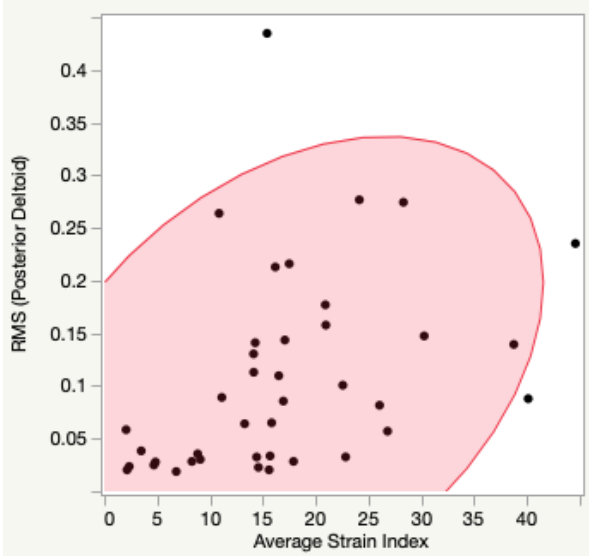
(b)



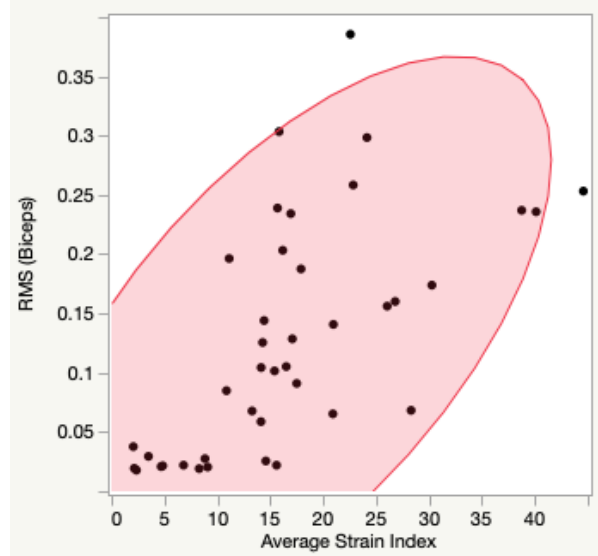
(c)



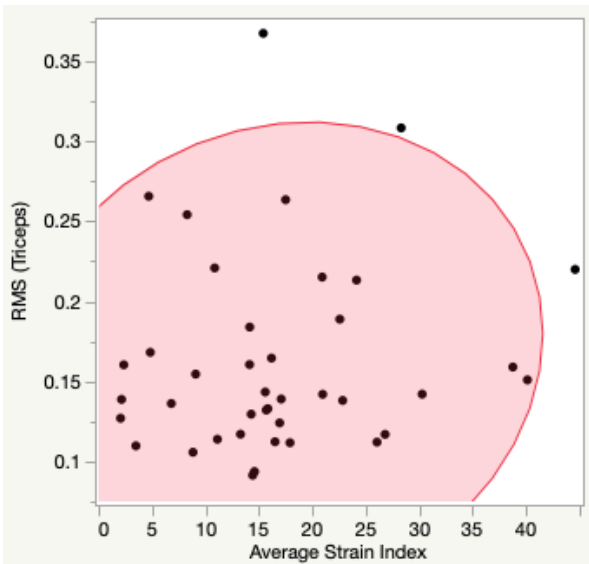
(d)



(e)



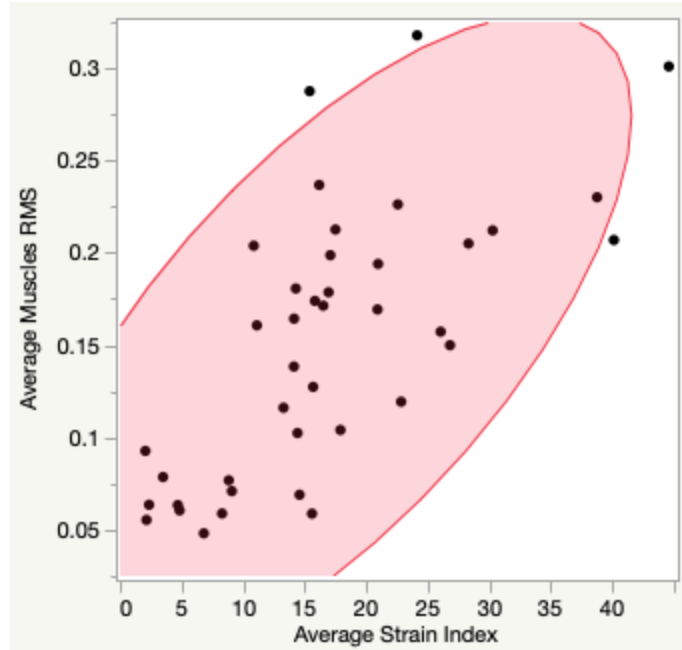
(f)



(g)

**Figure 5.6 Relationship between individual muscles and average strain index**

For assessing the relationship between the overall shoulder demand and strain index, a separate correlation analysis was performed. The test reported a correlation value of 0.6944 which indicates a strong relationship (Figure 5.4).



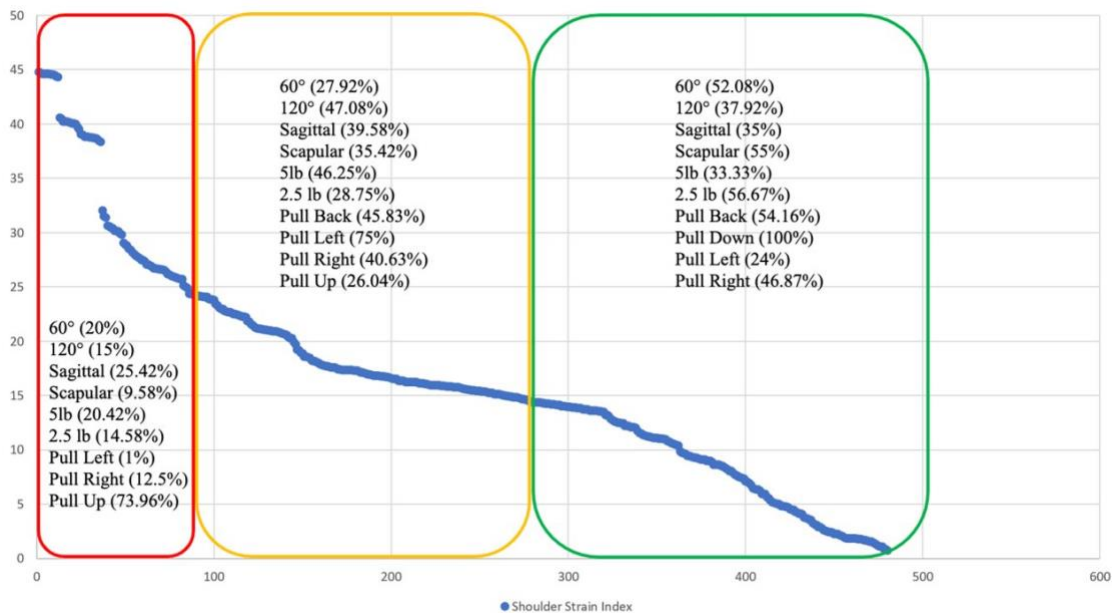
**Figure 5.7 Relationship between average muscle activation and strain index**

## Chapter 6: Discussion and Conclusion

In industrial and occupational settings, workers are often required to perform forceful arm exertions. Almost all forceful arm exertions involve active contribution by the shoulder joint. The main shoulder joint, glenohumeral joint, is the most versatile joint with range of motion covering almost 65% of a sphere. Humans can perform forceful arm exertions of various magnitudes, in any possible direction using postures that can be encompassed within 65% of sphere using the shoulder joint. The flexibility provided by the shoulder joint makes it one of the most functional joints in the human body. The functionality makes the shoulder joint also vulnerable to a range of work-related risk factors. In the literature, shoulder joint has been characterized as a joint with infinite functional degrees of freedom making it one the most challenging joint to study from risk assessment point of view. Although multiple studies have evaluated the load-response relationship for the shoulder joint due to the exposure to different work-related factors, there is scarcity of risk assessment tools. Very recently Chowdhury et al. (2016) developed a strain index to quantify the strain imposed on the main shoulder joint (glenohumeral joint) during forceful arm exertions. The strain index ranges between 0 and 100 and is derived based on the concept of concavity compression mechanism. For any forceful arm exertion, strain index can be computed provided the direction and magnitude of internal joint reaction forces acting at the glenohumeral joint are known. Upon development of the strain index, Chowdhury et al. (2016) performed some basic validation studies using simplistic static and dynamic exertions. A detailed and more comprehensive validation of the strain index is still lacking. It is critical that strain index is validated using a range of physical exertions to substantiate it as an appropriate risk assessment tool. In this study 40 forceful arm exertions were used to assess (1) sensitivity of the strain index to work-related factors (such as shoulder angle, shoulder plane, external force level and exertion direction); (2) the relationship of model-based strain index with physiological muscle activity data.

The strain index was found to be quite sensitive to the various work-related factors. Statistical significance was observed for the effect of shoulder plane, shoulder angle, external force and exertion direction on strain index. The mean strain index values ranged from 0.76 to 44.77 (Figure 6.1). Within the observed range, higher strain index values (25 to 44) were observed for the exertions performed in the pull up and right directions in sagittal and scapular planes, at 60°

and 120° shoulder angles using 2.5 and 5lb external loads. Moderate strain index values (15 to 25) were observed for the exertions performed in the pull back, left, right and up directions in sagittal and scapular planes, at 60° and 120° shoulder angles using 2.5 and 5lb external loads. The strain index values remained quite low (0.77 to 15) for the exertions performed in the pull back, down left and right directions in sagittal and scapular planes, at 60° and 120° shoulder angles using 2.5 and 5lb external loads.



**Figure 6.1 Shoulder strain index values observed in this study**

Higher strain was observed with an increase in the magnitude of the force. This was quite expected and consistent with previous studies. Antony et al., (2010) observed shoulder exertions with three hand load conditions (no load, 0.5kg and 30% grip) in three shoulder planes and four elevation angles. They found that an addition of 0.5kg hand load increased the shoulder muscle activity by 4% of MVC. For automotive assembly operations performed at different combinations of workpiece weights (3lb and 6lb) and hand-tool weights (1lb, 2lb and 4lb), shoulder postures and duration, Garg et al., (2006) found that ratings of perceived exertion, fatigue and pain increased with an increase in the weight of workpiece and hand-tool. Piscione et al., (2006) reported the

effect of backpack carrying load (0kg, 10 kg and 20 kg) on fatigue of middle deltoid and upper trapezius muscles. A significant increase in fatigability was observed in both muscles with increasing the mass of load, which was similar to the observations of the present study. However, the increase in the strain index values with an increase in the magnitude of force was small. This is mainly attributed to the small force levels (2.5lb and 5 lb) used in this study. These force levels are very small in comparison with the maximum controllable force by humans which ranges from 15.5 Newtons (3.49 lb) to 102.3 Newtons (23lb) (Tan et al., 1994).

Higher strain index was observed at 120° shoulder angle compared to 60° shoulder angle, which is consistent with previous studies (Antony et al., 2010; Sigholm et al., 1983; Järvholm et al., 1991). Brookham et al., (2010) investigated the impact of shoulder joint flexion (0°, 60°, 70°, 80°, 90°) and humeral rotation on shoulder muscle activation while exerting a light tool, and reported a positive relationship between shoulder flexion angle and shoulder muscle activity. Straker et al., (1997) investigated the effect of posture on workers' performance. The study was conducted at two shoulder joint flexion angles (0° and 30°) and two noise levels. Their findings suggested that shoulder joint flexed at 30° was associated with poor performance, high discomfort ratings, and higher fatigue compared to shoulder joint flexed at 0°. Overall, the current and the previous studies seem to indicate that shoulder strain increases with an increase in the shoulder angle.

A comparison between shoulder planes demonstrated that the strain index in the sagittal plane was significantly higher than in the scapular plane. Multiple previous studies have reported similar trends (Antony and Keir, 2010; Kai et al., 2012; Minning et al., 2007). Kai et al., (2012) compared the amount of fatigue on infraspinatus muscle during resisted arm elevation and reported that arm elevation in the sagittal plane resulted in higher amount of fatigue than scapular plane. In a study that analyzed the activation of scapular stabilizing muscles during isometric tasks, Oliveira et al., (2013) stated that the ratios between upper trapezius/lower trapezius and upper trapezius/serratus anterior were higher in the frontal plane compared to the scapular plane. Thus, the findings suggest that the shoulder strain is higher in sagittal plane than scapular plane.

When comparing the direction of force exertions, the strain index values were the highest in pull up direction, followed by pull left, pull right, pull back and pull down directions. Similar results were found in the study conducted by Alasim et al., (2019). In their study, the authors

observed the effects of pulling and pushing exertions in five directions (pull right, pull left, pull back, pull down and pull up) using three force levels. They found that exertions performed in up and right directions generated higher muscle activation and higher ratings of perceived exertion than the exertions performed in other directions. On the other hand, while observing the effects of multiple work locations and six directions (forward, backward, upward, downward, left and right) of force application (40N) on shoulder muscular activity, Cudlip et al., (2016) reported that backward exertions generated the highest mean overall muscle activity. Similar results were found in a study by Chopp et al., (2010) where they estimated the muscle load at two work configurations, four different angles with vertical axis, and five directions (backwards, forwards, downwards, sideways and upwards). They concluded that pulling in backwards direction generated the greater muscular demand than all other force directions. These observations contradicted with the findings of the present study as in this study the highest strain index values were found in the pull up direction. The differences in the study findings could be attributed to the experimental design related factors. For example, the external forces used in the current study were lower than some of the previous studies. This lower level of force might not have any significant effect on pull back directions. The findings related to the exertions performed in left, right and down directions are consistent with a study performed by McDonald et al., (2014). Pull down tasks in this study generated the least strain index values. Similar results were found in the studies conducted by Haslegrave et al., (1997) and Cudlip et al., (2016). These studies suggested that due to the assistance of gravitational force, pull down tasks were easier than the tasks in the other directions.

The correlation analysis drew a clear picture of the relationship between the model-based strain index and physiological loading of the shoulder joint. Previous studies have used correlation analysis to establish relationship between model driven data and physiological data. One such analysis was found in the study of Dickerson et al., (2006) where the external dynamic shoulder torque was calculated using an upper extremity biomechanical model and perceived shoulder muscular effort was rated by individual subject. The correlation analysis between these factors demonstrated that individual subject torque values were positively correlated with perceived effort scores ( $r^2 = 0.45-0.77$ ). Laursen et al., (1998) applied correlation analysis for validating an EMG based model used to estimate muscle forces at glenohumeral joint. They concluded that the correlation values between the shoulder joint moment and external force moment were highest at abduction/adduction moment (0.65 to 0.95) and flexion/extension moment (0.7 to 0.9).



Additionally, several studies have implemented correlation analysis for assessing the relationship between physiological data of muscle activation and output of biomechanical model (Dickersin et al., 2008; Hawkes et al., 2012; Alasim et al., 2019). Generally, correlation values within the range of 0 to 0.3 indicates weak positive correlation, 0.3 to 0.7 indicates moderate positive correlation, and 0.7 to 1.0 indicates strong positive correlation (Ratner, 2009). From the correlation analysis of this study, it was found that the shoulder strain index demonstrated a strong positive correlation with infraspinatus (0.760), while showed moderate correlations with anterior deltoid (0.672), biceps (0.637), middle deltoid (0.556) and supraspinatus (0.546). This result suggests that one rotator cuff muscle (infraspinatus) shows high relation and other (supraspinatus) shows moderate relation with the strain index values. The non-rotator cuff muscles exhibit moderate to low values of correlation. The result is consistent with the trend of higher activation values of rotator cuff muscles than other shoulder muscles during forceful arm exertions. Previous studies have suggested that the infraspinatus muscles show a high activation level during manual handling tasks (Goubault et al., 2021). In a study conducted by Alasim et al., (2019) it was found that rotator cuff muscles (infraspinatus and supraspinatus) and middle deltoid muscles showed higher activation than all other muscles during strenuous tasks. Sigholm et al., (1984) and Sporrang et al., (1996) also reported higher activation of infraspinatus and supraspinatus muscle during pull up and pull right exertions. The reason for higher activation level of rotator cuff muscles is that the forces produced by these muscles significantly contribute to the concavity compression process of glenohumeral joint stabilization (Wuelker et al., 1998).

Though in our study a strong correlation value was observed for the infraspinatus muscle, the supraspinatus muscle showed a moderate relationship with the strain index values. Additionally, the anterior deltoid muscles generated a better correlation value than supraspinatus muscles. This type of different muscle activation patterns were also found in the several studies. Cudlip et al., (2016) found that in overhead exertions anterior deltoid muscle had higher muscle activity (activity level >50%MVE) than supraspinatus muscle (activity level>45% MVE). Chopp et al., (2010) found that anterior deltoid muscles showed higher activation level compared to infraspinatus muscles at different combinations of work configuration, target angle and force direction. This different pattern of muscle activations could be explained by the fact that different configuration and combination of work-related factors had different effects on the activation level of shoulder muscles, thus they exhibited different pattern from general trend.

The lowest correlation value was observed between strain index values and triceps muscle (0.116). Similar pattern of muscle activation of triceps was observed in Alasim et al., (2016) study. The low activation of triceps muscles indicate that these muscles do not actively participate in concavity compression process of shoulder joint.

A high correlation coefficient of almost 0.7 was observed when activities of shoulder muscles were averaged and correlated with the strain index, further suggesting that shoulder strain index developed by Chowdhury et al. (2016) shows a strong relationship with the overall physiological demand of the shoulder joint. Such strong relationship validates the strain index as an appropriate tool to assess the risk of injury to the shoulder joint due to static work-related forceful arm exertions.

## **6.1 Study Limitations and Direction for Future Studies**

In this validation study, only static exertions were used. Workplace exertions are static as well as dynamic. In order to fully validate the strain index, future study can evaluate dynamic exertions. The force values used in this study were low (2.5 and 5 lbs.) considering the much higher force exertion ability of the shoulder joint. The low force values limited the strain index in a rather narrow range of <44, future studies can look at higher force values to test and validate shoulder strain index in higher ranges (50-75, 75 to 100, etc.). However, there is risk of creating higher strain and/or injury during the exertions with high strain index (especially >75) and therefore it will be challenging to design such studies. The effect of age of the participants was not considered in this validation study. In workplace settings there are workers from various age groups. Additionally, the total number of participants hired in this study was only 12. Future studies can include more participants from multiple age groups and consider the effect of age on strain index values. However, most of the existing biomechanical modelling systems including AnyBody do not have options to incorporate age as a factor in the biomechanical models. In this study, a pre-existing shoulder model from the repository of the AnyBody modelling system was used. All models are subjected to some levels of assumptions, future studies can consider other shoulder models (such as Open Sim) to further validate the strain index. In this study the models were scaled using basic participant data such as gender, height, and weight. There are newer modelling options

for creating more accurate participant specific models by incorporating more precise arm anthropometric measurement. Future studies can explore such participant-specific modelling approaches. The validation was performed using dominant (right) hand, future studies can focus on non-dominant hand.

## **6.2 Final Conclusion**

The study results indicate that the shoulder strain index developed by Chowdhury et al. (2016) is sensitive to the work-related factors that are typical during occupational forceful arm exertions. The study results also indicate that the shoulder strain index exhibits a strong relationship with the real physiological data. Thus, based on the findings of this study, it can be concluded that the shoulder strain index can be used as a risk assessment tool to accurately compare and predict the risk of shoulder WMSDs during work related static forceful arm exertions.

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## Appendix I: Shoulder Strain Index

Table A1: Shoulder Strain Index

Shoulder angle	Shoulder plane	load	Direction	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 9	Subject 10	Subject 11	Subject 12	Average
60	Sagittal	2.5	PB	13.99 27 12 52	14.11 60 87 28	16.81 08 65 23	14.47 81 40 09	14.82 85 72 24	14.35 87 80 16	17.71 42 33 95	14.94 28 87 33	12.10 89 25 16	15.27 86 66 78	12.53 89 76 37	13.87 93 99 38	14.58735387
60	Sagittal	2.5	PD	0.76 23 96 12	0.937 09 02 97	4.475 71 66 75	1.421 16 22 26	1.877 56 91 83	1.277 06 18 43	5.663 93 96 3	2.025 50 21 99	2.298 84 11 99	2.475 44 48 87	1.789 35 74 31	0.845 00 74 53	2.15432772
60	Sagittal	2.5	PL	17.58 79 43 94	17.64 07 07 51	19.22 88 56 89	17.82 75 60 15	18.01 58 02 44	17.76 95 53 53	19.95 42 13 5	18.07 57 18 57	16.27 05 94 37	18.24 91 48 79	16.66 01 90 22	17.61 37 49 13	17.90722825
60	Sagittal	2.5	PR	12.74 01 52 65	12.85 43 23 86	15.34 80 19 78	13.18 81 79 68	13.50 97 27 48	13.07 83 03 17	16.17 71 70 5	13.61 42 98 98	10.96 54 28 93	13.92 05 44 93	11.37 70 46 73	12.63 36 24 63	13.28390178
60	Sagittal	2.5	PU	26.52 56 33 05	26.58 06 38 27	27.77 63 74 15	26.74 18 85 26	26.89 77 65 97	26.68 87 49 36	28.17 64 33 28	26.94 85 75 61	25.71 72 80 23	27.09 77 12 04	25.89 94 11 59	26.61 67 68 42	26.80560227
60	Sagittal	5	PB	15.02 83 53 34	15.14 69 27 03	17.73 19 67 6	15.49 47 45 31	15.83 12 33 71	15.38 00 95 92	18.59 69 34 34	15.94 09 63 83	13.21 47 53 49	16.26 31 86 46	13.62 93 71 61	14.90 84 42 51	15.59724793
60	Sagittal	5	PD	9.205 98 67 24	9.008 92 72 99	4.805 84 28 56	8.428 95 59 39	7.862 45 13 37	8.623 08 63 5	3.544 78 94 52	7.683 84 16 34	12.23 01 41 25	7.149 40 06 72	11.54 81 49 87	9.405 55 15 54	8.291427079
60	Sagittal	5	PL	22.39 38 94 74	22.48 52 92 48	24.39 31 02 03	22.74 66 64 06	22.99 50 64 02	22.66 12 33 42	25.01 85 31 41	23.07 55 42 47	21.20 29 58 86	23.31 06 85 86	21.47 11 44 07	22.29 92 47 31	22.83778006
60	Sagittal	5	PR	13.64 56	13.73 97	15.98 53	14.01 42	14.27 75	13.92 39	16.85 27	14.36 16	12.19 69	14.60 74	12.53 14	13.55 03	14.14058657

				60 47	17 69	43 55	71 78	47 34	98 47	21 72	49 89	73 11	13 77	23 86	17 22	
60	Sagittal	5	PU	40. 22 74 23 96	40. 20 48 78 09	39. 70 79 89 55	40. 13 86 05 39	40. 07 44 18 19	40. 16 04 62 6	39. 53 85 92 13	40. 05 34 52 68	40. 57 10 07 12	39. 99 18 68 08	40. 49 27 17 93	40. 25 02 37 53	40.11763 777
60	Scapular	2.5	PB	8.1 93 35 79 49	8.3 16 95 84 31	11. 13 10 18 6	8.6 74 99 83 79	9.0 15 41 35 45	8.5 57 81 07 42	12. 24 13 51 49	9.1 25 71 38 63	6.4 51 92 26 85	9.4 67 16 27 9	6.7 97 28 11 99	8.0 67 60 53 3	8.836716 251
60	Scapular	2.5	PD	1.0 75 22 40 96	1.1 11 01 57 82	5.0 40 56 82 19	1.6 25 09 11 94	2.1 30 98 03 16	1.4 53 65 49 82	6.3 91 10 69 72	2.2 96 74 79 46	1.7 00 22 43 22	2.7 89 68 56 65	1.5 85 34 79 94	1.1 38 91 49 97	2.361546 874
60	Scapular	2.5	PL	13. 87 97 16 42	13. 99 39 33 3	16. 55 68 11 8	14. 32 23 28 22	14. 64 83 27 3	14. 21 54 81 29	17. 30 37 02 72	14. 77 09 80 13	12. 01 48 66 45	15. 12 07 65 32	12. 44 91 86 51	13. 76 34 65 57	14.41996 375
60	Scapular	2.5	PR	13. 63 64 14 77	13. 73 44 00 14	15. 95 91 26 59	14. 01 95 92 52	14. 29 23 90 32	13. 92 59 51 2	16. 85 47 85 91	14. 38 07 08 95	12. 09 04 06 39	14. 63 81 99 28	12. 45 06 12 95	13. 53 69 28 98	14.12662 65
60	Scapular	2.5	PU	25. 78 16 29 8	25. 83 72 70 73	27. 04 86 59 54	26. 00 05 27 91	26. 15 82 10 38	25. 94 67 12 24	27. 45 47 08 04	26. 20 95 12 85	24. 93 21 04 06	26. 36 01 82 39	25. 12 60 21 34	25. 72 53 69 41	26.04840 906
60	Scapular	5	PB	9.1 37 13 03 53	9.0 88 69 71 34	9.4 47 92 66 51	8.9 02 93 77 78	8.6 38 22 53 92	8.9 72 12 58 64	9.8 92 42 96 3	8.5 22 87 71 75	9.2 73 25 60 1	8.6 57 91 84 26	9.3 27 42 14 26	9.1 80 69 41 29	9.086803 331
60	Scapular	5	PD	5.0 88 21 10 02	5.0 07 47 83 28	3.2 84 90 88 31	4.7 80 60 33 69	4.5 67 23 95 63	4.8 53 19 38 39	2.6 27 43 10 74	4.4 97 50 80 72	6.2 32 29 51 46	4.2 92 28 74 89	5.9 74 87 76 93	5.1 52 35 81 25	4.696532 711
60	Scapular	5	PL	15. 14 00 69 04	15. 24 79 29 07	17. 61 37 69 47	15. 56 24 73 05	15. 86 76 64 44	15. 45 72 84 27	18. 51 22 95 42	15. 96 57 19 14	13. 64 54 23 76	16. 24 37 89 8	13. 98 04 81 97	15. 03 04 09 54	15.68894 241
60	Scapular	5	PR	17. 38 60 39 75	17. 37 45 08 27	18. 20 95 06 87	17. 32 80 72 11	17. 29 42 97 33	17. 34 55 76 16	18. 60 37 93 08	17. 33 39 14 24	17. 11 74 93 57	17. 43 19 91 32	17. 20 98 32 74	17. 39 56 74 33	17.50255 831
60	Scapular	5	PU	38. 83 02	38. 81 43	38. 46 50	38. 76 75	38. 72 21	38. 78 29	38. 34 74	38. 70 73	39. 07 31	38. 66 38	39. 01 78	38. 84 63	38.75320 428

				76 13	41 58	41 04	18 66	79 1	59 82	93 8	72 57	70 58	85 18	13 77	99 1	
120	Sagittal	2.5	PB	15. 79 22 39 53	15. 94 77 85 88	19. 19 94 66 03	16. 37 92 50 9	16. 80 28 09 28	16. 23 49 01 25	20. 29 13 69 84	16. 94 10 33 43	13. 51 56 66 17	17. 34 69 79 64	14. 03 51 91 34	15. 64 14 06 57	16.51067 499
120	Sagittal	2.5	PD	2.4 36 20 32 6	2.5 08 86 34 29	5.9 73 23 01 93	3.0 84 40 64 78	3.5 44 24 96 86	2.9 26 63 34 65	7.5 12 63 29 66	3.7 07 64 88 87	2.0 94 37 24 08	4.1 44 37 46 48	1.7 67 15 50 96	2.2 50 66 55 69	3.495869 674
120	Sagittal	2.5	PL	16. 59 05 56 67	16. 72 79 41 81	18. 18 24 62 37	16. 87 31 26 14	17. 07 97 57 48	16. 80 43 87 64	18. 88 11 31 92	17. 14 66 82 69	15. 45 02 00 84	17. 34 23 59 94	15. 71 34 71 33	16. 51 69 90 57	16.94242 245
120	Sagittal	2.5	PR	20. 39 62 04 52	20. 57 49 82 3	22. 95 43 51 05	20. 85 27 07 33	21. 17 19 36 28	20. 74 19 35 18	23. 82 90 73 78	21. 27 30 95 09	18. 46 03 60 4	21. 57 40 25 89	18. 92 19 15 44	20. 27 36 90 26	20.91868 979
120	Sagittal	2.5	PU	29. 91 54 96 49	30. 14 45 57 16	31. 54 19 90 71	30. 19 54 94 43	30. 39 76 30 46	30. 12 66 37 18	32. 06 37 05 55	30. 46 36 42 99	28. 82 97 10 95	30. 65 74 17 1	29. 07 74 41 51	29. 84 35 50 29	30.27143 957
120	Sagittal	5	PB	16. 38 54 76 62	16. 53 17 12 84	19. 72 19 75 73	16. 96 04 26 28	17. 37 52 90 61	16. 81 91 23 82	20. 79 12 78 26	17. 51 06 25 16	14. 15 40 04 2	17. 90 81 37 41	14. 66 32 71 72	16. 23 76 27 87	17.08824 588
120	Sagittal	5	PD	7.4 98 14 50 46	7.3 59 77 50 04	4.1 66 93 13 27	6.9 35 34 79 14	6.4 95 87 66 93	7.0 79 69 47 91	3.1 26 10 65 9	6.3 58 14 12 51	9.7 53 83 80 08	5.9 77 73 75 14	9.3 18 73 84 67	7.6 58 33 13 88	6.810721 999
120	Sagittal	5	PL	22. 28 91 99 25	22. 34 20 52 89	23. 42 06 98 46	22. 49 43 49 64	22. 63 88 76 54	22. 44 45 14 31	23. 76 86 98 51	22. 68 54 81 82	21. 80 50 68 96	22. 82 08 04 34	21. 89 44 72 99	22. 23 52 19 01	22.56995 306
120	Sagittal	5	PR	27. 50 96 75 31	27. 64 55 00 2	30. 60 67 54 9	28. 03 67 15 88	28. 40 71 16 05	27. 90 88 34 9	31. 40 01 44 57	28. 52 65 39 73	26. 65 64 74 81	28. 87 40 12 08	26. 74 52 04 55	27. 37 09 54 25	28.30732 727
120	Sagittal	5	PU	44. 62 85 51 43	44. 61 89 02 41	44. 40 70 93 88	44. 59 05 46 22	44. 56 31 27 96	44. 59 98 95 17	44. 33 55 10 29	44. 55 41 70 33	44. 77 58 87 24	44. 52 78 85 08	44. 74 22 66 5	44. 63 83 19 25	44.58184 631
120	Scapular	2.5	PB	14. 18 20	14. 20 73	14. 70 82	14. 28 20	14. 35 33	14. 25 74	14. 83 80	14. 37 63	13. 73 41	14. 44 22	13. 84 67	14. 15 70	14.28209 628

				86 89	90 09	45 34	36 41	81 28	71 68	09 74	17 79	30 37	96 41	65 05	24 37	
120	Scap ular	2.5	PD	1.8 22 15 22 93	1.8 24 26 75 89	2.9 29 36 16 59	1.8 45 47 28 81	1.8 67 08 89 99	1.8 41 75 48 26	3.8 11 81 00 78	1.8 75 81 74 59	1.5 58 83 93	1.9 06 10 89 67	1.6 61 19 89 2	1.8 35 13 89 33	2.064917 659
120	Scap ular	2.5	PL	11. 01 67 29 43	11. 04 73 20 72	11. 53 06 98 5	11. 13 47 26 28	11. 21 58 29 61	11. 10 63 05 47	11. 73 34 26 99	11. 24 15 44 17	10. 50 75 04 88	11. 31 52 21 29	10. 62 95 72 64	10. 98 54 13 6	11.12202 446
120	Scap ular	2.5	PR	10. 47 18 50 01	10. 56 10 33 5	12. 10 52 75 99	10. 80 97 27 77	11. 03 20 43 87	10. 73 02 57 44	12. 49 39 12 69	11. 10 16 97 19	9.6 64 13 20 97	11. 29 78 04 28	9.7 23 87 10 17	10. 38 54 26 19	10.86475 267
120	Scap ular	2.5	PU	16. 02 01 90 38	16. 05 36 06 91	16. 78 46 60 5	16. 15 16 90 49	16. 24 66 74 61	16. 11 93 49 39	17. 03 02 17 93	16. 27 76 61 39	15. 51 06 26 18	16. 36 87 22 75	15. 62 68 42 89	15. 98 64 08 05	16.18138 762
120	Scap ular	5	PB	20. 94 35 59 51	20. 95 66 67 44	21. 15 22 58 3	20. 99 08 62 65	21. 04 04 46 2	20. 97 99 76 31	21. 10 55 07 38	21. 05 47 36 94	20. 61 83 98 49	21. 09 33 00 52	20. 71 62 95 11	20. 92 94 55 75	20.96512 205
120	Scap ular	5	PD	5.3 65 95 92 9	5.1 99 81 02 59	2.5 91 28 79 73	4.7 09 69 09 75	4.2 35 09 28 37	4.8 71 76 11 55	2.3 63 21 54 24	4.0 95 21 86 11	8.0 35 12 47 69	3.6 81 28 54 63	7.4 23 53 01 61	5.5 33 62 26 49	4.842133 297
120	Scap ular	5	PL	15. 78 35 19 75	15. 79 78 89 84	16. 07 29 25 02	15. 83 94 13 68	15. 87 84 24 3	15. 82 58 51 5	16. 14 41 38 48	15. 89 08 76 36	15. 52 41 40 59	15. 92 66 92 56	15. 59 21 61 62	15. 76 87 24 01	15.83706 314
120	Scap ular	5	PR	15. 42 74 32 76	15. 39 00 37 71	14. 90 47 72 29	15. 27 14 81 37	15. 14 50 04 25	15. 31 19 84 44	14. 83 65 59 95	15. 10 08 94 11	16. 72 36 87 5	14. 96 38 57 06	16. 35 16 02 08	15. 46 36 46 01	15.40757 996
120	Scap ular	5	PU	24. 19 27 24 18	24. 18 10 70 89	23. 92 66 43 61	24. 14 68 49 29	24. 11 37 42 74	24. 15 81 30 99	23. 84 15 35 32	24. 10 29 45 4	24. 37 07 48 16	24. 07 12 32 97	24. 33 01 18 51	24. 20 45 20 65	24.13668 856

## Appendix II: RMS Values of Shoulder Muscles

Table A2: RMS values of seven shoulder muscles

Angle	Plane	Weight (lbs)	Direction	Supraspinatus	Infra spinatus	Middle deltoid	Anterior deltoid	Posterior deltoid	Biceps	Triceps	Average Muscle activity
60	90	2.5	PB	0.0846	0.08488	0.05314	0.11844	0.022646	0.02554	0.09365	0.068985143
60	90	2.5	PD	0.03426	0.06479	0.03138	0.07868	0.020324	0.019343	0.13873	0.055358143
60	90	2.5	PL	0.03074	0.11002	0.07211	0.1881	0.02833	0.18756	0.11166	0.104074286
60	90	2.5	PR	0.1319	0.19214	0.10484	0.13505	0.06407	0.06773	0.117	0.116104286
60	90	2.5	PU	0.14214	0.22799	0.12607	0.21952	0.05701	0.16009	0.1169	0.14996
60	90	5	PB	0.06891	0.07425	0.02852	0.05411	0.020315	0.02197	0.14335	0.058775
60	90	5	PD	0.021456	0.0563	0.016867	0.015795	0.0283	0.019099	0.25419	0.058858143
60	90	5	PL	0.03231	0.10611	0.06942	0.19869	0.03256	0.25836	0.13809	0.119362857
60	90	5	PR	0.18678	0.3103	0.14172	0.11039	0.11294	0.10436	0.18392	0.164344286
60	90	5	PU	0.18546	0.34914	0.169	0.27039	0.08785	0.2358	0.15093	0.206938571
60	45	2.5	PB	0.136	0.09952	0.06506	0.06837	0.03552	0.02749	0.10571	0.07681
60	45	2.5	PD	0.09325	0.07172	0.03522	0.04316	0.023492	0.017794	0.16035	0.063569429
60	45	2.5	PL	0.10158	0.10757	0.0845	0.15548	0.03245	0.144	0.09135	0.102418571
60	45	2.5	PR	0.20517	0.21156	0.13015	0.07231	0.13037	0.05872	0.16052	0.1384
60	45	2.5	PU	0.23064	0.18561	0.15358	0.18146	0.0816	0.15614	0.11204	0.157295714
60	45	5	PB	0.13136	0.10319	0.03588	0.02113	0.03023	0.020405	0.15455	0.070963571
60	45	5	PD	0.03507	0.06314	0.019634	0.01462	0.024894	0.020799	0.26568	0.063405286
60	45	5	PL	0.08538	0.12615	0.08702	0.18872	0.0335	0.23891	0.13207	0.127392857
60	45	5	PR	0.28715	0.3623	0.18051	0.08795	0.21591	0.09098	0.26354	0.21262
60	45	5	PU	0.31493	0.28626	0.22781	0.2462	0.13938	0.23702	0.15896	0.23008
120°	45	2.5	PB	0.20371	0.19011	0.21111	0.26754	0.10966	0.10515	0.11228	0.171365714

120	90	2.5	PD	0.06274	0.067 15	0.091 43	0.151 37	0.038 32	0.029 42	0.109 73	0.078594286
120°	90	2.5	PL	0.17856	0.162 47	0.195 23	0.270 55	0.085 57	0.234 21	0.124 05	0.178662857
120°	90	2.5	PR	0.17616	0.209 91	0.216 08	0.125 88	0.176 95	0.065 29	0.215 18	0.16935
120	90	2.5	PU	0.23662	0.245 7	0.252 08	0.287 38	0.147 39	0.173 86	0.142 02	0.21215
120°	90	5	PB	0.25083	0.200 58	0.246 3	0.282 25	0.143 51	0.128 57	0.138 97	0.198715714
120°	90	5	PD	0.03694	0.061 23	0.021 56	0.040 08	0.018 721	0.022	0.136 18	0.048101571
120	90	5	PL	0.19095	0.208 7	0.201	0.307 96	0.100 63	0.385 5	0.188 86	0.226228571
120°	90	5	PR	0.18361	0.271 31	0.234 75	0.094 7	0.274 27	0.068 25	0.308 4	0.205041429
120°	90	5	PU	0.31586	0.372 29	0.334 08	0.375 93	0.235 15	0.253 19	0.219 96	0.300922857
120°	90	2.5	PB	0.25248	0.148 01	0.235 97	0.232 36	0.140 97	0.125 48	0.129 52	0.180684286
120	45	2.5	PD	0.12207	0.080 13	0.110 09	0.113 75	0.058 41	0.037 64	0.126 96	0.092721429
120°	45	2.5	PL	0.19235	0.121 17	0.185 94	0.226 48	0.089 05	0.196 34	0.113 83	0.160737143
120°	45	2.5	PR	0.25544	0.182 14	0.287 06	0.132 78	0.263 76	0.084 85	0.220 87	0.203842857
120	45	2.5	PU	0.29187	0.182 07	0.311 16	0.291 51	0.212 82	0.203 28	0.164 55	0.236751429
120°	45	5	PB	0.27526	0.162 65	0.256 08	0.223 6	0.157 7	0.140 74	0.141 93	0.193994286
120°	45	5	PD	0.06848	0.073 71	0.034 86	0.029 21	0.027 76	0.021 5	0.168 13	0.060521429
120	45	5	PL	0.17987	0.135 35	0.156	0.245 2	0.064 97	0.303 4	0.132 97	0.173965714
120°	45	5	PR	0.32512	0.292 92	0.364 28	0.126 9	0.434 6	0.101 52	0.367 5	0.287548571
120°	45	5	PU	0.4037	0.298 84	0.385 9	0.347 72	0.276 6	0.298 39	0.213 28	0.317775714

<sup>i</sup> <https://www.sports-health.com/sports-injuries/shoulder-injuries/guide-shoulder-anatomy>

<sup>ii</sup> <https://physio-study.com/biomechanics-of-the-shoulder/>

<sup>iii</sup> [https://www.researchgate.net/figure/a-Position-of-the-humeral-head-with-respect-to-the-glenoid-cavity-b-Concavity\\_fig3\\_271215335](https://www.researchgate.net/figure/a-Position-of-the-humeral-head-with-respect-to-the-glenoid-cavity-b-Concavity_fig3_271215335)