An EMG Analysis of the Muscles Involved in the Incline and the Supine Bench Press

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By

Sylvain J. Lemelin

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AN EMG ANALYSIS OF THE MUSCLES INVOLVED IN THE INCLINE AND THE SUPINE BENCH PRESS

BY

SYLVAIN J. LEMELIN

A Thesis submitted to the Faculty of Graduate Studies of the University of Manitoba in partial fulfillment of the requirements of the degree of

MASTER OF SCIENCE

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ABSTRACT

An EMG Analysis of the Muscles Involved in the Incline and Supine Bench Press

One of the most commonly used exercises for testing strength and power of athletes is the bench press. The bench press is also considered to be one of the most fundamental exercises for developing and strengthening the muscles of the anterior shoulder, namely the pectoralis major and anterior deltoid. The incline bench press is another popular chest training exercise used by many athletes. Weight training dogma suggests that the incline bench press can specifically increase the activation level of the clavicular fibers of the pectoralis major. The purpose of the study was to compare the effect of four imposed bench press conditions (i.e., supine, 37°, 43°, and 50° incline bench presses) on the normalized EMG levels of sternal pectoralis major, clavicular pectoralis major, anterior deltoid, and middle deltoid muscles. Data was collected on 14 male subjects while performing four different bench press exercises (i.e., supine, 37°, 43°, and 50° incline), using a 75% RM load which had been determined from four separate 1 RM tests prior to data collection. Analysis of the normalized EMG data determined that the level of activation of the clavicular pectoralis major did not significantly change across all four conditions. Activation of the sternal pectoralis muscle during the supine bench press was significantly greater than during the 43° and 50° incline, but not during the 37° incline bench press. Both the middle and anterior deltoid muscles had significantly greater levels of activation during the three incline bench press exercises than during the supine bench press. Mean ascent and descent times were consistent across all bench press trials. Horizontal displacement of the bar across all trials did not change. Vertical bar displacement was significantly greater during the 43° and 50° incline than the supine bench press. Peak vertical velocity during the eccentric and concentric phases were greater in the three incline bench presses than the supine bench press. Peak downward acceleration during the 43° incline bench press was significantly greater than the supine, yet no different than the 37° and 50° incline. Finally, peak upward acceleration in the eccentric and concentric phases did not change significantly. The results suggest that the incline bench press does not selectively recruit clavicular pectoralis major; and the activity

of this muscle is consistent over the four types of exercise. It appears that the sternal pectoralis major plays ann important role during the supine bench press while the anterior and middle deltoid muscles play a more important role during the incline bench press.

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"AN EMG ANALYSIS OF THE MUSCLES INVOLVED IN THE INCLINE AND THE SUPINE BENCH PRESS"

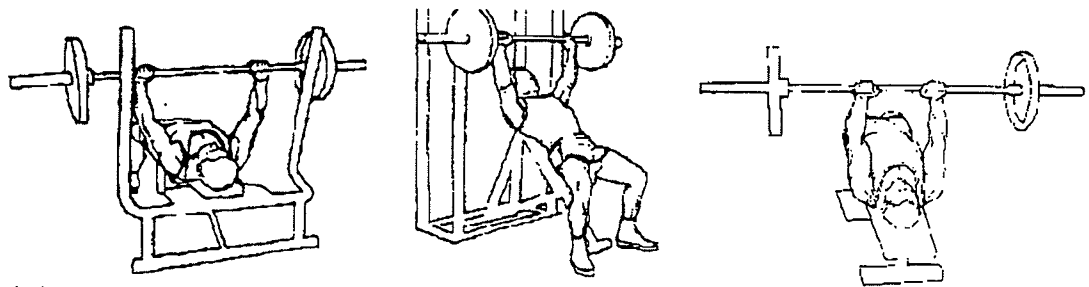
CHAPTER 1

INTRODUCTION

One of the most commonly used exercises for testing and training of strength and power of athletes and bodybuilders is the bench press. The bench press is also considered to be one of the most fundamental exercises for developing and strengthening the muscles of the anterior shoulder, namely the pectoralis major and anterior deltoid (Fleck and Kraemer, 1987). The movement involves lowering a weight from an extended elbow position while the athlete is lying supine on a bench. Once the bar touches the chest (horizontal extension of the humerus and flexion of the elbow), the weight is lifted or pushed vertically upward to its original starting position by horizontally flexing the upper arm and extending the elbows.

The traditional exercise bench was a versatile piece of equipment that inspired the creation of other equipment designed to provide a greater variety of exercises for the chest. Examples of variations in the equipment are the incline bench press and the decline bench press (Figure 1-1). The use of these alternative bench press modes in conjunction with the traditional bench press have been incorporated into the weight lifting programs of athletes to provide what is believed to be a more "complete chest workout" (Reynolds, 1991).

Weight training dogma indicates that these alternative modes of the bench press elicit specific training adaptations (i.e., increase strength and development in distinct regions of the pectoralis muscle). Weight training manuals, books, and popular magazines (Baechle, Earle, Allerheiligen, 1994; Bijarnson, 1986; Moran & McGlynn, 1990; O'Shea, 1976; Pearl & Moran, 1986; Reynolds, 1991) suggest that the movements performed on the incline bench press would increase the activation level of the clavicular pectoralis via: (1) an increase in discharge rate per activated motor unit, or (2) an increase number of motor units in the clavicular portion (i.e., upper part of the chest) of the pectoralis muscle, which in turn would enhance strength and size in that area.



(a) (b) (c)

Figure 1-1 Alternate modes of the bench used to train the chest. **(a)** Supine bench press; **(b)** incline bench press; and **(c)** decline bench press. (From Pearl & Moran, 1986, p. 245).

The assumption of increased EMG from the clavicular pectoralis during the incline bench press has been explained by describing the angle of the bench in relation to the horizontal and vertical planes, as well as what type of movement is required from the shoulder at each of these angles. The closer the angle of the bench was to the vertical, the greater emphasis placed on the deltoid; the closer to the horizontal, the more emphasis placed on the sternal pectoralis (Bijarnson, 1986). The movement required at the vertical (i.e., humeral abduction) is such that the deltoid muscle becomes the prime mover. Similarly, a bench angle at the horizontal requires horizontal flexion which is mainly performed by pectoralis major. Therefore, according to these authors, a bench press angle lying between the horizontal and vertical such as the incline bench press at 45 degrees would place an emphasis on the upper portion of the chest. Conversely, Reynolds (1991) described this angle as being too steep to elicit greater activation from the upper portion of the chest and recommended a 30-degree incline bench to increase the load on the clavicular portion of pectoralis major.

The reasons for the change in bench inclination (i.e., from supine to incline) have been poorly explained. The change in the incline may alter the orientation of the humerus (i.e. greater shoulder flexion and more external rotation) which in turn changes the load on the clavicular pectoralis, sternal pectoralis, and anterior deltoid, so that the clavicular pectoralis becomes a more important mover for this movement. This suggestion was acquired from the findings of Inman, Saunders, and Abbott (1944) who determined the clavicular pectoralis to be most active at 115 degrees (as measured from the anatomical position) of forward flexion of the shoulder . In order to keep the path of the bar close to vertical during this movement, the subject may be required to use an angle of forward shoulder flexion similar to the results of Inman and colleagues (1944), which in turn would elicit greater involvement from the motor units innervating the clavicular pectoralis muscle fibers as the angle of forward flexion for the arm is close to 115^{o}.

To lend more credibility this postulation, Manktelow et al., (1980), Chaffai and Mansat (1988), and Milroy et al., (1988), demonstrated that the pectoralis major muscle had independent neural and vascular components, suggesting that it has a potential for independent function. According to the the fact that the muscle is innervated by three seperate nerves, the incline bench press may then elicit greater activation from the clavicular pectoralis major.

Although the suggestion that the incline bench press actually imposes a greater training effect on the clavicular pectoralis muscle seems reasonable and has been accepted by the majority of the weight lifting community, to date no research has been conducted to support it. Acceptance of unproven hypotheses appears to be a problem in the field of body building where many of the practices have been established based on untested assumptions. These assumption are then presumed to be true by athletes and coaches. Examination of muscle activity using EMG techniques can assist in determining these muscle activation patterns.

Purpose of the Study

The purpose of the study was to determine if any differences exist between the normalized muscle activity in the sternal pectoralis, the clavicular pectoralis, the anterior deltoid, and the middle deltoid muscles during the supine bench press and the incline bench press.

Null Hypothesis

The normalized muscle activity in the sternal portion of the pectoralis, the clavicular portion of the pectoralis muscle, the anterior deltoid, and in the middle deltoid will not differ between the supine bench press and the incline bench press.

Rationale for the Study

The bench press and its variations are an integral part of a normal weight training session. Many coaches and athletes, especially bodybuilders, have planned their chest workouts around the assumption that the incline bench press elicits a greater training effect on the clavicular fibers of the pectoralis muscle than does the supine bench press. Determination of whether or not the incline bench press elicits greater activation of the clavicular portion of the chest would be useful information to the weight lifting community. The results of such a study could either change the weight lifting strategy in terms of chest workouts employed by athletes or it may strengthen the theory that the incline bench is an effective method of specifically training the clavicular pectoralis. Another important reason to conduct the study is that research will provide knowledge, as well as fill in a gap in understanding about this particular movement. Finally, the study may stimulate more questions and generate more research in exercise analysis and motor control theory, especially in the field of weight lifting.

Limitations and Delimitations

1. All subjects tested in the study were strength trained college males.

2. The experimental group was selected from a fairly stringent criteria: they had a minimum 2 years of weight training experience, and they were able to bench press at least 125% of their body weight. This limited the number of participants as well as decreased the generalizability of the study. These criteria were used to ensure consistency in terms of biomechanical technique as well as group homogeneity.

3. The testing protocol was designed to imitate parts of a typical workout session in order to make the results of the study more pertinent to the area.

4. The cadence was controlled by using a metronome to set the upward and doward phases of the exercise. Although this control may have reduced the generalizability, it was necessary to enhance reliability of the study.

5. No comparisons of raw EMG amplitude between the medial deltoid, anterior deltoid, sternal pectoralis major muscle, and the clavicular portion of pectoralis major muscle were attempted because according to Gowland , deBruin, Basmajian, Plews, and Burcea (1992), absolute EMG amplitude of muscle activity in these three different muscles would be influenced by variable distances of recording electrodes from active tissue.

Definition of Terms

acceleration phase In the research literature, it was defined as the portion of the ascent phase of the bench press exercise where the bar reached its peak acceleration and peak velocity (Madsen & McLaughlin, 1984). This occurred, as noted in the literature, between the start of the ascent and the onset of the sticking region.

EMG amplitude The quantity which expresses the level or the amount of signal activity; usually measured in microvolts (μV).

bipolar Having two detection surfaces; use of two electrodes.

concentric contraction Type of contraction in which the activated muscle is shortening under tension and the net moment is greater than the moment caused by the external forces acting on the segment to which the muscle is attached (Rodgers & Cavanagh, 1984).

concentric phase Term given during the ascent phase in which the prime mover muscles are contracting concentrically to produce the upward movement of the bar.

deceleration phase Portion of the ascent phase of the bar during the bench press exercise where the velocity of the bar is decreasing.

eccentric contraction Type of muscular contraction in which the muscle lengthens under tension because the net external moment acting on the segment is greater than moment developed by the muscle (Rodgers & Cavanagh, 1984).

eccentric phase Term given during the descent phase where the prime mover muscles are acting ecccentrically to control the downward movement of the bar.

electromyographic (EMG) signal This is the name given to the total signal detected by an electrode which is the algebraic sum of all active motor unit action potentials within the pick up area of the electrode (Basmajian & De Luca, 1985).

impedance The resistance to the transmission of an electrical current (Basmajian and De Luca 1985).

mechanical advantage The ratio of the effort-force lever arm to the resisting-force lever arm. It is the amount by which the effectiveness of an applied force is multiplied (Rodgers and Cavanagh, 1984).

moment A force acting away from the axis of rotation that produces angular motion. The magnitude is equal to the product of the force and the perpendicular distance from the line action of the force to the axis of rotation.

moment arm The perpendicular distance from the line of action of a force to the axis of rotation (Rodgers & Cavanagh, 1984, p. 1888).

monopolar Having one detection surface.

motor unit An individual motor nerve and all the muscle fibers it innervates (Fox, Bowers, & Foss, 1989, p. 699).

1 RM The maximal load lifted during the supine and incline bench press which corresponds to maximal effort. In the study, the 1 RM defined the strength of subjects.

muscle fiber action potential (MAP) The name given to the detected waveform resulting from the depolarization wave as it propagates in both directions along each muscle fiber from its motor end plate (Basmajian & DeLuca, 1985, p. 8).

myoelectric signal The total electrical signal observed at an electrode or differentially between two electrodes. It is also an alternate name given for the electromyographic signal (EMG) and muscle fiber action potential (Dainty and Norman, 1987).

noise Unwanted signal which is detected together with the wanted signal (Basmajian and De Luca, 1985).

prime mover A muscle considered to be a major contributor resposible for producing a specified movement .

trained In this study, a trained subject was defined as one who had a minimum two years of weight training experience and could bench press at least 125% of their body weight for one repetition.

sticking region The instant of minimum vertical upward bar acceleration, where the vertical forces exerted on the bar are minimal during the ascent phase. It occurrs as a result of a limitation in moment generation capacity where a relatively large moment arm of the load (about the elbow and shoulder axis). For maximal loads (100% RM), this position occurred between 33.9% (Madsen and McLaughlin, 1984) and 34.7% (Elliott, Wilson, and Kerr, 1989) of the vertical distance from the chest to peak height of the bar.

synergist a muscle that actively provides a contribution to a particular movement during a movement sequence (Basmajian & DeLuca, 1985).

CHAPTER 2

REVIEW OF LITERATURE

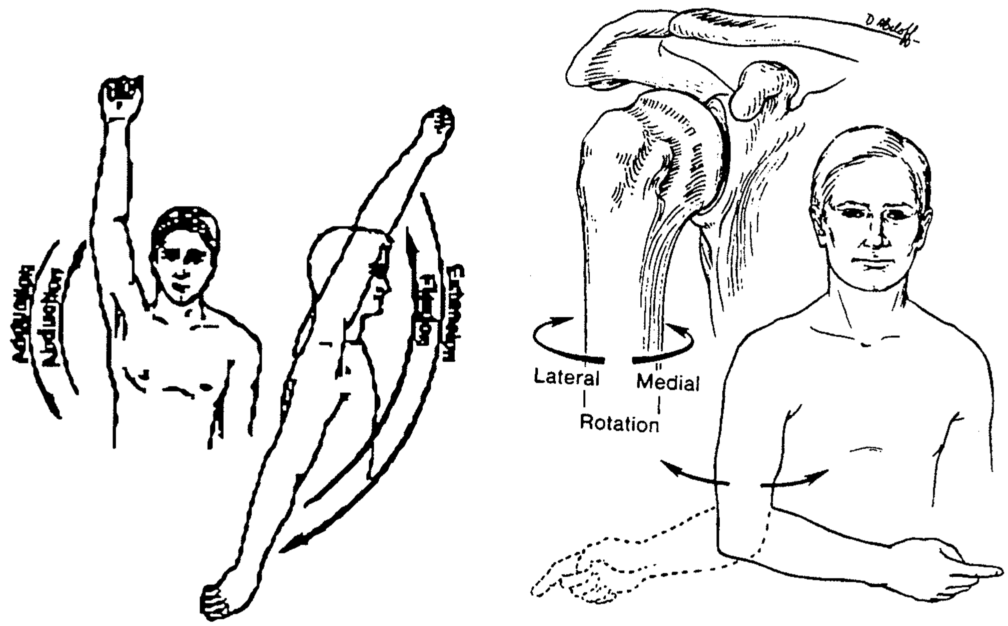
Introduction

This chapter will include a review of articles that are relevant to the movements performed on the incline and supine bench press in terms of anatomy, kinematics, kinetics, and electromyography. The literature review will contain (1) the anatomy of the shoulder and elbow joints with emphasis on those muscles involved in the movements required to perform the bench press exercise, (2) a discussion of the electromyographic apparatus, (3) a brief review of cinematography, and (4) a kinematic description and mechanical analysis of both the incline and supine bench press movements.

There are several muscles involved in the production of the bench press movement. The pectoralis major and anterior deltoid horizontally adduct the upper arm while the triceps brachii serves to extend the elbow joint and assist in shoulder extension. These three muscles along with latissimus dorsi, teres major, serratus anterior, rhomboids and trapezius, the rotator cuff muscles, brachialis, and biceps brachii will be discussed in detail because of their involvement in the bench press exercise.

Anatomy of the Shoulder Joint

The shoulder joint is one of the most interesting and most complex joints of the body. It is defined as a multi-axial ball and socket joint that has a large number of degrees of freedom; allowing approximately 180 degrees of shoulder flexion and abduction, 90 degrees of internal and external rotation, 60 degrees of shoulder extension, 135 degrees of horizontal flexion, and 45 degrees of horizontal extension (Zuckerman & Matsen, 1989). According to Basmajian (1982), this extreme mobility is gained at the expense of stability. In fact, the shoulder is constructed so loosely that it is frequently dislocated (Crouch, 1985). The stability of this joint is dependent on the coordinated contraction of the rotator cuff muscles which attempt to secure the humeral head in the most stable position while still providing the maximum leverage (Bradley & Tibone, 1991).



(a) (b)

Figure 2-1 **(a)** Movements of the shoulder region (scapula, clavicle, and humerus). **(b)** Movements occurring at the shoulder joint (scapula and humerus). (Taken from Basmajian, 1982, p. 88 & 89).

It is common to consider the whole shoulder girdle as one unique joint, but in reality it is not one joint. Basmajian (1982) and Crouch (1985) stated that movements of the shoulder joint should be distinguished from the movements of the shoulder region (clavicle and scapula), although their movements often occur together. For example, movements of the shoulder region involve the clavicle and scapula, whereas movements at the shoulder joint occur solely between the scapula and the humerus (Figure 2-1).

The articular surfaces of the shoulder joint are the shallow pear shaped glenoid fossa of the scapula and the large head of the humerus. Basmajian

(1982) stated that the head of the humerus was three times larger than the glenoid fossa. Therefore the area in contact between the two articulating surfaces is small, which in turn increases the instability of the shoulder. To increase the depth of the socket and give it more stability, a rim of fibrocartilage is attached around the outside of the glenoid fossa.

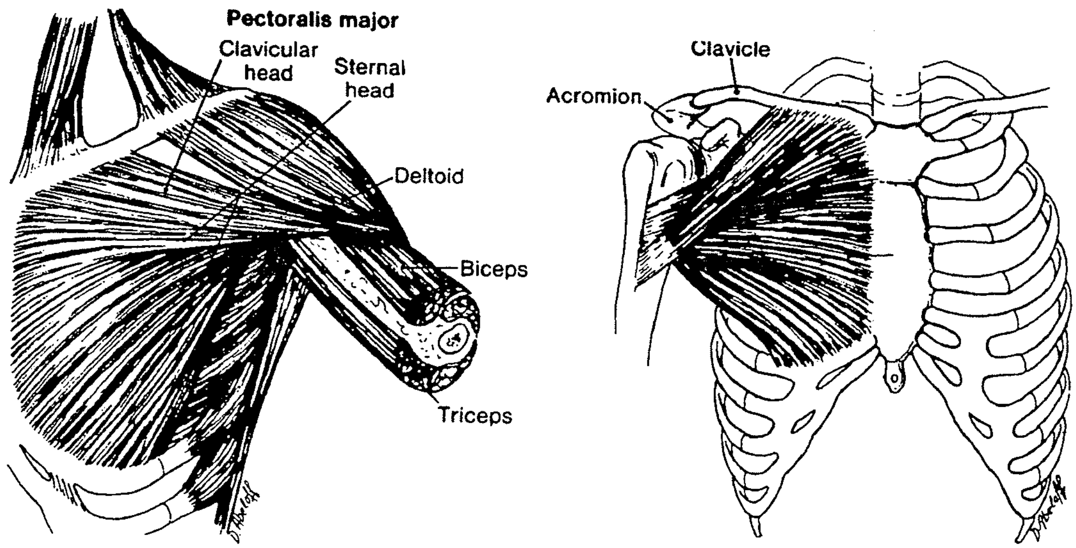
The articular capsule of the shoulder is extremely loose, probably the loosest joint capsule in the body. In fact, it can allow as much as 2.5 cm of separation between the two bones (Crouch, 1985). According to Basmajian (1982), most of the stability of the shoulder joint during shoulder movements is provided by the muscles surrounding it. The reinforcement of the articular capsule is made possible by four muscles that blend with it. It is therefore reinforced in the front by subscapularis, above by supraspinatus, behind by teres minor and infraspinatus, and below by the long head of triceps brachii, although weakly (Crouch, 1985).

The main movements of the shoulder joint complex include abduction, adduction, flexion, extension, medial rotation, and lateral rotation (Figure 2-1). The major muscles involved with these movements are the deltoid, pectoralis major, latissimus dorsi, teres major, rhomboids, trapezius, subscapularis, supraspinatus, infraspinatus, and teres minor.

Pectoralis major

The pectoralis major is described as a large, thick, fan-shaped muscle that covers the superior part of the chest and originates from the mid-clavicle, sternum, ribs and the external oblique fascia. The fibers converge from this broad origin towards their insertion which is comprised of a flat tendon that inserts on the lateral lip of the bicipital groove of the humerus. It is composed of two portions: a sternal head (or sternocostal head) and a clavicular head

(Figure 2-2a). The entire muscle acts as an adductor and medial rotator of the humerus. Alone, the clavicular head flexes the humerus (i.e., raises the arm forward), and from this position, the sternocostal head extends the humerus to the side of the body (Basmajian, 1982; Crouch, 1985).



(a) (b)

Figure 2-2 (**a**) The left pectoralis major and surrounding muscles. (**b**) The right pectoralis major with insertion uncovered. (From Basmajian, 1982, p.145.)

The clavicular head comprises the uppermost quarter of the pectoralis muscle (Figure 2-2a & 2-2b). It arises from the medial half of the clavicle and its fibers passes downward and laterally, overlapping the sternal head to insert on the lateral lip of the bicipital groove (Basmajian, 1982). The sternocostal head is much larger than the clavicular head. It originates from the length of the sternum and from the costal cartilage of the 2nd to 6th thoracic ribs. The fibers of the sternal head run laterally and upward, crossing under the clavicular head to insert on the superior lateral lip of the bicipital groove, and therefore share the same insertion as the clavicular head (Figure 2-2b).

The tendon of the pectoralis major muscle is flat (about 5 cm broad) and consists of two laminae, placed one in front of the other which gives the muscle a twisted appearance. The anterior lamina comprises of the fibers from the clavicle and upper sternum (Figure 2-2b) and has the lowest insertion. The posterior lamina consist of the most distal fibers and deep fibers that run laterally and upward, inserting superiorly to the anterior laminae. Finally, the anterior lamina measures approximately 1 centimeter in length while the posterior lamina is about 2.5 centimeters long (Wolfe, Thomas, Wickiewicz, and Cavanaugh, 1992).

It appears that the pectoralis major muscle has a potential for independent function between clavicular and sternal portions of the muscle as Manktelow et al., (1980) demonstrated independent neural and vascular components in 70% of the inferior one-fifth of the muscle in cadavers. In addition, Wolfe et al., (1992) found that this portion of the pectoralis major had multipennated fibers while the remainder of the muscle had unipennated fibers. The findings of Manktelow et al., and Wolfe et al., suggest that the lower portion of the sternal head may have been a separate muscle in lower mammals. Although the role of this muscle was not speculated on, it probably served a function prior to the evolution from quadriped to bidped locomotion. In addition, Chaffai and Mansat (1988) and Milroy and Korula (1988) noted that the clavicualr portion of the pectoralis muscle that had it's own vascularization and innervation, suggesting that this portion had also a potential for idependent function.

Another unusual or unique characteristic of the pectoralis major is that it contains different fiber lengths. According to Wolfe et al., (1992), different fiber lengths within a muscle allow for differential shortening velocities (depending on the number of sarcomeres in series within each fiber). These differential fiber lengths may allow the pectoralis major to optimize power production over a broad range of muscle shortening velocities (Freehafer et al., 1979).

Deltoid

Crouch (1985) described the deltoid muscle as a large, thick, triangular muscle that helps form the roundness of the shoulder. Superficially, the muscle appears to be one continuous mass of flesh, but in reality it has 3 portions. These portions are the anterior, posterior, and middle fibers (Figure 2-2a & Figure 2-3).

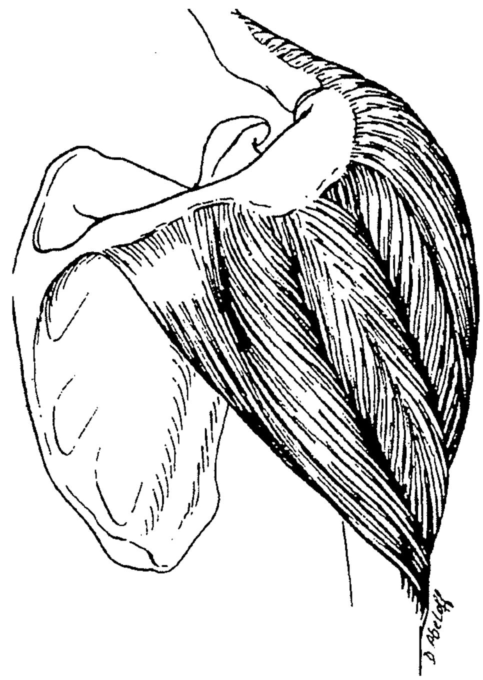


Figure 2-3 The posterior view of the deltoid. (Taken from Basmajian, 1982, p. 148).

The anterior fibers of the deltoid arise from the lateral third (anterior border) and upper surface of the clavicle. The middle fibers originate from the lateral border and upper surface of the acromion. The posterior fibers arise from the inferior lip of the crest of the spine (Basmajian, 1982; Crouch 1985). All three portions converge into a thick tendon that inserts on the middle lateral surface of the humerus, on the deltoid tuberosity.

According to Basmajian (1982) and Crouch (1985), the deltoid as a whole is a powerful abductor of the humerus. The anterior fibers have the same actions as the clavicular head of the pectoralis major which are flexion, horizontal adduction, and medial rotation. The middle fibers are the strongest of the three portions. This strength is attributed to its bipennate muscle fiber arrangement. The middle fibers are solely involved in abduction of the humerus. Finally, the posterior fibers, considered the weakest fibers of the deltoid, are involved in lateral rotation and extension of the humerus as well as adduction of the abducted arm.

Latissimus dorsi

The widest muscle in the back, the latissimus dorsi has its origin from the thoracolumbar fascia (which is located on the lower six thoracic ribs), the posterior half of the iliac crest, from all the lumbar and upper sacral spines, and from the inferior angle of the scapula (Basmajian, 1982; Crouch, 1985). The insertion of latissimus dorsi is made possible by a thin, ribbon-like tendon that is located at the bottom of the bicipital groove of the humerus (Figure 2-4).

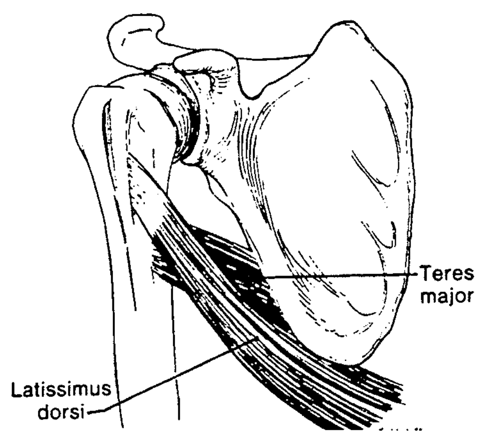


Figure 2-4 The insertion of right latissimus dorsi and right teres major. (From Basmajian, 1982, p. 146).

The latissimus dorsi extends, adducts, and medially rotates the humerus. It works with the pectoralis major in preventing the shoulder girdle from being driven upward in movements such as ordinary pushing with the arms (e.g., bench press) or raising the body and suspending it (Basmajian, 1982).

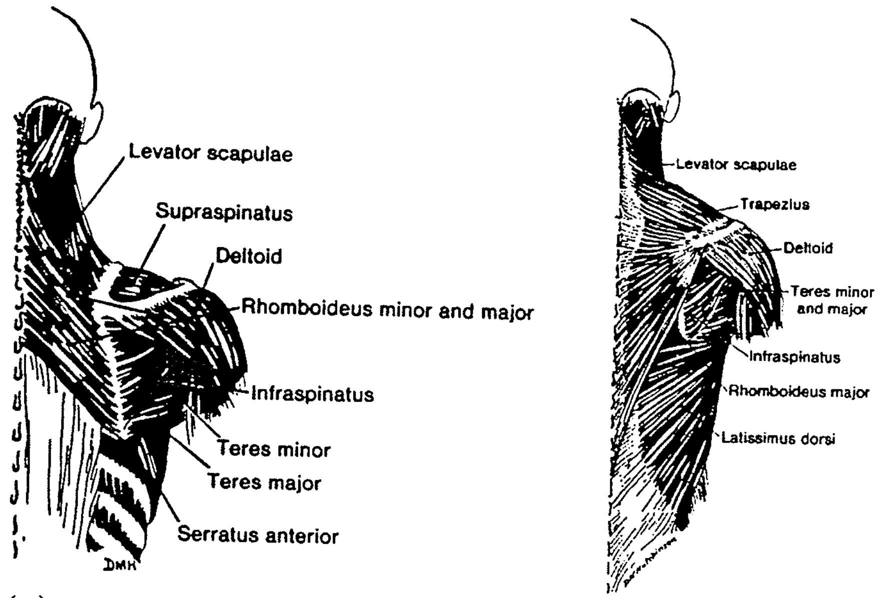
Teres Major

The teres major arises from the inferior angle of the axillary border of the scapula and inserts on the medial lip of the bicipital groove, just behind the insertion of latissimus dorsi (Figure 2-4). According to Basmajian (1982), the teres major has similar actions to those of the latissimus dorsi. These movements are extension, medial rotation, and adduction of the humerus. It also serves to stabilize the upper end of the humerus during abduction (Basmajian, 1982).

Rhomboids and trapezius

Although the rhomboids and trapezius are not directly responsible for the major movements of the shoulder, they contribute significantly as stabilizers of the scapula, especially during abduction, flexion, and lateral rotation of the humerus (Basmajian, 1982).

The rhomboids (major and minor) lie underneath the trapezius and arise from the first four thoracic spines (T1-T2) and the lower end of the ligamentum nuchae (Figure 2-5a). The muscle fibers run downward and laterally where they insert on the vertebral border of the scapula; the attachment extends from the inferior angle to the level of the spine (Basmajian, 1982; and Crouch, 1985). These muscles retract (i.e., pull back) and elevate the inferior angle of the scapula, or in other words, turn the glenoid fossa downward.



(a) (b)

Figure 2-5 (a) Rhomboids minor and major. (b) Trapezius and latissimus dorsi. (Taken from Basmajian, 1982, p. 141).

Crouch (1985) and Basmajian (1982) described the trapezius as a large, flat, triangular muscle that covers the back of the neck and upper half of the trunk (Figure 2-5b). Crouch (1985) stated that the trapezius originates on the occipital bone of the skull, the ligamentum nuchae, and the vertebral spines of the seventh cervical and all of the thoracic vertebrae. The muscle can be divided into three portions of fibers that have separate insertions. Basmajian (1982) stated that the upper fibers descend and insert on the posterior lateral border of the clavicle, the middle fibers run horizontally and insert on the length of the spine of the scapula, and the lower fibers rise and insert on the tubercle of the spine of the scapula that is approximately 3-4 cm from the vertebral border.

Crouch (1985) stated that the insertion of these 3 groups of fibers indicated the function of the trapezius. The upper fibers serve to elevate and rotate the scapula as well as elevate the tip of the shoulder. The middle fibers retract the scapula, and they also stabilize the scapula during the initial movements of flexion and abduction of the humerus. The lower fibers pull the medial end of the scapula downward, cooperating with the upper fibers to rotate the glenoid fossa upward (Basmajian, 1982).

Serratus anterior

Basmajian (1982) described the serratus anterior as one of the most powerful muscles of the girdle, and probably the most difficult to visualize. It is illustrated in Figure 2-6 as a broad and flat muscle that arises by finger-like strips from the outer surfaces of the upper 8-9 ribs and inserts into the superior and inferior angles and medial border of the scapula (Basmajian, 1982).

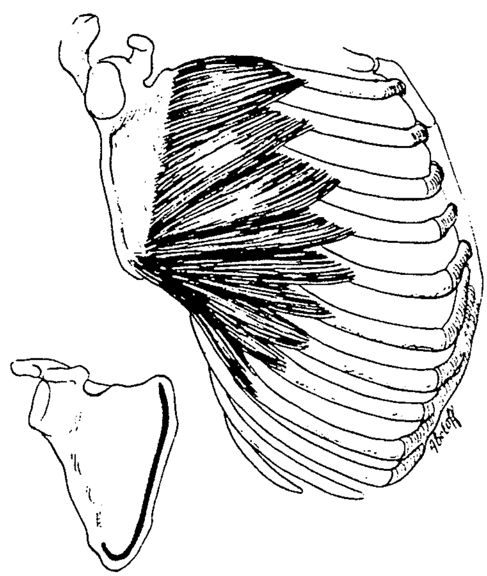


Figure 2-6 Right serratus anterior. The clavicle (bottom left) shows its insertion. Note: the scapula has been moved to expose all of the muscle. (From Basmajian, 1982, p. 143).

The serratus anterior is the most powerful protractor of the scapula and it also rotates the glenoid cavity upward. The upward rotation enables the arm to be raised above the head. It also keeps the scapula fixed to the chest wall, acting as an anchor to allow for movements of the humerus. It is therefore one of the most important stabilizers of the scapula, especially in the bench press movement.

Subscapularis, supraspinatus, infraspinatus, and teres minor

These four muscles form the rotator cuff muscles of the shoulder joint. They are often referred to as the guardians or the dynamic ligaments of the glenohumeral joint (Basmajian, 1982). These muscles, according to Kreighbaum and Barthels (1985), provide very large stability components (Figure 2-7).

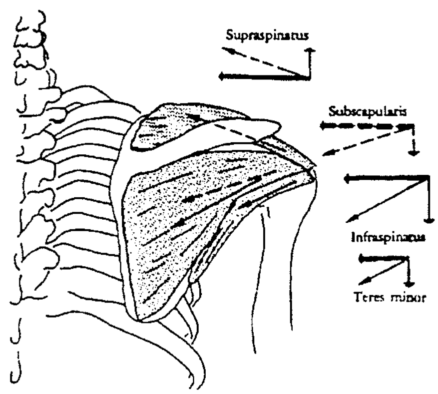


Figure 2-7 The rotator cuff muscles of the shoulder joint and their stabilizing components on the scapula. (From Kreighbaum & Barthels, 1985, p. 200).

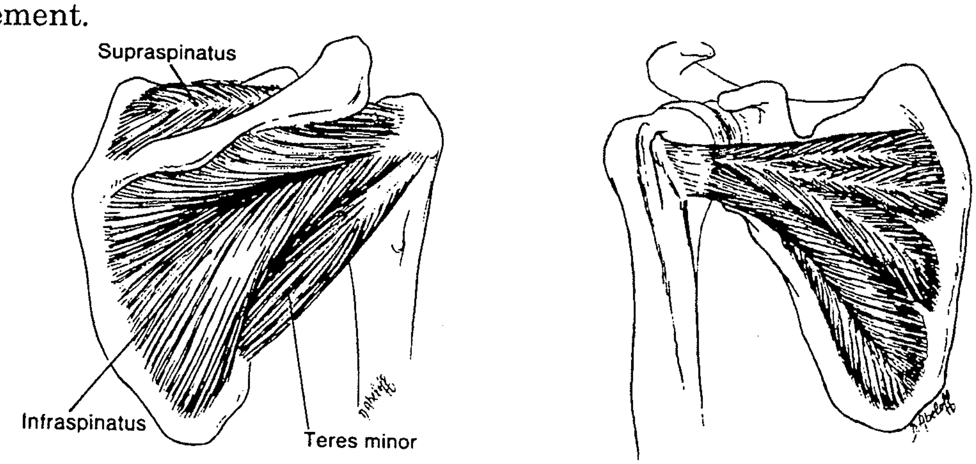
Subscapularis originates in the subscapular fossa and then converges towards the lesser tubercle where it inserts along with the pectoralis major, latissimus dorsi, and teres major. Its action is medial rotation of the humerus (Figure 2-8b).

Supraspinatus arises from the supraspinous fossa of the scapula and passes directly above the shoulder joint and inserts on the greater tubercle of the humerus (Figure 2-8a). Basmajian (1982) stated that the reinforcement this muscle lends the shoulder is made possible by blending its tendon with the joint capsule. Supraspinatus functions in abduction of the humerus, especially at the start of the movement. It also stabilizes the head of the humerus during abduction.

The infraspinatus, according to Basmajian (1982) and Crouch (1985) has a multipennate origin inside the infraspinous fossa of the scapula (Figure

2-8a). This muscle inserts on the greater tubercle just above the teres minor. It also abducts the humerus as well as provides stability by pulling downward on the head of the humerus.

Teres minor arises from the axillary border, just above the insertion of the teres major. Its fibers pass upward and laterally behind the shoulder joint to insert on the greater tubercle of the humerus (Figure 2-8a). Along with infraspinatus and teres major, teres minor produces adduction of the upper arm. By pulling downward it also serves to stabilize the head of the humerus. According to Basmajian (1982), these muscles act together as a force couple, meaning that they pull in different directions to achieve a single desired movement.



(a) (b)

Figure 2-8 The rotator cuff muscles of the right shoulder **(a)** The three muscles of the posterior scapula and **(b)** subscapularis. (Taken from Basmajian, 1982, p. 147).

Anatomy of the Elbow Joint

The elbow joint is described as a hinge-type, synovial joint. It consists of the inferior or lower end of the humerus and the superior or upper ends of the ulna and radius.

The articulation of the elbow joint involves movement of the trochlear notch of the upper ulna accompanied by the upper surface of the radial head on the trochlea and capitulum of the humerus. The trochlea is described by

Basmajian (1982) as a pulley-like surface that fits deep into the trochlear notch of the ulna. It resembles an hourglass laid on its side with one side (medial) having a greater circumference than the other (lateral). The superior radio-ulnar joint is not part of the hinge joint itself. It shares the same articular capsule and the same continuous joint space as the elbow, and should be mentioned for those reasons (Basmajian, 1982).

The elbow joint is surrounded by an articular capsule that is both thin and loose (anteriorly and posteriorly) as well as thick and strong (medially and laterally). The constriction of this capsule therefore allows movement in the anterior-posterior direction while limiting movement in the lateral and medial directions.

The undesirable side-to-side movements of the elbow are prevented in part by the articular capsule and thickenings called collateral ligaments. The two ligaments are the medial or ulnar collateral ligament and the lateral or radial collateral ligament. According to Basmajian (1982) and Crouch (1985) the medial collateral ligament is attached to the medial epicondyle of the humerus and spreads to the medial edge of the trochlear notch of the ulna (Figure 2-9). A cord located in the anterior part of the ligament strengthens the ligament further and plays the most crucial role in preventing abduction of the elbow.

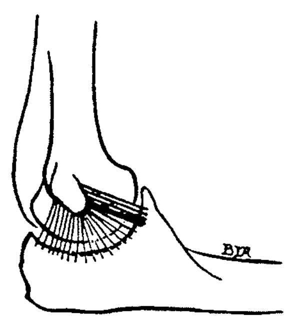


Figure 2-9 Medial collateral ligament of the left elbow joint. (From Basmajian, 1982, p. 91).

The lateral collateral ligament is fastened to the lateral epicondyle and attaches to the outer surface of the annular ligament (a ligament that holds the radius against the ulna). Finally, the muscles of the arm that primarily act on the elbow joint are the triceps brachii, brachialis, and biceps brachii.

Triceps brachii

Crouch (1985) described the triceps brachii as a large muscle that occupies the whole posterior surface of the humerus. It is consists of three portions; the long, the lateral, and medial heads, in which the long head is the only one that crosses the shoulder joint (Figure 2-10).

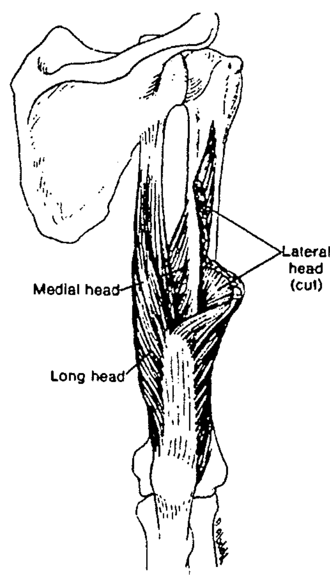


Figure 2-10 Right triceps brachii from behind. (From Basmajian, 1982, p. 152).

The long head of triceps brachii originates from a rough area below the glenoid fossa known as the infraglenoid tuberosity (Crouch, 1985; and Basmajian, 1982). The lateral head arises from a linear origin on the posterior lateral surface of the humerus, and according to Basmajian (1982) it extends 5-8 cm from just below the greater tuberosity of the humerus. The medial head, which lies deep to the long and lateral heads, has its origin on the lower two thirds of the posterior humerus. All three heads of triceps brachii are joined by a large aponeurosis that narrows into a tendon and inserts on the posterior edge of the olecranon process of the ulna (Basmajian, 1982; Crouch, 1985).

According to Basmajian (1982), the triceps brachii is a powerful muscle and the most important elbow extensor. It also serves as a stabilizer of the elbow when the upper limb is pushing against a resistance. It acts as a first class lever system (Figure 2-11) in which the axis (elbow joint) is located between the motive force (triceps contraction) and the resistive force (Kreighbaum & Barthels, 1985). The long head, which is a two joint muscle, also helps to extend the humerus at the shoulder.

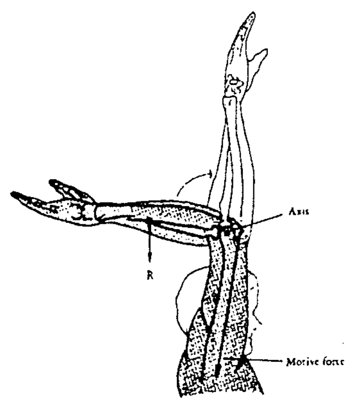


Figure 2-11 The triceps brachii act as a first class lever to extend the elbow. (From Kreighbaum & Barthels., 1985, p. 182).

Brachialis

The brachialis has an extensive origin that covers the lower half of the anterior surface of the humerus. It crosses the front of the elbow where its strong tendon inserts on both the tuberosity and coronoid process of the ulna (Basmajian, 1982; Crouch, 1985). The brachialis is an important flexor of the elbow and according to Basmajian (1982), it is never credited for the movement it produces because it is deep to biceps brachii (Figure 2-12a).

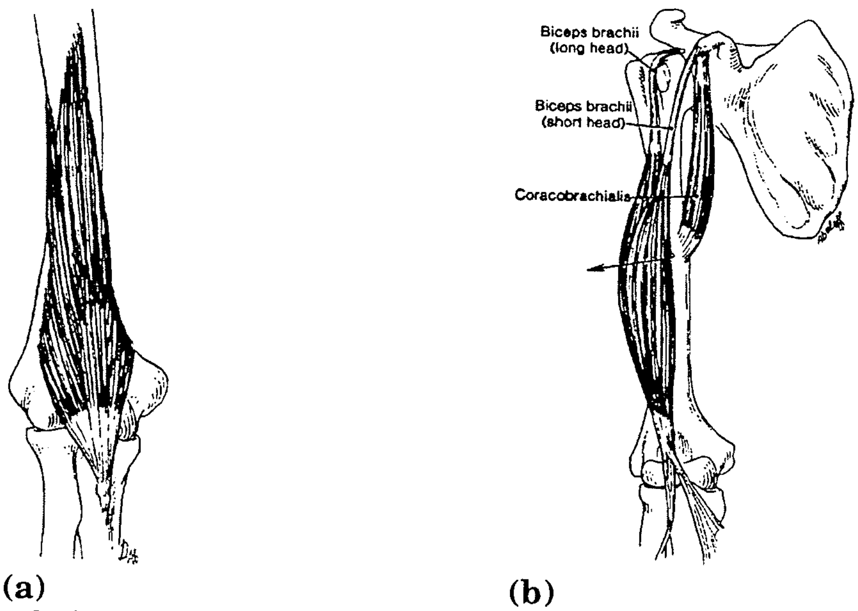


Figure 2-12 **(a)** Anterior view of right brachialis. **(b)** Right biceps brachii. (From Basmajian, 1982, pp. 149-150).

Biceps brachii

Crouch (1985) described the biceps brachii as a long, spindle-shaped muscle that has two heads (long and short). The short head shares its origin with the coracobrachialis at the tip of the coracoid process of the scapula (Basmajian, 1982). The long head arises from the supraglenoid tubercle that is located above the upper end of the glenoid fossa. The two heads of the muscle join halfway down the humerus. It is then inserted by a cord-like tendon, into the radial tuberosity.

A special feature of the long head is that it lies within the capsule of the shoulder joint (Figure 2-12b). It also runs through the bicipital groove and is provided with a synovial sheath that communicates with the shoulder joint (Basmajian, 1982; & Crouch, 1985).

The biceps brachii is a supinator of the forearm and according to Basmajian (1982) it is a power flexor, meaning that it serves to flex the elbow against a resistance as well as supinate the forearm in the absence or presence of a resistance. It also assists with shoulder flexion and abduction.

Electromyography

The use of electromyography (EMG) to measure muscle activity has been used since the late 18th century (Basmajian & DeLuca, 1985). In the field of biomechanics, EMG has contributed to the understanding of how the muscles of the body function in relation to one another, especially how they act together to produce movement. EMG has also provided the scientist with a knowledge of muscle contractions, muscle force and tension, and muscle fatigue. The EMG data collection system has found its place in the laboratories of physical education and sports, vocational medicine, rehabilitation, physical and occupational therapy, and anatomy (Stevenson, 1984).

Generation of a muscle action potential

According to Grieve (1975), the electromyogram is an index of the electrical activity occurring at the muscle cell membrane during excitation. This excitation consists of depolarization and repolarization of the membrane resting potential (Rau & Reucher, 1984). The depolarization of the muscle fiber membrane establishes a small electrical potential (muscle fiber action potential) across the membrane which is a result of the transfer of the charged ions potassium, calcium, and sodium (Stevenson, 1984; Chaffin & Andersson, 1991). The small electrical potential generated across the muscle cell membrane can be recorded by placing an electrode either within or on the surface of the skin (Chaffin & Andersson, 1991). This signal can then be amplified for analysis, and because it indicates the actual electrical activity of a muscle, it is possible to use this information to evaluate how muscles function in certain movements (Stevenson, 1984).

The electromyographic system

The major components of the electromyographic system (Figure 2-13) are the recording electrodes, the amplifier, and the monitor, which can be a cathode ray oscilloscope, a loud speaker, a magnetic tape recorder, or a computer system (Aminoff, 1987). The electrodes pick up the myoelectric activity generated as the result of muscle fiber membrane depolarization. The small electrical signal is then amplified by a preamplifier (usually differential) and then re-amplified by a second stage amplifier, which finally allows the signal to be recorded in the form of an observable EMG signal (Basmajian & DeLuca, 1985; Aminoff, 1987).

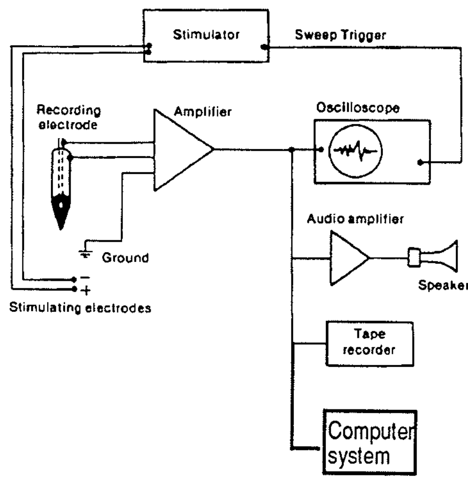


Figure 2-13 Components of an EMG data acquisition system (Adapted from Aminoff, 1987, p. 39).

Electrodes

Electrodes are low impedance devices placed over the muscle of interest. They record the algebraic sum of all motor unit action potentials (MUAP) being transmitted along a muscle fiber during contraction (Winter, 1990). Also, electrodes will better detect those MUAPs closer to it than those further away.

According to Stern, Ray, and Davis (1980), electrodes should be made of either silver and silver chloride or of carefully chlorided silver. These are the material of choice because they are non polarizable, allowing bi-directional flow of current between the electrode and electrolyte interface (Soderberg and Cook, 1984). These types of electrodes should be of low impedance, non-polarizing, stable (i.e., not subject to movement), light, and small in size.

Two types of electrodes are commonly used in recording muscle action potentials, these are surface and indwelling. According to Chaffin and Andersson (1991), surface electrodes are useful for recording the activity of superficial muscles. They are most often used to study movements in an occupational and sports setting, especially for kinesiological studies involving time-force relationships (Basmajian et al., 1985). Indwelling electrodes, such as the needle type, are most common in clinical electromyography, while wire electrodes are more frequently used in kinesiological studies of deep muscles (Chaffin & Andersson, 1991; Grieve et al., 1975). Both indwelling and surface electrodes come in monopolar and bipolar configuration.

Electrodes should be placed on the same muscle groups (when they are paired), and they should be placed along the length of muscle fibers with their orientation perpendicular to the fibers. Electrodes should be spaced according to the length of the muscle and whether the goal is to observe the activity of a single motor unit or the whole muscle (Stern et al., 1980).

Amplifier and preamplifier

Along with the myoelectric activity, the electrodes pick up various unwanted signals (noise) that can contaminate the EMG signal (Aminoff, 1987; Winter, 1990). A preamplifier is placed on or near the recording site to enhance the myoelectric signal and reduce noise or artifacts. Before the MUAPs may be displayed or recorded, further amplification is needed. According to Aminoff (1987), this can be achieved by a differential amplifier which also helps reduce distortions of the signal by rejecting interference signals.

Stern et al., (1980), Basmajian and DeLuca (1984), Aminoff (1987), and Winter (1990), listed four considerations and requirements for the EMG amplifier: (1) amplifier gain, (2) input impedance, (3) frequency response, and (4) common-mode rejection.

Amplifier gain was defined by Aminoff (1987) and Winter (1990) as being the ratio of the amplified output voltage to input voltage. The units of gain are called decibels (dB) and can be calculated using the formula given by Aminoff (1987):

$$\text{Gain (dB)} = 20 \log \frac{\text{Output Voltage}}{\text{Input Voltage}}$$

Stern et al., (1990) stated that amplifier gain should be high to sufficiently output the recording. For example, a 2 millivolts input and a gain of 1000 would result in an output voltage of 2 volts (Winter, 1990). According to Winter (1990), good amplifiers should accommodate a wide range of input signals and should possess a range of gains from 100 to 10 000. Also, the gain of a bioamplifier should be able to render the output signal with an amplitude of approximately 1 volt (Basmajian & DeLuca, 1985). In addition, Winter (1990) stated that signal amplitude should be reported in microvolts (μV).

According to Winter (1990) and Stern et al., (1980) the input impedance must be high enough to ensure that the EMG signal is not reduced as it connected to the input terminals of the amplifier. A minimum input resistance of 1 MOhms was recommended by Winter (1990), yet it is preferable to have at least 10 MOhms. Consider Figure 2-14, a simplified diagram of the biological amplifier for recording electrode potentials. In this diagram, R_{s1} and R_{s2} have replaced the electrode-skin interface.

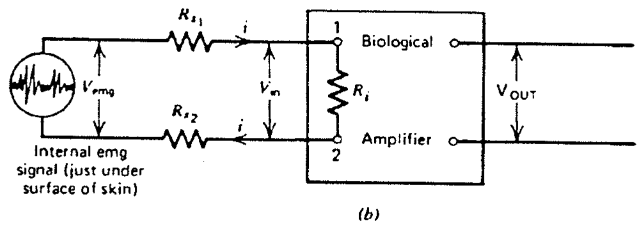


Figure 2-14 Biological amplifier for recording electrode potentials (From Winter, 1990, p.198).

Small EMG signals caused a current to flow through the R_{s1} and R_{s2} resistance to the impedance of the amplifier R_i. Winter (1990) stated that due to resistance encountered at R_{s1} and R_{s2}, a decrease in voltage would be noticed as a result of voltage division. The voltage at their input terminals (V_{in}) would actually be less than the actual signal at (V_{EMG}). If the electrode resistance (R_{s1} and R_{s2}) was 10 000 Ohms and the input impedance of the amplifier (R_i) was 80 000 Ohms, an EMG signal of 2 mV would be reduced to 1.6 mV at the input terminals (V_{in}). According to Winter (1990) this would translate to a voltage loss of 0.2 mV across each electrode. It is therefore important to maximize the imput impedance of the amplifier.

To minimize the voltage loss, one could increase R_i to 1 MOhm and decrease R_{s1} and R_{s2} by preparing the skin better, (or one could increase R_i to 50 MOhms and not care about skin preparation). This would result, according to Winter (1990), in a minimal reduction of the 2 mV EMG signal to 1.998 mV (V_{in}). Therefore it is desirable to prepare the skin (if you have a very low input impedance) to reduce the impedance of the electrode skin interface to 1000 Ohms or less and to have an input impedance of at least 1 MOhm (Basmajian & DeLuca, 1985; Chaffin & Andersson, 1991; Winter, 1990).

Aminoff (1987) defined frequency response as the range of frequencies between the low and high cut off points. The difference between these two cut off points are known as the frequency bandwidth (Basmajian et al., 1985; Winter, 1990). The frequency response of an amplifier should be such that it allows all frequencies of the EMG signal to be amplified equally.

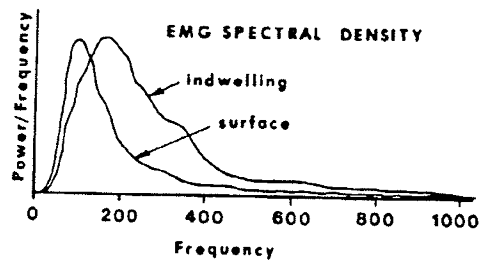


Figure 2-15 The frequency spectrum of EMG. (From Winter, 1990, p. 199).

According to Winter (1990), the surface EMG signal (or myoelectric) has been reported to range between 5 Hz and 2000 Hz (Figure 2-15). Basmajian and DeLuca (1985) recommended varying bandwidths depending on the type of electrodes used and the purpose of the experiment. Frequency responses of 20-500 Hz and 20-1000 Hz were suggested for surface and indwelling EMG electrodes, respectively. Aminoff (1987) and Lenman & Ritchie (1983) recommended a bandwidth of 2 Hz to 10 000 Hz, while Winter (1990) suggested a frequency response of 10-1000 Hz for surface EMG, and 20-2000 Hz for indwelling electrodes. Winter (1990) also suggested that upper frequencies be increased to 5 and 10 KHz for computer pattern recognition of individual MUAPs. Since motion artifact occurs generally at frequencies less 30 Hz, a frequency response of 30-1000 Hz would be sufficient. (If motion artifact is expected, the high pass filter can be set at 30 Hz and if none is expected, the filter can be set at 10 Hz or less.)

Common-mode rejection was defined by Aminoff (1987) and Basmajian & DeLuca (1985) as being the ability to reject common mode signals (hum and artifact). The ability of an amplifier to reject common mode signal is indicated by its common mode rejection ratio (CMRR). The CMRR can be given by the following equation:

$$\mathrm{CMRR} = 20 \log \frac{V_{\text{common mode in}}}{V_{\text{difference mode out}}}$$

Basmajian and DeLuca(1985), Aminoff (1987), and Winter (1990) stated that the higher the ratio, the greater the amplifiers ability to reject common mode signals.

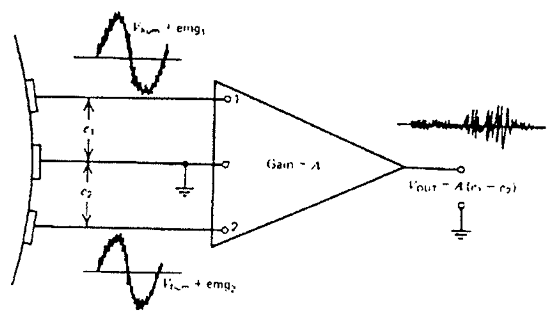


Figure 2-16 Amplifier showing how differential amplifiers reject common signal. (From Winter, p. 202).

In order to reject common mode signals or noise, a differential amplifier is required (Winter, 1990; Aminoff, 1987). Figure 2-16 illustrates how a differential amplifier rejects common mode signals by subtracting the hum (60 Hz signal) of each active terminal. For example, at terminals 1 and 2 the net signals are $V_{hum} + emg_1$, and $V_{hum} + emg_2$, while the amplifier has a gain of A (Winter, 1990):

$$\begin{aligned} e_0 &= A(e_1 - e_2) \\ &= A(V_{hum} + emg_1 - V_{hum} - emg_2) \\ &= A(emg_1 - emg_2) \end{aligned}$$

Winter (1990) stated that the output e_0 was an amplified version of the differences between the EMG signals on terminals 1 and 2. The above description was very idealistic; realistically hum is always present to some extent unless the pre-amplifiers and amplifiers are powered by battery (as electrically powered equipment contributes a 60 Hz noise signal), and operated far away from domestic power sources (Winter, 1990).

CMRR is often expressed as a logarithmic ratio and the units are in decibels (dB), (Aminoff, 1987; Basmajian & DeLuca, 1985; Winter, 1990). According to Winter (1990), quality biological amplifiers should have a minimum CMRR of 80 dB.

Processing of the EMG signal

Although visual inspection on a mechanical chart recorder of the raw EMG recording was the most common way of analyzing muscle activity as it changed with time, the reproduction of its higher frequencies was almost impossible to record on a pen recorder (Winter, 1990). This is because the frequency response of such recorders is 0-60 Hz and cannot record the higher frequencies encountered during periods of high muscle activity (Winter, 1990). Another problem with raw EMG is that it is not suitable for quantification or correlation studies. According to Dainty and Norman (1987), and Winter (1990), the EMG signal must be changed into another processed form in order to be able to quantify the "amount of activity". Winter (1990) and Basmajian & DeLuca (1985) listed the following types of on-line processing as the most common forms of post-processing: half or full wave rectification, linear envelope detector, integration of full wave rectified signal over entire contraction, integration of the full wave rectified to a preset level then set to zero (Figure 2-17).

Rectification

Rectification (Figure 2-17a) of the raw EMG signal involves generating only positive deflections of the signal (Basmajian & DeLuca, 1985; Winter, 1990). According to Basmajian and DeLuca (1985), this is accomplished by either eliminating the negative values (half-wave rectification) or by inverting the negative values (full wave rectification). Full wave rectification is preferred because it retains all the energy of the signal and is a good indicator of changing contraction levels of the muscle (Basmajian & DeLuca, 1985; Winter, 1990). According to Winter (1990), the quantitative use of the full-wave rectified signal alone is limited. It usually serves as an input to other types of processing.

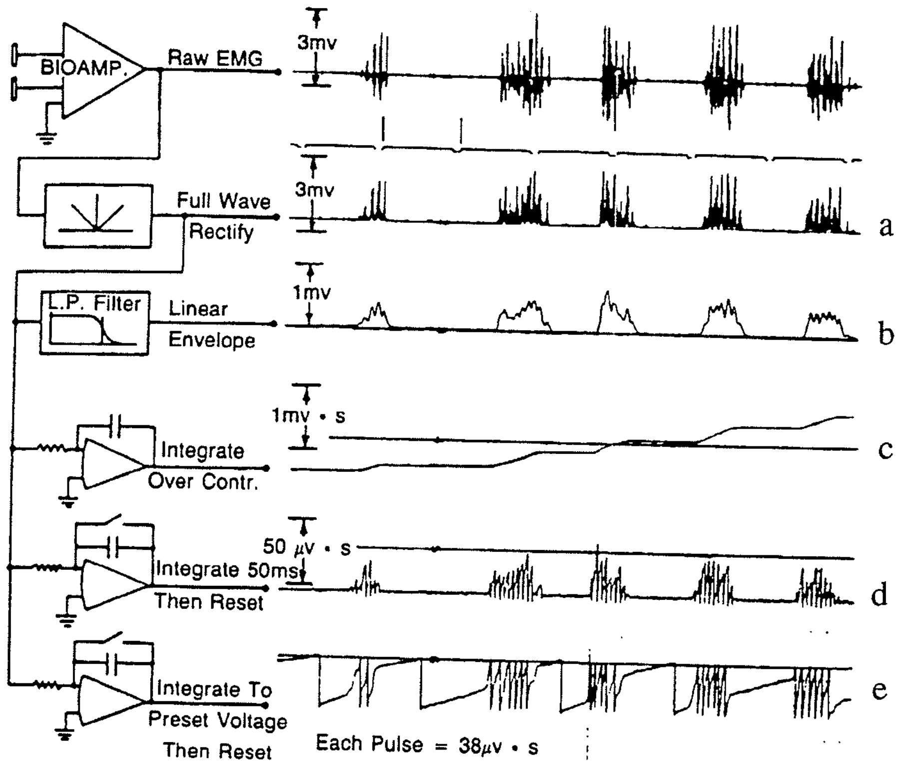


Figure 2-17 Example of several common EMG Processing systems. (From Dainty & Norman, 1987, p. 121).

<u>Linear envelope</u>

A linear envelope is a result of filtering the full-wave signal with a low pass filter (Figure 2-17b). It was described by Basmajian and DeLuca (1985) and Winter (1990) as a moving average because it closely followed the EMG trend and resembled the tension curve. It is a valuable method in assessing how EMG activity changes with time over the period of contraction as well as providing an typical pattern of EMG activity for repetitive movements.

Intergration

According to Basmajian & DeLuca, (1985) and Dainty & Norman (1987), integration is the most commonly used data reduction procedure in electromyography. The purpose of integration is to calculate the area of the rectified EMG signal, or to "measure the area under the curve" (Winter, 1990, p. 206).

The simplest form of integration (Figure 2-17c) begins at a predetermined time and continues during the total time of muscle activity (Dainty & Norman, 1987; Winter, 1990). This type of integration allows muscle activity to be recorded at a desired time, which could be a single contraction or a series of contractions. Average EMG during a specific contraction can be calculated by dividing the integrated value by the time of contraction (Winter, 1990)

The second form of integration (Figure 2-17d) involves resetting the integrated EMG signal to zero at regular time intervals ranging from 40-200 msec. This type of design produces peaks that represent the trend of EMG amplitude with time (Dainty & Norman, 1987; Winter, 1990). According to Dainty & Norman (1987), each peak represents the average EMG over a given period of time, and a series represents the moving average (similar to the linear envelope). One problem with this type of integration is possible noise as a result of frequent resetting. Also, a high reset would not allow the integrator to follow the rapid fluctuations of EMG activity (Winter, 1990).

The third method (Figure 2-17e) involves a voltage level reset. This type of integration allows the measurement of the strength of muscle contractions by evaluating the frequency of resets (Dainty & Norman, 1987). According to Dainty and Norman (1987), this is possible because the integrator quickly charges up to the reset level when muscle activity is high, and when it is low, the integrator takes longer to reach the reset level. Therefore the frequency of reset pulses indicates the level of muscle activity.

Root mean square

Another type of processing used in sports biomechanics is the root mean square method (RMS). The RMS method, like the types of processing discussed earlier, is a time domain analysis. In a sense, it is very similar to linear envelope processing. This type of processing, according to Soderberg and Cook (1984) can provide nearly instantaneous output of the characteristics of the EMG signal. Basmajian and DeLuca (1985) and Soderberg and Cook (1948) recommended the use of this type of processing over any other type of time domain analysis because it possessed a sounder mathematical basis.

Power spectral analysis

Power spectral analysis (also called frequency, harmonic, or Fourier analysis) can be described as a method of decomposing a signal by fitting different frequencies of sine and cosine waves to it. Mathematically, it transforms a signal from the time domain to the frequency domain (Basmajian & DeLuca, 1985; Linstrom & Petersen, 1983). Therefore, a muscle action unit potential having duration and amplitude will be measured from the first deflection from baseline to the point at which the potential returns to baseline. The Fourier transform then dissects the signal into a phase component and a magnitude component, with frequency as the horizontal axis (Spitzer, Wang, Luo, Ward, & Hassoun, 1992).

Originally, spectral analysis of raw EMG signals was accomplished by way of a filter bank that split the signal up in different frequency components, after which the power level of each component was measured. The present method of analysis incorporates the use of a computer which employs a Fast Fourier transform (FFT) to the signal. The steps involve analog-to-digital conversion of the EMG signal into discrete data points which are then multiplied with a time window. Finally, the conversion to the frequency domain is achieved by applying a Fast Fourier transform. The power spectrum component can be derived by squaring the absolute values of the transformed signal.

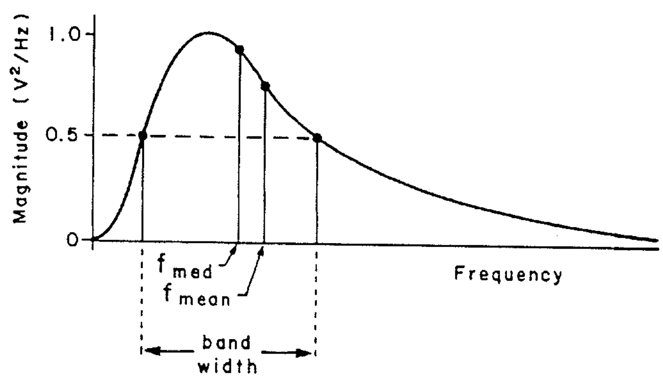


Figure 2-18 Frequency spectrum of the EMG signals. (From Basmajian & DeLuca, 1985, p. 99).

According to Basmajian and DeLuca (1985) analysis in the frequency domain involves the measurement of parameters which describe distinct aspects of the frequency spectrum of the raw EMG signal. As mentioned above, this type of analysis is most commonly achieved by using the Fast Fourier transform. Figure 2-18 illustrates an idealized version of the power density spectrum of the EMG signal; along with the three most commonly reported parameters of the spectrum. According to Basmajian and DeLuca (1985) the linear scales used in this diagram provide a more direct expression of the power distribution and are favored over logarithmic scales which tend to compress the spectrum as well as distort the distribution.

The three parameters shown in Figure 2-18 can be used to provide useful measures of the spectrum which according to Dainty and Norman (1987) possess statistical similarities. These measurements are the median frequency, the mean frequency, and the spectrum bandwidth. Other parameters such as mode frequency, maximum frequency, and segment ratios of the power spectrum have been used, but are not recommended by Basmajian and DeLuca (1985) because they tend to produce less reliable measures. The use of the above values have been used to statistically describe the shape of the following signal parameters: duration, phase, amplitude, and density. Basmajian and DeLuca (1985) cited a study by Stulen and DeLuca (1981) to investigate the reliability of estimating the various parameters of the power density spectrum. They found the median and mean frequency to be most reliable, with the median frequency being less sensitive to noise. Similarly, Hausemanowa-Petrusewicz and Kopec (1983) described the mean and median to have the greatest diagnostic power in assessing motor neuron disease, peripheral neuropathy, and myopathy. For single muscle unit action potentials Spitzer et al., (1992), suggested other parameters such as the low or high frequency energy (area under the curve between specified frequency limits which can be absolute or relative), total energy component (area under magnitude component), and half-power frequency (point where the area under the curve is split into two equal halves).

Power spectrum analysis is very useful in the observation of individual motor units, especially under conditions where muscles are contracting maximally (i.e., a great deal of interference from other motor unit potentials), or when the EMG signal is contaminated. This method has the ability to describe the shape of the spectrum (Figure 2-18) and extract information from noisy signals, unlike the conventional methods of analysis mentioned earlier. According to Lindstrom and Petersen (1983), the ability to describe shape changes in the spectrum opens possibilities to relate EMG to physiological events and mechanism.

The analysis of the power spectrum, according to Dainty and Norman (1987), is a mathematical definition which is an estimate of the spectra. Since estimation errors can occur from the mathematical analysis, Dainty and Norman (1987) suggested that when reporting results from spectral analysis, the estimation procedure as well as bias and variance should be stated. According to Lindstrom and Petersen (1983), the averaging nature of the Fourier transform may be a limitation in itself because it may not be sensitive to the composition of the signal (i.e., firing pattern); and therefore any distinct properties of the signal will be hidden. The Fourier transform allows motor units that have high amplitude signals or that fire frequently to dominate the shape of the spectrum.

Normalization

One of the primary uses of EMG in biomechanics is to make comparisons of the signal between different muscles and individuals. Given the fact that subcutaneous fat, muscle geometry, and other variables are very specific to individuals and different muscles, these type of comparisons are usually not valid (Soderberg & Cook, 1984). According to Basmajian and DeLuca (1985), and Soderberg and Cook (1984) these comparisons can be achieved by normalizing the variables of the EMG signal with respect to their maximum value. The maximal values are obtained by recording the EMG during a maximal voluntary contraction (MVC). The recorded test values are then converted to a percentage of the EMG produced during the MVC, hence, individual variations that limit direct comparisons can be taken into account because of the relative activity measure.

According to Mirka (1991) the point of reference can also be a sub-maximal voluntary contraction (RVC). Along with either the MVC or the RVC reference point, the resting value is also collected in order to quantify the resting level of activity required to hold a segment in a posture (Mirka, 1991). This measure can then be used as a low end reference making the normalized EMG value more reflective of the muscle's activity level during the task (Mirka, 1991).

Normalization also makes the comparison of EMG values over several test days feasible because all the day to day effects due to slight changes in skin impedance, electrode position, and other variables can be controlled.

A limitation, however, is the extent to which individuals can actually elicit a MVC, is unknown. Chaffin and Andersson (1991) also suggested a second problem with normalization which was the fact that myoelectric activity is not linearly related to the produced force over the entire force range. They recommended the use of more than one contraction level to establish a force-electrical activity curve. This should be performed at all the angles of the movement under study to establish this curve.

Mirka (1991) suggested that normalization can be used improperly. According to Mirka (1991), several researchers have tried to normalize the EMG signal taken from dynamic activities such as walking, climbing, and pedaling with respect to the EMG of a single MVC. This procedure was usually done by selecting a neutral or mid-range joint position as the reference angle from which a static MVC EMG signal was obtained, resulting in data that are scaled down versions of the original raw data (Mirka, 1991).

Mirka's (1991) study on the quantification of EMG normalization error proved that using a single EMG signal as a reference for normalization during dynamic exertion was subject to large errors. Under the conditions of the trunk flexion and extension experiment cinducted by Mirka (1991), the errors were as high as 99% for those muscles bearing the greatest torque producing loads (as measured by a Kin/Com). These muscles were the erector spinae group. For the antagonists, whose absolute activity were lower, the errors were similar. The further the segment moved away from the joint location used to normalize the data the greater the error was in normalization.

Mirka (1991) attributed these errors to the fact that by changing the segment position (in this case, the trunk), the muscles move relative to the skin, and therefore the viewing area of the electrodes change. Also, other parameters such as the length/tension relationship, the composition of the muscle, and the number of fibers being sampled can contribute to the error.

It was suggested that the maximum EMG signal used to normalize the task EMG should be taken at several joint angles throughout the range of motion. In turn, this would provide a more accurate representation of the percentage of MVC EMG value at a specific joint angle (Chaffin & Andersson, 1991; Mirka, 1991).

A limitation of this technique, however, is that for every position of interest a resting and maximum EMG must be collected for each muscle. According to Mirka (1991), this can become time consuming as well as fatiguing to the subject, but it is necessary to accurately assess the relative activity of a muscle.

Use of EMG in biomechanics

In biomechanical analysis, electromyography has been used in a variety of ways. It has been used in kinesiological and biomechanical studies for determining muscle function or relating EMG activity to muscle tension. In biomechanics, studies have been directed at muscle length changes, types of contractions, EMG/tension relationships, and muscle fatigue.

Temporal processing

When muscles are relaxed, the EMG signal will be virtually zero as there is no or little activation occurring. As the muscle starts to contract, the activity level will increase proportionally (Stevenson, 1984). This information has been used to identify sequential patterns of various muscle groups in specific movements (Morris, Jobe, Perry, Pink, Healy, 1989; Pink, Perry, & Jobe, 1993); it has also been used to assess the effectiveness of exercises in terms of muscle activation. The ability to observe the timing of muscle contractions has also contributed significantly to the understanding of how muscles work in the everyday setting, especially in the occupational environment (Chaffin & Andersson, 1991; Stevenson, 1984). Temporal elements of the sequences have implications for biomechanists, as muscle activity relative to a particular phase of a movement indicates which muscles are involved during that specified time period.

As mentioned above, temporal analysis EMG has contributed to the understanding of the specific muscle firing patterns in the trunk during a golf swing (Pink et al., 1993), and in the elbow during a tennis forehand stroke, a back hand stroke, and a serve (Morris et al., 1989). Using a reference value (i.e., a maximum voluntary isometric contraction - MVC) produced by the erector spinae and the abdominal oblique muscles, Pink et al., (1993), found definite muscle firing patterns (in terms of percent of reference) throughout the entire golf swing. Both erector spinae and the abdominal oblique muscles were active throughout the golf swing; with ranges in muscle activity between 28-75% MVC for the trunk extensors and 20-64% MVC for the oblique abdominals. Pink et al., (1993) also recorded different firing patterns between the left and right sides of both muscles. Based on the these findings, and the fact that 59% of PGA golfers reported injuries to the trunk, they concluded that golfers should engage in a rigorous strength training program that targets the trunk muscles.

Using fine wire electromyography, Morris et al., (1989), examined the relative activity (i.e., % of MVC) of 8 muscles (biceps, triceps, brachialis, extensor digitorum communis, extensor carpi radialis brevis, extensor carpi radialis longus, pronator teres, and flexor carpi radialis) during three different tennis strokes. All 8 muscles showed low activity during the preparation phase for both the backhand and the forehand strokes (<25% MVC). During the acceleration phase muscle activity increased (>40% MVC), especially in three wrist extensors (60% MVC). The forehand stroke also exhibited greater activity in the biceps and the brachialis. The follow-through showed a marked reduction of muscle activity in all muscles (<29% MVC).

All muscles showed low activity during the wind-up phase of the serve (<25% MVC), while the wrist extensors increased their activity during the cocking phase (25-40% MVC), especially in the late part of cocking, prior to the onset of the acceleration phase (>40% MVC). During the acceleration phase, the triceps, the flexor carpi radialis, the extensor carpi radialis brevis, and the pronator teres increased their activity (>40% MVC). By follow-through all muscle activity was low (>25% MVC), except for biceps which increased during the late-follow-through (25-40% MVC).

Since the wrist extensors were highly active in some portions of all three strokes, this relatively high activity may be a reason why these three muscles are most commonly injured. Given that most elbow injuries are found at the origin of the extensors, on the lateral epicondyle of the humerus, athletes and trainers may want to incorporate a specific weight training program that strengths these muscles.

Ross, Breit, and Britten (1993) used surface EMG to assess the effectiveness of a lumbar support device in maximizing abdominal muscle activity while minimizing the activity of hip flexors. Muscle activity was monitored by surface electrodes placed over the upper and lower rectus abdominus, rectus femoris, and the L3 level of the erector spinae during six different exercises (3 with and 3 without the device) as described in the product manual. Two types of abdominal exercises were designed to primarily affect either the upper and lower abdominal muscle at three levels: beginner, intermediate, and advanced. No significant differences in mean EMG values were observed between the use and non-use of the lumbar support device at all four electrode sites (i.e., upper and lower abdominal). This contradicted the product literature which claimed that the device was better at "isolating" the abdominal muscles.

Ross et al., (1993) also found that hip flexor activity did not decrease significantly and the erector spinae muscle activity remained unchanged. The muscle activity in rectus femoris actually increased significantly in exercises requiring leg elevation which contradicted the statement that the device also made sit-ups safer for the lower back by reducing the involvement of the hip flexors. Finally, the pattern of muscle activity across exercise levels was similar for the upper and lower rectus abdominus, with the advance exercises eliciting greater activity than either beginning or intermediate exercises, (especially in lower rectus abdominus). Since the findings of Ross et al., (1993) suggest that the use of a lumbar support device does not appreciably alter the involvement of either the hip flexor or abdominal muscle, the use of such product does not offer any advantage over similar exercises performed without it.

Piering, Janowski, Moore, Snyder, and Wehrenberg (1993) tested the hypothesis that different abdominal exercises (long lying crunch, bent knee crunch, leg raise, and vertical leg crunch) have specific effects on one of four quadrants of the rectus abdominus. EMG recordings were taken from all four quadrants of the rectus abdominus, and the integrated EMG (IEMG) for each electrode pair was normalized to the highest IEMG for the four exercises within each quadrant (and expressed as a percentage of that highest value). Analysis of the normalized EMG activity revealed no significant difference in the mean muscle activity between the four different quadrants and no differences in the mean muscle activity between the four different quadrants and no differences in the mean activity between the four different exercises. The results of Piering et al., (1993) suggest that in terms of strengthening the rectus abdominus muscle, one exercise trained the rectus abdominus just as well as any other, and employing all the exercises during a training session may not be necessary.

Muscle tension

When tension is changing with time, processing EMG through a linear envelope detector is commonly accepted method for comparing the EMG-tension relationship (Winter, 1990). Winter (1990) also stated that for constant tension experiments, one can calculate the average of the full-wave rectified signal. Finally, Winter (1990) suggested a second method of relating muscle force to EMG activity, by counting the action potential over a given period of time, where count rate would increase with muscular tension.

EMG has also been used to predict tension (Chaffin & Andersson, 1991). Predicting muscle tension is accomplished by expressing the smooth rectified EMG as a fraction of the EMG at short maximal voluntary contractions (Hof, 1984). Using this technique is not without error as determining and eliciting a true maximal voluntary contraction is difficult and at times subjective. Hof (1984), Basmajian & DeLuca (1985), and Chaffin & Andersson (1991) suggested that this method may be accurate for certain muscles while being totally inaccurate for others.

Varied results have been reported regarding the relationship of EMG activity to muscle tension. According to Gregor (1993), Winter (1990), and Stevenson (1984), both linear and non linear relationships have been found (Figure 2-19). Stevenson (1984) further specified that a linear relationship between EMG amplitude and tension was found at the beginning of maximal isometric contraction, and after 30% of maximal isometric contraction, a non linear relationship was observed.

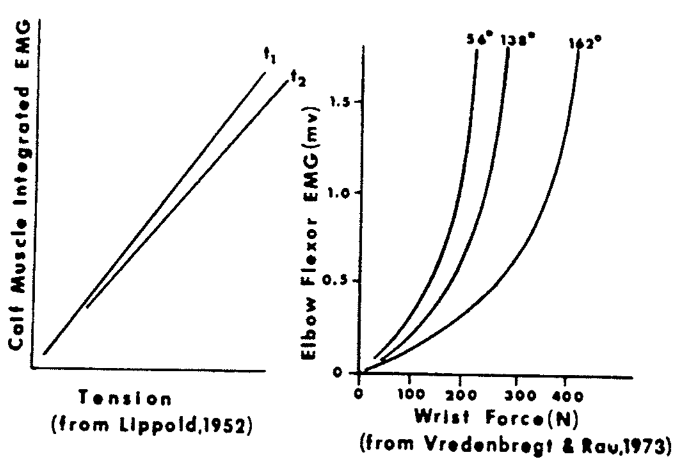


Figure 2-19 Relationship between the average EMG amplitude and muscle tension in isometric contraction. (From Winter, 1990, p. 207).

A study by Woods and Bigland-Ritchie (1983) designed to examine muscles that reflected varying fiber composition, fiber distribution, and force generating patterns found that muscles of near uniform fiber composition produced linear relationships between EMG and force. Non-linear relationships were observed in muscles of mixed fiber composition. Woods and Bigland-Ritchie (1983) suggested that factors such as uneven fiber distribution, motor unit potential amplitudes, and possible supratetanic motor unit activity may have contributed to the non linear EMG/force relationship.

Although fairly constant results are observed in semi-static isometric contractions, this is not true for ballistic movements, or non-isometric contractions (Hof, 1984; Winter, 1990; Gregor, 1993). According to several investigators (Hof, 1984; Solomonow, Baratta, Shoji, and D'ambrosia, 1990; Chaffin & Anderson, 1991), EMG data by itself should not be used to predict muscle force in ballistic movements. Gregor (1993) stated that dynamic movements require lengthening and shortening of muscle and acceleration of body segments, which tends to complicate the relationship between muscle force or joint torque and EMG.

To further illustrate the problems of EMG as a predictor of muscle tension, Black, Woodhouse, Suttmiller, and Shall (1993) conducted a study to determine the influence of hip position on electromyographic and torque production in the knee. Black et al., tested 12 males and 12 females isokinetically at 30 degrees per second at two different positions. Torque values, as recorded by Black et al., were significantly greater for isokinetic knee flexion in the sitting position compared to the supine position, yet EMG activity did not change relative to hip position. Knee extension torque and EMG activity did not change during the sitting or supine position. This may have been a result of muscle length changes during the different movement protocols.

Under dynamic conditions, muscle tension has been seen to lag behind the EMG signal (Winter, 1990). This has been termed electromechanical delay. Electromechanical delay can be determined by comparing the onset of EMG activity with the acceleration of the body segments. According to Soderberg and Cook (1984) these have ranged from as little as 26 msec to as much as 100 msec. Although these delays seem small, they need to be considered in the evaluation of high velocity movements. This delay may be an important factor where tension needs to be generated before a specific event such as the muscular pre-tension required before heel strike during touchdown in running (Soderberg & Cook, 1984).

A recent experiment by Marsh and Martin (1995) sought to determine if a relationship existed between pedaling cadence and the EMG of five lower limb muscles in cyclists and non-cyclists. The study was conducted to lend insight into a current notion, (albeit mostly observational), suggesting that muscle activity is less during a cycling cadence that is preferred by the individual. EMG data was collected (for seven crank cycles) from the vastus lateralis, rectus femoris, biceps femoris, lateral soleus, and medial gastrocnemius during six randomly selected cadences (50, 65, 80, 95, 110 rpm, and the individual's preferred cadence at a constant external power output of 200 W). EMG data from all muscles was normalized to a maximal voluntary isometric contraction.

Other factors that could affect the ability to show a relationship between muscle force and EMG activity are specific muscle characteristics such as: different firing rates, degree of synergistic action of the muscle groups, muscle length, muscle fatigue, signal cancellation (related to different firing rates of motor units), and muscle temperature (Basmajian & DeLuca, 1985; Chaffin & Andersson, 1991; Enoka and Fuglevand, 1993; Gregor, 1993). Other possible limitations included measurement factors such as electrode size and type as well as their impedance (Chaffin & Andersson, 1991).

Contraction type

EMG has been successfully used to illustrate the differences between the types of contractions. According to Stevenson (1984) and Winter (1990), EMG amplitude was greater in eccentric contractions than isometric contractions at the same joint angle. It was also found that eccentric contractions produced greater EMG signals than the concentric contractions. These results, according to Stevenson (1984) and Winter (1990), indicated that EMG amplitude was useful as a relative measure of muscle metabolism.

Soderberg and Cook (1984) state that when the IEMG was plotted against tension, greater slopes were noted at the faster concentric speeds. Little difference between the eccentric contractions was observed, but both were lower than the slope created by the isometric contraction. According to Soderberg and Cook (1984), maximal voluntary contractions produced values that were independent of the velocity and type of contraction. This strengthened the statement made earlier that muscle tension can't be accurately determined by using EMG values.

EMG has also been used to determine the effects of detraining on the types of contraction. Hortobayi, Houmard, Stevenson, Fraser, Johns, Israel (1993), examined the effects of 14 days of detraining on strength, muscle size, and muscle activation. No changes in performance were noted between pre- and post detraining in the bench press, parallel squat, isometric and isokinetic knee extension force, vertical jumping, body mass, and body composition. The percentage of muscle fiber types remained unchanged (vastus lateralis). Type I fiber area did not changed while Type II fiber area decreased significantly (6.4%). Isokinetic eccentric knee extension force decreased significantly. Peak EMG activity of vastus lateralis decreased significantly by 8.4%, 10.1%, and 12.7%, during the isometric, isokinetic eccentric, and concentric knee extension tests, respectively (yet were non-significant). Additionally, peak EMG activity did not change significantly in the biceps femoris for the isometric, isokinetic knee flexion tests. It appears that short term periods of detraining may specifically affect eccentric strength or the size of Type II muscle fibers, while not affecting other components of neuromuscular performance.

According to Hortobagyi et al., (1993), the theory that strength retention can be associated with the phenomenon of muscle memory was not supported by their results. The theory implies that strength retention can occur as a result of increased neural activation or muscle fiber hypertrophy. Clearly this was not observed as EMG activity did not change significantly and the Type II fiber area decreased. Practically, the results of Hortobayi et al., (1993), imply that the concerns voiced by many strength athletes regarding the negative effects (i.e., decrease in strength) of short term detraining may be unfounded.

Fatigue

As a muscle begins to fatigue, more motor units will be used or increased firing rates will be required to maintain the same muscle tension which may be a way to compensate for muscular fatigue. This pattern of neural activation is reflected in the EMG activity where the amplitude or frequency of the signal changes. Winter (1990) cited that higher frequency components of the EMG signal to decrease as the muscle tension remained constant. Also, another change in EMG associated with fatigue was a tendency for the motor units to fire more synchronously, which was noted in signal fluctuations during sustained contractions (Winter, 1990). Others have found EMG activity to increase while force production decreased (Nummela, Rusko, and Mero, 1994).

According to Soderberg and Cook (1984), studies have found linear increases in EMG amplitude throughout the length of the fatiguing contractions (for adductor pollicus longis, quadriceps femoris, and the hand grip muscles). For the biceps brachii, however, a non-linear relationship was observed.

Using power spectrum analysis, amplitude changes have been noted. Significant differences in the mean power frequency have been recorded between fast-twitch fibers and slow-twitch fibers (Basmajian & DeLuca, 1985; Soderberg and Cook, 1984). In subjects with a high proportion of fast-twitch fibers, a significant decline in the mean power frequency of the EMG signal was noted. Only a slight decrease was recorded for subjects with a high proportion of slow-twitch fibers in the same muscle. Other studies using the median frequency have found that parameter to be relatively independent of tension level during brief isometric contractions (Soderberg and Cook, 1984). The studies have also indicated an almost linear decrease with time during fatiguing isometric contractions (for median frequency).

As muscle can no longer sustain maximal contraction (i.e., fatigue), the median frequency of the power spectrum has been shown to change, where a marked compression of the EMG spectrum is noticed (Roy, DeLuca, Snyder-Mackler, Emley, Crenshaw, and Lyons, 1990). This according to Roy et al., (1990), is a result of decreased conduction velocity at the muscle tissue due to localized muscular fatigue. A reduction in median frequency (MF) was observed at the following back extensor muscles (both left and right sides): longissimus throracis, iliocostalis lumborum, and multifidus. Furthermore, Roy et al., (1990), were able to predict which rowers suffered from low back pain, and classify which subject was a port-side rower or a starboard rower. The rowers who suffered from low back pain had not recovered as much as their healthy counterpart when measured 1 minute (5 sec., contraction at 80% MVC) after the test (i.e., their relative median frequency had not changed). The same was true for the port side and starboard side rowers, where the non-dominant side did not recover as quickly as the dominant side (thus the median frequency was more compressed for the non-dominant muscle).

Nummela, Rusko, and Mero (1994) recorded the EMG activity of four lower extremity muscles (medial gastrocnemius, biceps femoris, vastus lateralis, and rectus femoris) of 10 male sprinters during fatigued and non-fatigued running (20 m speed test, 400m, and 20m submaximal run at the average speed of the first 100m of the 400m). Average EMG activity for each muscle was collected during three different phases of running: Preactivation (50 ms prior to ground phase), breaking, and propulsive.

EMG analysis of the four muscles revealed that the average EMG for all muscles differed significantly between the 3 runs in all three phases. EMG activity was greatest during the breaking and the total ground phase at the end of the 400 meters than in the submaximal 20 meter run. Average EMG was greater during the maximal 20 meter run than at the end of the 400m rundduring the propulsion phase and the total ground phase.

Numella et al., (1994), also noted a decrease of 19% in running speed during the 400m which could be partly explained by a decrease in stride length and ground reaction forces. The resultant ground reaction force (as determined by a force platform) decreased 13% in breaking phase and 10% in the propulsive phase between the submaximal 20m run and the 400m run. Average EMG increased 28% in the breaking phase but only 8% in the propulsive phase. The increased EMG activity noted during the 400 m run seen in the study of Numella et al., (1994), may have been an attempt to compensate for failure of the contractile capacity of muscle fibers that had been already recruited. In the context of the study, the increased activation may have been a strategy employed by the nervous system to offset the effects of muscular fatigue.

Limitations

Several factors need to be considered when employing the EMG as an analytical tool. Most of the physiological limitations were discussed earlier and therefore this section will focus on the measurement problems associated with electrodes and amplifiers.

Cross-talk

One of the biggest concerns with electrodes has been cross-talk from adjacent muscles (Stern et al., 1980). For indwelling electrodes this is a minor concern because these electrodes detect signals from nearby muscle fibers, but for surface electrode, a potential problem exists (Basmajian & DeLuca, 1984). Cross talk was defined by Vink et al., (1989) as the superposition of myoelectric activity recorded from muscles other than those for which the electrodes are meant.

Electrode placement

Basmajian & DeLuca (1985) suggested that the detection surfaces of the electrode be oriented in a direction perpendicular to that of the muscle fibers. Although, Snow (1992) agreed that the direction of the muscle fibers should be considered when placing electrodes, he disagreed with the notion that detection surfaces should be oriented in perpendicular fashion. Snow (1992) noted that although this would decrease cross-talk, it would also decrease common mode rejection. Electrodes should also be of bipolar configuration with a small detection surface with the smallest allowable separation between the detection surfaces. Basmajian & DeLuca (1984) recommended a spacing of 1 cm for surface electrodes. They stated that since the amplitude of the muscle fiber action potential decreased approximately 75% of the peak amplitude within a distance of 1 mm (Figure 2-20), the reduction in peak amplitude at a distance of 3 mm was so great that potential for cross-talk became almost negligible.

Muscle trauma

A major limitation of the indwelling electrodes is that they are invasive and may cause a certain degree of muscle trauma (Basmajian & DeLuca, 1985). This may cause a release of calcium ions near the detection surface which in turn creates a small yet measurable voltage potential. This is recorded but is not really indicative of muscle activity. Although fine wire electrodes require insertion by hypodermic needle, they are not painful. They are however, susceptible to breakage during movement. Also, the use of the hypodermic needle may cause an efflux to the site calcium that would bias the recording (Basmajian & DeLuca, 1985).

Reliability

Conflicting reports on the reliability of the EMG have been found. Elliott and Blanksby (1976) conducted an experiment to interpret the reliability of EMG recording during a treadmill run. On the basis of cross-correlation techniques of two different trials, Elliott and Blanksby (1976) concluded that averaged integrated EMG recorded during running was reproducible on an intra-day basis.

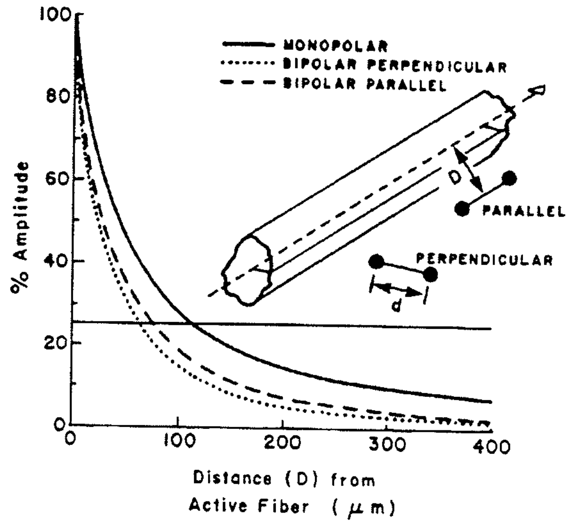


Figure 2-20 Decline of the peak-to-peak amplitude of the muscle fiber action potential. (From Basmajian & DeLuca, 1984, p. 41).

Similarly, Gollhofer et al., (1990) conducted an experiment on the reproducibility of EMG recordings in stretch-shortening type contractions. Four different movements were performed and electrode position on the muscle was varied. These movements were then compared from day to day and week to week through reliability coefficient correlations. Reliability of at least .90 for parts of the EMG pattern as well as the total EMG pattern was found. Gollhofer et al., (1990) concluded that surface EMG was a reliable method for studying stretch-shortening type contractions, especially for reproducing timing EMG patterns. A decrease of electrical amplitude was also noted when the electrode position was changed, therefore suggesting a need for accurate and consistent placement of electrodes when performing any repeated measures experiment.

The need to keep electrode placement constant was reaffirmed by Veiersted (1991). Large deviations were noted in EMG response when electrodes were repositioned within a radius of 3 mm (for the calibration procedure) on the trapezius muscles. A significant depression in EMG response was also reported when electrodes were moved over the middle part of the muscle.

Skin temperature

Skin temperature is another factor that may influence the EMG response. Winkel and Jorgensen (1991) reported an average power frequency reduction of the EMG signal by 59 Hz and a mean amplitude increase of 62 mV when the skin temperature was cooled from an average 32.9°C to 21°C. These changes in the magnitude component of the Fourier domain were significant at the $p=0.004$ and the $p=0.02$ levels, respectively. Winkel and Jorgensen (1991) suggested that these temperature changes measured at the skin could have occurred normally during working hours and therefore these changes were reasonable. Since the mean frequency amplitude and mean power frequency are considered powerful diagnostic parameters in Fourier analysis, the statistical significances mentioned above could cause an experimenter to draw conclusions based on temperature changes at the skin surface, rather than making a statement regarding the effect of a particular treatment. Winkel and Jorgensen (1991) concluded that skin temperature should be carefully monitored, especially during vocational studies. In a laboratory setting, skin temperature can be controlled more readily by monitoring the temperature of the testing environment.

Noise rejection

The most common concern with amplifiers is their ability to reject or eliminate hum (noise). Unfortunately, the majority of this noise comes from domestic power lines that possess an interference frequency of 60 Hz.

According to Winter (1990) this hum would lie right in the middle of the EMG spectrum and nothing can be done to filter it out. However, precautions can be taken to eliminate much of the undesirable noise. The best solution involves using battery powered preamplifiers and amplifiers. The second is to operate this equipment away from possible sources of noise (e.g., fluorescent lights).

Movement artifact also produces noise in EMG signals. This type of noise can be caused by movements of the electrodes on the skin, and movement of the wires leading to the amplifier. Fortunately, these types of artifacts are low frequency noise and lie in the 0-30 Hz range. According to Winter (1990) these artifacts can usually be eliminated by low-frequency filtering (i.e., using a band pass filter or high pass filter that does not allow these low frequency signals to pass). Other solutions include using microamplifiers right at the skin surface, fastening the electrodes to the skin more firmly, and using high quality cabling.

Cinematography

According to Kennedy, Wright, and Smith (1989), and Abraham, (1987), high speed cinematography and video taping are the two methods most frequently used for filming athletic performance. Cinematography and video filming have been used for both kinematic and kinetic calculations, which include: angular and linear displacement, angular and linear velocity, angular and linear acceleration, and the estimation of forces (using the acceleration terms from film).

Cinematography versus video filming

The limitations of video technique include the number of frames that can be filmed per second and the accuracy of the coordinate values due to resolution and quality of the video image (Angulo & Dapena, 1992). High speed cinema cameras have the capability of recording at 500 frames per second, whereas regular video cameras are limited to 60 frames per second. Thus, the video filming of movements that require large accelerations is limited by the relatively low frequency response of the system. For very high accelerations, such as 600 ms^{-1} and 10^5 degrees per second (as found in baseball pitching), reported by Denoth et al., (1984), even high speed cinematography was limited. Therefore high acceleration values such as the ones reported by Denoth et al., would definitely tax the video analysis system, unless high speed video cameras were available.

One of the most important considerations when comparing video to high speed filming is the accuracy of the coordinate values. These tend to be affected by the resolution and quality of the video image; which is limited by the size of the pixels used on the monitor (Angulo & Dapena, 1992).

Kennedy et al., (1989), compared the accuracy of the two methods for repredicting points in the x, y, and z planes. They found the average error of the coordinates of the points of their 2 meter control object to be 4.8 and 5.8 mm, for film and video, respectively. Although the cinematographic method was significantly more accurate ($p>0.05$), Kennedy et al., (1989), stated that the 1 mm difference was not large enough to consider cinematography to be more accurate in terms of point reproduction.

Angulo and Dapena (1992) found greater error in accuracy for video when using a large field of view (8 meters). A resultant error of 10 mm was recorded for the reconstructed coordinates of the control object, compared to the 4 and 5 mm for the large and small film image, respectively. For the external landmarks in the "xy" plane, the resultant error for the video technique was larger (39 mm) than in the larger (29 mm) and smaller (28 mm) film image techniques. Although the accuracy of the video analysis technique was affected by a larger field of view, Angulo and Dapena (1992) noted that within the volume of the control object, the video technique was accurate enough for most applications.

More recently, Sholtz and Millford (1993) evaluated the accuracy and precision of the Peak Performance Technologies video motion analysis system for three dimensional angle reconstruction and found the system to be accurate and reliable. Two cameras, positioned so that the optical axes formed an angle of 60 degrees, video taped the pendular motion of a bar at three different positions with respect to optical axes of the two cameras. The video taped motion was digitized and 32 angles between 18 markers were calculated. Intraclass correlation coefficients (ICC) between trials within each pendular orientation and across orientations revealed excellent ICC's of greater than 0.999 which suggested that calculated angles from each trial were extremely consistent from trial to trial. Scholtz and Millford (1993) also compared the angles derived from the system with those calculated using trigonometry. Deviations from the actual angle were relatively small and averaged 0.0 - 0.8 degrees across all angles and orientations. In addition, intraclass correlation coefficients were greater than 0.999, suggesting that the results were consistent. In terms of planar movements, such as the one used in the experiment of Scholtz and Millford, it appears that the Peak Performance Technologies motion analysis system can provide accurate and reliable results.

The use of high speed cinematography has obvious advantages over video filming, however, Dainty and Norman (1987), and Abraham (1987) stated that the prohibitive cost of high speed cinematography has made the use of video taping for movement analysis a popular option for many researchers. Abraham (1987) also stated that for moderate speed movements, the affordable video system provided reasonable image resolution, freeze-frame analysis, and single frame advance. Kennedy et al (1989) and Dainty and Norman (1987) were in agreement with Abraham's statement; they also added that video film analysis was relatively easier to use and had an inexpensive and shorter film processing time. In fact, the images are immediately available which can allow the investigator to control the quality of the image during the recording session (Angulo & Dapena, 1992).

Kinematics, kinetics and mechanics of the bench press

Technique of the bench press

The bench press exercise can be broken down into three different phases that generally describe the motions occurring during this skill; these are the ascent phase, pause, and the descent phase. The descent phase begins from the starting position (Figure 2-21a) where the individual lies supine on the bench with the bar at arm's length (i.e., the elbows are fully extended) and perpendicular to the anterior deltoid. As the bar is lowered to the chest (descent phase), the following muscles contract eccentrically to control the descent of the bar: anterior deltoid, pectoralis major, and triceps brachii. Finally, the training handgrip for this movement requires the forearm to be pronated with hands placed on the bar slightly wider than shoulder width (O'Shea, 1976).

The descent phase of a properly executed bench press has a characteristically slower descending bar speed than its ascending counter part. A slower descending bar speed, according to Hemba (1990), served to: decrease the potential trauma to the chest, the shoulder region, and the sternum (especially for athletes that employ a wide grip on the bar); increase muscle fiber recruitment (i.e., greater eccentric loading), and ensure that the prime mover muscles contract through the full range of bar motion.

The pause starts when the bar touches the chest (Figure 2-21b). Hemba (1990) stated that the contact point should be at the highest part of the chest to maximize the mechanical advantage of the triceps brachii and deltoid muscles. However, a higher contact point will in effect reduce the contribution of the pectoralis muscle (i.e., muscle is not fully stretched, and therefore the contraction will not be as forceful). This position would also decrease the range of motion through which the three prime movers will be required to contract. Once the bar touches the chest it should lay motionless on the chest for a mental count of "one thousand and one" (approximately 0.5-1.0 second).

After the pause phase is finished (Figure 2-21b), the ascent phase begins and the weight is pressed to the starting position by vigorous extension of the elbow (with a left-right axis of rotation through the elbow) and horizontal flexion of the shoulder (through a longitudinal axis about the glenohumeral joint). The movements in the ascent phase are achieved through concentric contraction of the pectoralis major, anterior deltoid, and triceps brachii.

Table 2-1 Ascent and Descent times reported in various studies

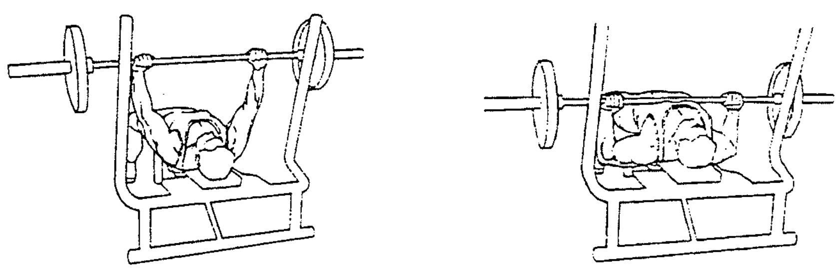
Load (% of 1 RM)	Descent time (sec)	Ascent time (sec)	Authors
100% RM	2.34	1.91	McLaughlin & Madsen (1984)
100% RM	-	2.27	Elliott et al., (1989)
81% RM	-	1.50	
100% RM	1.158#	1.758#	Madsen & McLaughlin (1984)
	1.723*	2.211*	

\# Novice group
* Expert group

The average time for a repetition is about 3 - 4 seconds with the descent phase comprising the bulk of the time (approximately 1.5 - 2 seconds); and the ascent phase lasting about 1-1.5 seconds, with a pause of 0.5 seconds (Hemba, 1990). The time needed (Table 2-1) to complete the movement will also be dependent on the load (i.e., maximal loads would take longer than submaximal loads) as well as the number of repetitions performed (fatigue effect).

O'Shea (1976) also stated that no bouncing or heaving should be permitted when the bar comes into contact with the chest. Also, the buttocks must remain in contact with the bench throughout the pressing movement, with the elbows pointed outwards or in at the sides. Finally, the subject should inhale as the bar is lowered and exhale as it is lifted. In competitive terms, Hemba (1990) listed the following requirements of a legal bench press:

1. Shoulders and hips must remain on the bench at all times;
2. Feet must remain flat on the floor throughout the lift;
3. The bar must be raised evenly throughout the ascent phase; and,
4. Arching of the lower back (bridging) is permitted only if the shoulders and hips remain in contact at all times.



a b

Figure 2-21 The bench press. (From Pearl & Moran, 1986, p. 245).

The incline bench press employs the same technique described by O'Shea (1976) and Hemba (1990) with the exception that the trunk is inclined during the movement rather than supine (Figure 2-22). No studies have been located which determine the existence of distinct technical aspects in the incline bench press, yet one would expect to find similar movement patterns to those the supine bench press. Although these movements are similar, authors of popular weight training manuals (O'Shea, 1976; Bijarnson, 1986; Pearl & Moran, 1986; Moran & McGlynn, 1990; Reynolds, 1991) suggest that the incline bench press places a greater emphasis on the clavicular portion of pectoralis major.

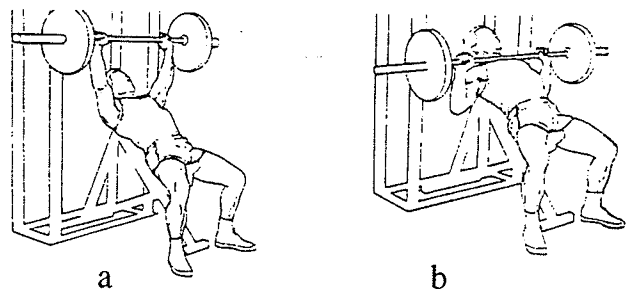


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Figure 2-22 The incline bench press. (From Pearl & Moran, 1986, p. 245).

The rationale describing why a change in bench inclination poses a greater stress on the clavicular pectoralis, has been poorly explained. By examining the direction of muscle fibers for the pectoralis major (Figure 2-2a & 2-2b), the line of pull for that muscle can be determined. The line of muscle pull for a muscle can be drawn as a straight line starting at the origin of the muscle, passing through the muscle belly, and ending at the point of insertion for that muscle. The pectoralis major muscle has two distinct lines of muscle pull (Figure 2-23a). In addition, the angle at which the tendon pulls on the bone will also influence the magnitude of the muscle force. A tendon to bone attachment angle of 90 degrees is considered optimal because the moment arm for the line of pull of the muscle is greatest, which in turn increases the mechanical advantage of the system. Although such an arrangement is not common in the human body, the muscle attachment with the closest angle to 90 degrees will have the capability of producing the greatest torque.

The line of muscle pull and it's angle will also change throughout the range of motion. To date, no studies have been published attempting to describe how the line of muscle pull for the pectoralis major changes with different orientations of the humerus, (as seen in the incline and supine bench presses).

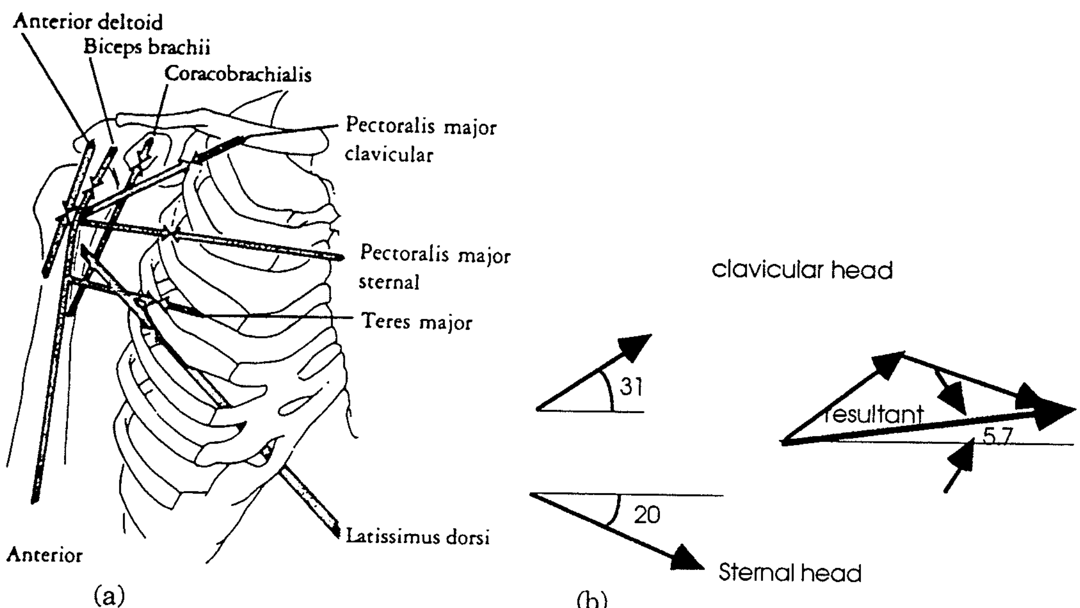

Figure 2-23 **(a)** Line of muscle pull for the pectoralis major (sternal and clavicular) and surrounding muscles (From Kreighbaum & Barthels, 1990, p.196). **(b)** Representation of the angle of muscle pull for both heads of the pectoralis major and resultant line of muscle pull (Adapted from Enoka, 1994).

However, Enoka (1994) has provided values (as measured from cadavers) for the angle of muscle pull for both the clavicular and sternal heads in the anatomical position (Figure 2-23b). The angle of muscle pull for the clavicular head was approximately 31 degrees above the horizontal and the line of muscle pull for the sternal pectoralis was 20 degrees below the horizontal. Using a graphical method (as depicted in Figure 2-23b), the resultant line of muscle pull for the entire muscle can be calculated (i.e., tip to tail method). When the vector quantities were drawn to scale (muscle length measured from cadavers) and at the appropriate angulation, the resultant line of muscle pull was calculated to be 5.7 degrees above the horizontal.

The change in the incline may alter the orientation of the humerus (i.e., greater shoulder flexion and more external rotation) which in turn changes the line of muscle pull of the clavicular pectoralis, sternal pectoralis, and anterior deltoid, so that the clavicular pectoralis may become a more important mover for this exercise. This suggestion may have been acquired from the findings of Inman, Saunders, and Abbott (1944) who determined the clavicular pectoralis to be most active at 115 degrees of shoulder flexion. The inclination of the bench press may be such that in order to maintain a vertical bar path, the subject must adopt an angle of forward flexion of the humerus that is close to 115 degrees, which according to the results Inman et al., (Figure 2-29), would elicit greater involvement from clavicular pectoralis.

Biomechanics of the bench press

<u>Muscle action in the bench press</u>

The muscles primarily involved at the glenohumeral joint to perform the upward movement of the bench press are clavicular pectoralis, sternal pectoralis, and anterior deltoid; the prime mover at the elbow is the triceps brachii (Hemba, 1990). According to Kreighbaum and Barthels (1990) the posterior rotator cuff muscles are also involved in bench press exercise, providing stability. Two major moment arms exist for this compound movement: one about the shoulder through the left-right and longitudinal axes of rotation (Wagner, Evans, Weir, Housh, & Johnson, 1992), and one about the elbow through a left-right axis (Elliott, Wilson, and Kerr, 1989). These produce resistance torques about the respective joints.

<u>Moment arms in the bench press</u>

As the execution of the skill progresses in time, the moment arms of the bar weight (mentioned above) change in length, therefore altering the magnitude of the resultant joint moment for both joints. At the beginning of the skill the moment arm at the shoulder is at a minimum and at maximum for the bar movements in the sagittal and transverse planes, respectively. This can be simply observed by inspecting Figure 2-24. Figure 2-24a illustrates the moment arm about the shoulder at two different instants (full elbow extension and chest contact). As one can see the moment arm about the shoulder is influenced by elbow extension, as it increases as the elbow moves to complete extension. At chest contact, the elbow is flexed at approximately 90 degrees and the moment arm of the shoulder is minimal. It should be noted that the moment arm for the elbow is minimal if almost negligible at chest contact because the weight of the bar is acting through the elbow' s axis of rotation. As well, part of the bar's weight is being supported on the chest, so the weight torques about the elbow and shoulder are decreased.

The resultant moment arm of the weight about the shoulder axis in the sagittal plane can be easily estimated by referring to Figure 2-28 (ascending bar path of expert and novice bench pressers). Since the moment arm is affected by the excursion of the bar relative the shoulder joint, and assuming that the end of the bar is represented by the event markers on the diagram, one can see that as the bar moved upward the moment arm decreased.

The torque required by the muscles to execute this movement will also depend on the relative length of the muscle, the moment arm of the muscle, and the speed of contraction. Since studies have shown that the path of the bar is not a vertical one, the moment arm of the load about the shoulder in the sagittal plane changes constantly, and will be somewhat dependent on the angle of ascent and descent of the bar.

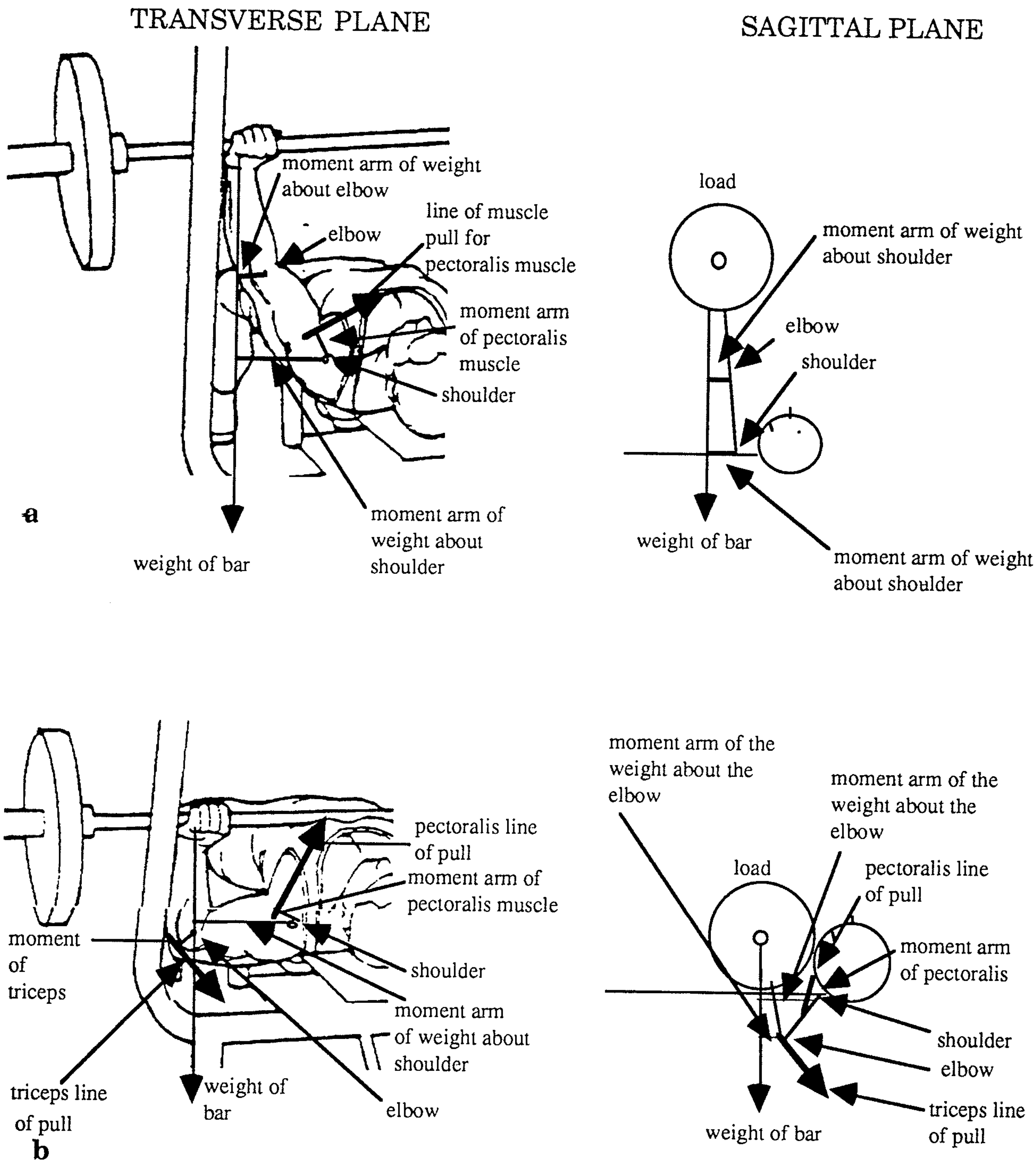

Figure 2-24 Example of moment arms occurring at the shoulder and elbow through a longitudinal axis and a left right axis in both the sagittal and transverse planes. (a) Elbow extension and (b) pause phase (chest contact). (Note that these are also illustrations of moments occurring about the shoulder and elbow joints, not just moment arms).

Elliott, Wilson, and Kerr, (1989), measured the moment arm of the weight about the shoulder and the elbow joint axis for the bar movements in the sagittal and transverse planes, at three different bench press loads (81%, 100%, and 104% RM). They found moment arms about the shoulder and elbow joint to be similar over three different load conditions (i.e., the resultant moment arm decreased throughout the ascent phase as the bar was displaced towards the shoulder). The moment arm of the 81%, 100%, and 104% loads decreased by 8.6%, 8.3%, and 8.6%, respectively.

The resultant moment arm about the elbow decreased throughout the first half of the ascent, reaching a minimum (5.1 cm) during the sticking region. By the end of ascent, the moment arm had increased by 4.6 cm (100% load). Elliott et al., (1989) postulated that since the resultant moment arm about the elbow axis (for the failed load) increased by 15.6% in 0.19 seconds, (from peak acceleration during the maximum strength region to peak vertical bar displacement), thi may be the reason why 8 of 10 subjects failed to lift the 104% RM load during the deceleration phase. This sudden increase in moment arm may have been brought about during elbow extension as the elbow joint rapidly moved away from the line of action of the weight in the transverse plane. Based on these findings, athletes interested in improving their bench press performance may want to incorporate a more rigorous strength training program that focuses on the elbow extensors.

Accompanying the changes in the moment arm length of the weight about the shoulder and elbow axis are variations in the moment arm lengths of the prime movers (anterior deltoid, sternal pectoralis, clavicular pectoralis, and triceps brachii). Few studies have been conducted regarding the length of moment arm (and changes in length during movement) for selected muscles. Unfortunately, no studies located reporting the changes in the moment arm for the pectoralis major. However, changes in the moment arms for the anterior deltoid, middle deltoid, posterior deltoid and several other shoulder muscles have been reported in Soderberg (1986). Since the anterior deltoid produces relatively the same movements (e.g., horizontal adduction, humeral flexion, and medial rotation) as the pectoralis major, it is not unreasonable to assume that variations in the anterior deltoid moment arm during movement would reflect length changes in the pectoralis moment arm. Soderberg (1986) reported a study by Walker et al., (1977) , which found that from 30 degrees to 150 degrees of abduction, the moment arm of the anterior deltoid increased from 0.7 cm to 4 cm (measured in the transverse plane). Given the location of the pectoralis major in relation to the glenohumeral joint, it is likely that the moment arm lengths for this muscle to be slightly longer (1-2 cm) than the moment arm for anterior deltoid.

At chest contact, the four muscles most active in the bench press are in a stretched position, and as a result, the muscle line of pull is relatively close to the joint axis of rotation that these muscles span (except at the elbow). At this point, their respective moment arms lengths are minimal. As these muscles contract, and the humerus is adducted, the angle of pull changes. This would appear to have the effect of increasing the perpendicular distance (moment arm) from the line of action of the muscle to the axis of rotation. The moment arm from the line of muscle pull of each muscle to the axis would then be maximal at the end of the ascent or start of the descent. However, at the end of the lift when the bar is at the highest position, the muscles are in a shortened position and therefore, the force output is decreased due to the length tension relationship.

Elliott, Wilson, and Kerr (1989) found linear displacements of the elbow to be inconsistent across subjects during the initial portion of the acceleration phase. Subjects either moved their elbows medially, laterally, or maintained a stationary position in relation to the shoulder axis. After this brief period, Elliott et al., stated that all subjects moved their elbows laterally throughout the remainder of bar ascent. This had the effect of keeping the elbow joint closer to the line of action of the bar in the transverse plane as well as keeping the elbows under the bar in the sagittal plane. This movement, according to Elliott et al., (1989), served to reduce the moment arm about the elbow in the first half of the ascent. Finally, Elliott et al., noted a consistent change in elbow angle for the 81% and 100% loads, where the angle increased by an average of 55.5 degrees (from the end of the acceleration phase to the end of the lift). The failed load (104%) had a significantly smaller elbow angle of 103^{o} than the 108^{o} reported during the 100%, especially at the sticking point.

Kinematics of the bench press

<u>Phases of the bench press</u>

The following phases of the kinematics of the bench press have been identified by Wilson et al., (1989), Elliott et al., (1989), Madsen and McLaughlin (1984), McLaughlin and Madsen (1984), and Lander et al., (1985): Acceleration phase, sticking region, maximum strength region, and deceleration phase (Figure 2-25). In addition, the existence of a sticking region was dependent on the load, as it was present for maximal loads, but not for all sub-maximal loads (Madsen & McLaughlin, 1984; McLaughlin & Madsen, 1984; Lander, Bates, Sawhill, and Hamill, 1985; Elliott et al., 1989; Wilson et al., 1989).

Although Figure 2-25 illustrates a force/time curve of a representative maximal load (Wilson et al., 1989), the general patterns of bar acceleration can be described. According to Newton's second law, force applied is directly proportional to acceleration (F = ma) , therefore an increase in acceleration will result in a corresponding increase in force applied to the bar. By observing the force/time curve, the trends in bar accelerations can be observed. The numbers printed along the force/time curve in Figure 2-25 correspond to specific events in the ascent phase:

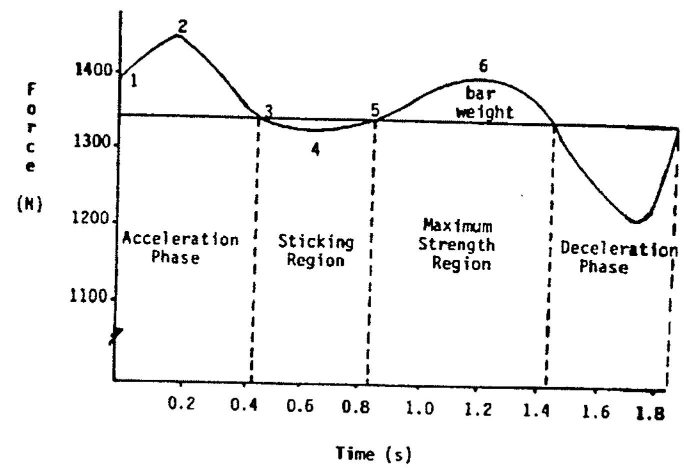

Figure 2-25 Force time curve for the ascent phase of a maximal load. (from Wilson et al., 1989, p. 394).

The acceleration phase was the portion of the ascent of the bar, which started at the (1) chest, encompassed (2) peak bar acceleration, and ended at the point where the bar reached its (3) peak velocity. The sticking region was the period from peak velocity (3) to minimum velocity (5) and incorporated the instant of minimum bar acceleration (4). This is also the point where the subject exerted minimum force on the bar. The maximum strength region began after the point of minimum bar velocity (5) and consisted of a point of peak acceleration (6); it ended at an instant of reduced velocity where the force exerted on the bar was similar to points 3 and 5. The deceleration phase therefore started after the instant of reduced velocity in the maximum strength region and ended at the point of maximal vertical displacement.

The relative distances from the chest where the above points occurred have been reported; with the exception of (3) peak velocity. The point of maximum force exertion (2) was reported to occur at 2.08% (Elliott et al., 1989), 1.66% (Lander et al., 1985), and 2.6% (Madsen and McLaughlin, 1984) of the vertical distance from the chest. Minimum force exertion (4) occurred between 34.7 % (Elliott et al., 1989) and 33.9% (Madsen and McLaughlin, 1984) of way through the ascent phase. Lander et al., (1985) reported the position of minimum bar velocity (5) to be at 41.8% of the vertical distance from the chest while Elliott et al., (1989) and Madsen and McLaughlin, 1984 measured it at 47.9% and 50.2%, respectively. Finally, peak force (6) exerted during the maximum strength region occurred 60.1% of the way through the vertical movement in the Elliott et al., (1989) study, while Lander et al., reported it to occur at 67.7% of the vertical distance.

The relative duration of the above "sub-phases" have been measured for the ascent phase of various load conditions. The relative duration (expressed as a percent of total time) of the acceleration phase has been recorded to be 25.1% for a 75% RM load and 15.8% for a 90% RM load by Lander et al., (1985)., 16.2% for a 100% RM by Elliott et al., (1989), and Wilson et al., (1989), and 20.4% for 104% RM (Elliott et al., 1989).

Lander et al., (1985) reported the sticking region to comprise of 26.9% and 26% of the ascent time for their 75% and 90% loads, respectively. For the 100% load, Elliott et al., (1989) and Wilson et al., (1989), found the sticking region to last 28.8% of the ascent time. The 104% load displayed a sticking region that accounted for 40.8% of the total time. The maximum strength region for the 75% RM, 90% RM, 100% RM, and 104% RM, accounted for 21.5%, 40.2%, 31.6%, and 24.1% of total ascent time, respectively (Lander et al., 1985; Elliott et al., 1989; Wilson et al., 1989).

Finally, Lander et al., (1985) found the deceleration phase of their 75% RM load to comprise 26.5% of the total time, while their 90% RM load lasted

18% of the total ascent time. Elliott et al., (1989), and Wilson (1989) et al., (1989), found the deceleration phase of their respective 100% RM loads to consist of 23.3% of the ascent time. The 104% RM trial made up 13.9% of the total ascent time. It appears that the load had an effect on the duration of this phase. Heavier loads would require the subjects to accelerate the weight over a greater period of time to reach a peak in velocity and then slow down and therefore the length of deceleration was shorter. The 104% RM load invariably had a brief deceleration phase because subjects failed at successfully completing the lift and therefore the duration of the movement was shorter.

In a study comparing a free weight bench press and an isokinetic bench press, Lander et al., (1985), recorded the same characteristics previously described, for submaximal loads (75% and 90% RM). Conversely, Wilson et al., (1989), and Elliott et al., (1989) only found an acceleration phase and a deceleration phase for their 81% RM load (Figure 2-26). Given the experimental design of the Lander et al., (1985) study, these disparities may have been attributed to an order effect as well as a possible fatigue effect (i.e., subjects were required to perform 5 trials at 90% followed by 5 trials at 75%).

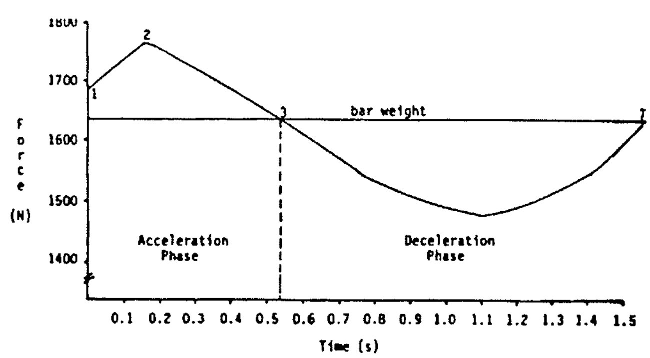

Figure 2-26 Force-time curve for the ascent phase of a typical submaximal load (e.g., 81% RM). (From Wilson et al., 1989, p. 396).

McLaughlin and Madsen (1984) found trends in bar acceleration for their 100% RM trial similar to the ones described earlier where four distinct sub-phases were recorded during the ascent (Figure 2-25). In the Elliott et al., (1989) study, the 81% RM load had very different patterns, where only 3 of 10 subjects exhibited similar patterns as the 100% load (i.e. four distinct sub-phases). The seven other subjects displayed only an acceleration and deceleration phase (Figure 2-26). The differences encountered in the Elliott et al., (1989) study, were due to the fact that the seven subjects produced much greater bar accelerations during the first phase of the ascent. These subjects spent 48.3% of the total movement time in the acceleration phase while the remaining portion consisted of decelerating the bar. The remaining three subjects resembled the bar patterns of the maximal load, displaying a small sticking region and maximum strength region.

Grip width in the bench press

Grip width has been shown to have an effect on bench press performance, bar path, and force exerted on the bar. In a study comparing the effect of grip width on the bench press performance of experienced weight lifters, Wagner et al., (1992a), stated that bar path was significantly affected by the three grip widths analyzed (95%, 165%, and 270% biacromial breadth). They found an inverse relationship between the grip width and the distance of the bar from the shoulder. As the grip width increased, the horizontal and vertical distances of the bar from the shoulder axis decreased (Figure 2-27). This decrease in distance produced a reduced moment about the shoulder axis in the sagittal plane, at five of the six technical aspects in the ascent phase of the bench press (Figure 2-27). (By assuming that the six points, 1-6, represent the line of force of the load, the changes in the length of the moment arm about the shoulder in the sagittal plane can be visualized by observing the mean path of the bar for these three grip widths).

The sticking (3) region began at a significantly higher point for the moderate grip (165%) than the wide grip (270%). The sticking region of the ascent encompassed a shorter relative time for the moderate grip (11.4%) than for either the narrower grip (17.3%) or the wider grip (22.5%) (Wagner et al., 1992a).

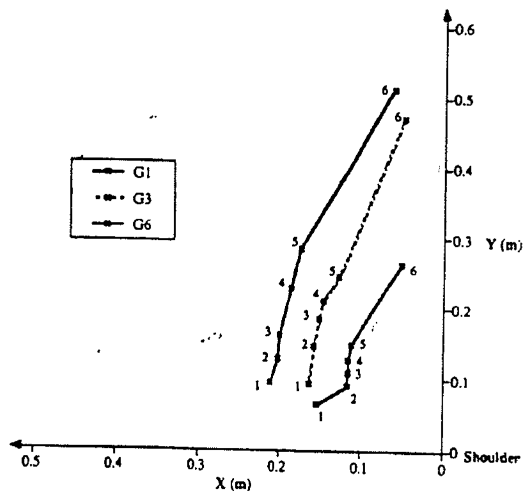

Figure 2-27 Mean bar path during the ascent phase of three different grip widths. G1 - 95%; G3 - 165%; and G6 - 270% of biacromial breadth. (From Wagner et al., 1992, p. 7).

Wagner et al., (1992a), also found the greatest bench press strength at the two moderate grip widths (165% and 200 % of biacromial breadth). No significant difference in bench press strength was recorded between the two narrow grips (95% and 135% of biacromial breadth) and the two wide grips (235% and 270% of biacromial breadth). The moderate grip width produced a small arm-to-trunk angle up to 90 degrees which increased the contribution of pectoralis major, thus permitting a greater force production at the beginning of the ascent. These results should consider the effect of training, since the strongest grip width tended to be the grip closest to the subjects' normal training grip (Wagner et al., 1992a).

In related study performed on novice and experienced female bench press athletes, Wagner, Evans, Weir, Housh, and Johnson (1992b), found that grip width significantly affected the bench press performance of experienced bench pressers, while it had no significant effect on bench press strength of novice lifters. Similar to the previous experiment, the grip width that produced the most significant increase in bench press performance for the experienced group was the grip that closely resembled their normal training grip (165% of biacromial breadth).

Harman (1984) conducted a three dimensional film analysis of the bench press exercise in an attempt to determine how grip width, weight, and movement speed affected kinematic and kinetic parameters. By using three dimensional data and an electronic force transducer (to measure forces on the bar), Harman (1984) found the narrow grip to produce the greatest mean peak elbow torque (60.6 Nm). Conversely, the wide grip produced the lowest mean peak elbow torque (22.7 Nm) and the highest mean peak shoulder torque (97.7 Nm). For each grip, the torque was greater with the heavier weights. Heavy lifts also showed a greater mean angle between upper arm and trunk (63.9 degrees) while light lifts had the smallest mean angle between upper arm and trunk (55.1 degrees). Finally, Harman (1984) found lift speed to affect mean shoulder torque, which was greatest for faster lifts.

<u>Motion of the bar</u>

Although the motion of the bar seems relatively simple (i.e., a vertical path), it is actually possesses a distinct excursion which seems to be dependent on the load (Lander, Bates, Sawhill, & Hamill, 1985; Wilson, Elliott, & Kerr, 1989; Elliott, Wilson, & Kerr, 1989), the grip spacing (Wagner, Evans, Weir, Housh, & Johnson, 1992a), and the skill of the bench presser (Madsen & McLaughlin, 1984; McLaughlin & Madsen, 1984). Several technical aspects have been noted and studied, in both the downward and upward phases of maximal and submaximal loads; as performed by elite and novice weight lifters (Madsen & McLaughlin, 1984; McLaughlin & Madsen, 1984; Elliott et al., 1989; Lander et al., 1985; Wilson et al., 1989).

Since the path of the bar is not completely vertical, it follows an arc like excursion during both the downward and upward phases. The angle of the arc has been shown to be different between skilled and novice bench pressers (Madsen and McLaughlin, 1984). The results of Madsen and McLaughlin (1984) indicated that during a maximal lift (i.e., 1 RM), the expert bench pressers moved the bar off the chest, towards the shoulder, at a 60 degree angle with the horizontal, while novice bench pressers used an 80 degree angle (Figure 2-28). Madsen and McLaughlin also noted that the novice lifters almost followed the same path during the descent, whereas the expert group did not. The expert group used a more vertical path during the descent than the ascent.

In a related study, McLaughlin and Madsen (1984) stated that bar displacement data (Figure 2-28) revealed that the heavy experts (mean body weight=1245.5 N) maintained a horizontal displacement further from the shoulder than did the light experts (mean body weight=775 N). Horizontal bar position throughout the ascent for the heavy experts (0.099 to 0.198 meters) actually resembled that of the novice group (0.132 to 0.189); the light experts recorded a horizontal displacement ranging 0.044 to 0.134. The greater the horizontal bar displacement from the shoulder the more pronounced (i.e., the less vertical) the angle of ascent and descent. However, the sequence of movement was similar to that of the light experts in Madsen and McLaughlin's study. Also, the heavy experts maintained a fairly similar mean angle of ascent and descent (62-66 degrees to the horizontal). The bar displacement differences noted between the heavy experts and the light experts was probably due to the fact that there exists a competitive restriction in grip spacing (81.5 centimeters). With a narrower grip (relative their body size), the heavy experts may have to displace the bar horizontally further from the shoulder rather than increasing elbow flexion.

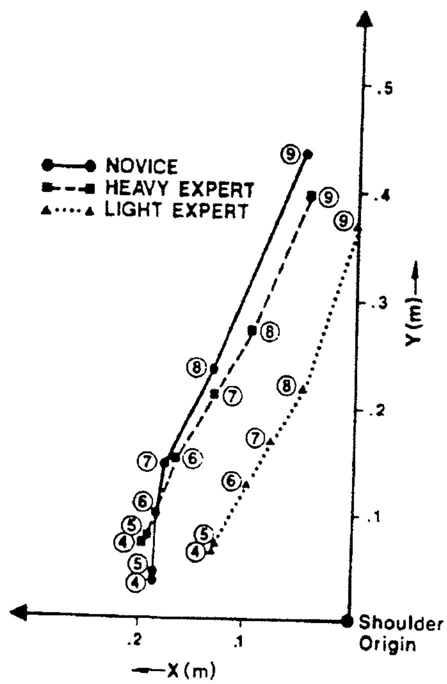

Figure 2-28 Comparison of mean bar paths between Madsen and McLaughlin (1984), and McLaughlin and Madsen(1984). (From McLaughlin and Madsen, 1984, p. 64).

Madsen and McLaughlin (1984) found that the heavy experts took a longer time to lower the bar (2.4 seconds) than the novice group (1.16 seconds)) and finished the movement with the bar closer to the chest; the light experts took 1.72 seconds to lower the bar. The light experts took a longer time to raise the bar than did the heavy experts or the light novices (2.21, 1.91, and 1.76 seconds respectively).

A study was conducted by Elliott, Wilson, and Kerr (1989) on 10 power lifters to analyze the ascent patterns of 3 different load conditions (81%, 100%, and 104% of RM). Elliott et al., (1989), recorded a consistent descent path over the three load conditions (17.8 degrees with the vertical from start to maximum downward velocity; and 24.1 degrees from maximum downward velocity to resting). These results were similar to those of Madsen and McLaughlin (1984), and McLaughlin and Madsen (1984).

The bar path during the ascent phase was different for all three load conditions. As the load increased, so did the horizontal displacement of the bar towards the shoulder. Horizontal displacement from chest to peak bar acceleration was 88.2, 60.1, and 57.3 degrees from the vertical, for the 81% RM, 100% RM, and 104% RM loads, respectively. From peak acceleration to peak velocity, the bar was displaced at 70, 65.8, and 51 degrees from the vertical, for the 81%, 100%, and 104% loads, respectively. Finally, the bar was displaced at the following angles (from peak velocity to end): 64.2, 65.8, and 10.3 degrees to the vertical for the 81%, 100%, and 104% loads.

The greatest horizontal bar displacement apparently occurs in the sticking region. Elliott et al., (1989), noted that the sticking region represented 28.8% of the total time and 36.5% of the total horizontal bar displacement (100% load). The sticking region for the 104% condition accounted for 40.8% of the total time and represented 57.4% of the total horizontal bar displacement. During the entire sticking region, horizontal displacement of the bar towards the shoulder axis was reduced by 38.5% for the 100% load and 31% for the 104% condition (Elliott et al., 1989). The increasing horizontal displacement observed during the ascent (especially in the sticking region) in the Elliott et al., study, was necessary to reduce the moment arm of the weight about the shoulder axis in the sagittal plane. Figures 2-24 and 2-28 illustrate how the moment arm decreased through bar ascent as it moved towards the shoulder (Figure 2-28).

In a study comparing maximal and submaximal efforts of expert bench pressers, Wilson et al., (1989), found the bar path of the ascent and descent to differ, especially for the heavier load where an increased horizontal displacement (ascent phase) towards the shoulder was noted; this consistent with the results of Elliott et al., (1989). Mean horizontal displacement for the 100% condition, during ascent, was 0.13 meters (which occurred mostly during the sticking region). For the 81% condition, mean horizontal displacement was only 0.09 meters. At the end of the lifts, the horizontal distance from bar to shoulder axis was significantly shorter for the 100% load (Wilson et al., 1989). In addition the most significant increase in horizontal bar displacement occurred during the sticking region. It appears that as the subject reaches the sticking, the bar is displaced more horizontally towards the shoulder in an attempt to reduce the moment of arm of the weight about the shoulder in the sagittal plane (Elliott et al., 1989).

Kinetics

Several investigators have used cinematography to estimate the forces exerted on the bar by the subject . The following formula has been used to calculate the instantaneous vertical component of force (F_y) from film (Wilson et al., 1989):

$$F_y\,(t)_i = ma\,(t)_i + W$$

where; $i = 1$ to n,

n = number of data points,
a = acceleration of bar (obtained from cinematographic data),

m = mass of bar, and

W= weight of bar.

The above formula provides a simple method of estimating the instantaneous vertical component of force. It is used because only one variable needs to be calculated (i.e., acceleration of bar), the remainder of the values are taken from the mass and weight of the load being manipulated. The values calculated from this equation can identify whether the vertical force being exerted against the bar is greater, equal to, or less than the weight of load. Information such as this can describe the force profile of a movement from start to finish and thus provide information regarding where greatest or least amount of force is being exerted during a particular movement (in this case, the bench press).

Typical force profiles for maximal and submaximal loads are illustrated in Figures 2-25 and 2-26. In the Madsen and McLaughlin study, comparing expert to novice bench pressers, a uniform force pattern (i.e. similar to Figure 2-25) was noted for the expert group. A greater difference was noted between the two groups in the position where minimum force was exerted than the position where maximum force exerted on the bar. The position where the maximum force and the minimum force were exerted, occurred at a horizontal distance of 0.062 meters and 0.096 meters further from the shoulder for the novice than the expert group, respectively. (Note that the normalized values for these horizontal distances were significantly different, yet they are not reported here because Madsen and McLaughlin failed to described how these values were normalized.) Madsen and McLaughlin (1984) attributed these results to either training specificity or a shift in bar path. Since the elite lifters used a bar path that was closer to the shoulder, it was suggested that this may have reduced the moments about the shoulder, when compared to the novice group.

In a related study, McLaughlin and Madsen (1984) observed that the heavy expert group (i.e., body weight of 1242.5 N) exhibited similar trends in force exerted on the bar when compared to the light experts. The average maximum and minimum forces exerted on the bar by the heavy experts during ascent was 2620.9 N and 2243.6 N compared to 2004 N and 1736 N for the light experts, and 1349 N and 914 N for the novice group, respectively. In addition, there was less difference between the point where the maximum and minimum forces were exerted on the bar during the ascent phase, between the heavy expert group and Madsen and McLaughlin's novice group. The point of maximum force exerted on the bar by the heavy exerts, the novice, and the light experts occurred at a horizontal distance (from the shoulder) of 0.198, 0.189, and 0.134 meters, respectively. The point of minimum force exertion occurred at a distance of 0.134, 0.174, and 0.078 meters for the heavy expert, novice, and light expert groups.

The novice group lowered the bar on average 1.18 seconds and 0.56 seconds faster than the heavy expert and the light experts, respectively. As a result, the novice group had to exert greater upward force to bring the bar to rest at the chest. Madsen and McLaughlin confirmed this result by noting that the expert group handled 79% more weight than the novice group, yet the difference in maximum force exerted while lowering the bar was only 43%. This result may have some implications for injury prevention, as a more controlled lowering of the bar is favored to a sudden burst of force to decelerate the weight.

Although the heavy experts lifted 30% more weight and had a 60% greater body weight relative to the lighter experts, the estimated torques at various points for the shoulder of heavy experts were twice the values of Madsen and McLaughlin's two lighter groups. The torque at the chest was 501

Nm for the heavy experts compared to 261 Nm and 255 Nm, for the light experts and the novice group. McLaughlin and Madsen (1984) attributed the large increase in the torque about the shoulder of heavy experts to: competitive restriction in hand grip spacing, bar displacement (i.e., bar was at a greater horizontal distance from the shoulder), and greater acceleration of the bar.

Elliott et al., (1989), investigated the biomechanical aspects of the bench press at three different loads (i.e., 81%, 100%, & 104% RM), and found different force patterns for all three conditions. The 100% load produced a consistent force profile for all subjects; Figure 2-26 illustrates a representative force curve for the 100% condition.

The 81% load was the most variable; only 3 of the 10 subjects exhibited a similar pattern to the 100% load. The seven subjects displayed bar accelerations and force patterns that consisted only of an acceleration phase and a deceleration phase. The absence of the other movment patterns displayed in the 100% RM (such as a sticking region and a maximum strength region) was a result of large forces relative to the bar weight during the acceleration phase (mean peak force of 112.1% of the bar).

Peak force at the failed condition (104% load) occurred at significantly closer vertical distance from the chest than the 100% load. Elliott et al., (1989), also noted that the position of peak force production during the maximum strength region occurred at a significantly closer distance to the chest for the failed load than the 100% condition (6.8 cm vs 16.1 cm). Furthermore, the position of maximal force exertion occurred at 2.08% of the vertical distance from the chest while the minimum force exertion occurred at 34.7% of the way through the movement. Minimum bar velocity occurred at 47.9% of the vertical distance from the chest. Peak force during the maximum strength region occurred 60.1% of the way through the ascent of the bar.

The failed attempts recorded by Elliott et al., (1989), produced a general movement pattern similar to the 100% load condition. Only 1 of 8 subjects who attempted the 104% load, achieved maximal vertical displacement during the sticking region. The other seven failed during the deceleration of the lift. The relative duration of the sticking region of the failed load to be significantly longer than the 100% load (40.8% vs 28.8%). The failed load also had a relatively shorter deceleration phase (13.9% vs 23.3%) (Elliott et al., 1989).

Wilson, Elliott, and Kerr (1989), using the same experimental protocol and analysis techniques as Elliott et al., (1989), compared the bar path and force profile of maximal (100%) and submaximal (81%) loads in the bench press. They found significant differences between the force profiles of the 81% and 100% load conditions for 7 of 10 subjects. Wilson et al., reported the same key features noted in Elliott et al., (1989), for the 100% load. This differed from Lander, Bates, Sawhill, and Hamill (1985) who found the same force profile for their maximal and submaximal conditions (i.e., 100% lift, 90%, and 75%). Given the nature of their experimental design, this could have been attributed to an order effect (i.e., 5 trials at 90% followed by 5 trials at 75%).

Lander et al., (1985), comparing free-weights to isokinetic bench pressing, found similar mean forces between the two lifting devices for both 75% and 90% loads (1138.12 N vs 1209.07 N and 1370.76 N vs 1439.78 N for free-weights and the isokinetic device, respectively). However, differences in force patterns were observed in the acceleration and deceleration phases. The forces recorded in the free weight condition during the acceleration phase was greater (for both loads) than the isokinetic condition (1600.63 vs 941.55 N and 1704.26 vs 1050.21 for the free-weight and isokinetic condition, respectively). The sticking region of the 75% RM and 90% RM loads also differed between the two devices where subjects exerted greater force on the isokinetic device

(1476.89 N vs 1101.66 N) than on the free-weights (1600.32 N vs 1229.36 N). Lander et al., (1985), suggested that these disparities were a result of the inherent differences of the two bench press styles. For example, subjects must generate muscular force to balance and stabilize the bar while this is not required on the isokinetic machine. In addition, the muscles may have been prestretched during the free-weight bench press as the subjects were allowed to lower the bar before pressing it upward.

The patterns of force production, however, were similar during the sticking region and the maximum strength region. All subjects produced large initial peak forces in the free weight condition. For the isokinetic condition, the first maximum forces occurred later in the range of motion. In addition, the magnitude of these forces varied in relation to second maximum values. The second maximum values were greater in the isokinetic condition. The fact that peak forces occurred later in the range (for the isokinetic condition may have some training implications in terms of increasing strength in the area where failure is most likely to occur. The isokinetic device may in fact allow a subject to improve strength during the sticking region and the maximum strength region (Lander et al., 1985).

Three different patterns of force production were recorded by Lander et al., (1985), for all six subjects. The first group exhibited a moderately large first maximum (1871 N) and a very large second maximum (2308 N). The second group exhibited a small first maximum value of 1672 N and an equally small second maximum (1652 N). The third group displayed a large first maximum value (2087 N) and a small second maximum (1648 N). Lander et al., noted that the above patterns related to the free-weight 1 RM capabilities of the subjects where the three groups had mean 1 RM values of 1689 N, 1522 N, and 1355 N.

EMG analysis of bench press

Only a few studies have used electromyography to assist in the understanding of the bench press and its related movements. Elliott et al., (1989) recorded moderate activity and on occasion noticed maximal activity in pectoralis major, triceps brachii, and anterior deltoid, throughout the bar descent. A distinct reduction in activity during the pause (i.e. chest contact) was noticed in all 8 subjects analyzed.

During the acceleration portion of the ascent phase (heavier loads), Elliott et al., (1989), recorded a large sustained increase in muscular activity from the pectoralis major, triceps brachii, and anterior deltoid. This activity continued throughout the remainder of the ascent. A slight delay in reaching maximal activity was noticed in triceps brachii. Biceps brachii displayed moderate activity at the onset of the acceleration phase and reached its highest level towards the end of the sticking region.

For the 81% condition, 7 out of 10 subjects produced moderate levels of activity during the acceleration phase; the activity greatly subsided during the deceleration phase. The other 3 subjects produced activation patterns similar to the heavier loads.

On the basis of their results, Elliott et al., (1989), suggested that since the moment arm decreased the most during the sticking region and the muscular activity from the prime movers was maximal and remained unchanged, one should reject the assumption that the sticking region is a result of an increase in resultant moment arm or a decrease in muscular activity. They proposed that the sticking point was caused by a poor mechanical force-producing position (possibly due to changes in muscle length as well as possible neural factors) combined with a gradual disappearance of elastic strain energy.

Wilson, Wood, and Elliott (1991) attempted to assess the neuromuscular contribution in the stretch-shortening cycle during the bench press mechanics of 12 experienced weight lifters. They found a significantly greater level of muscular activity in the deltoid and triceps brachii muscles during the concentric phase of the rebound bench press (i.e., subject lets the bar bounce off the chest in a controlled manner) compared to the non-rebound lift (i.e., a pause at chest contact). The muscular activity of the pectoralis muscles, however, did not change between movement conditions. Wilson et al., (1991), also stated that neither concentric or eccentric EMG of muscles tested was related to performance increase prior to stretch. Power spectrum analysis revealed that the stretching stimulus did not alter the firing characteristics of motor units. Wilson et al., (1991) concluded that performance of stretch-shortening movements such as the rebound bench press was primarily affected by the elastic strain energy of the prime mover muscles. Since their results did not show a neural change (i.e., increased firing) for the rebound bench press, it would seem futile for individuals to adopt this type of movement pattern in their training. It would therefore be advised (for training purposes and safety) that subjects use a pause before continuing with the ascent sequence of the bench.

In a recent study, McCaw and Friday (1994), calculated linear envelopes (IEMG) of the triceps brachii, anterior deltoid, medial deltoid, pectoralis major, and biceps brachii during the ascent and descent phases of two types of bench press, one using free weights and the other using a guided weight machine. Muscle activity collected during the performance of two loads (60% and 80% of 1-RM) on both conditions. The mean values over five trials were calculated for each muscle during descent and ascent for all four lifts. Mean IEMG activity for triceps brachii and pectoralis major was greater in the free weight condition (although not statistically significant) in three of the four phases (i.e., ascent and descent). The IEMG mean values for the biceps brachii were higher for the machine lifts during the descent phase, but higher for the free weights during ascent, (yet not significantly different). Although the sample size was small (n=5), the results of the study appear to support the notion that free weight training requires greater muscle activity than machine lifting. In addition, the significantly higher IEMG values for the anterior deltoid and medial deltoid muscles during the free weight condition suggested that these muscles play an important role in stabilizing the shoulder and assisting in pressing the load. Anterior deltoid and medial deltoid IEMG values were greater for free weights than for machine lifting (statistically significant at the low load).

Rosentwieg, Hinson, and Ridgway (1980) conducted an electromyographic comparison of three speeds of isokinetic bench pressing, as performed by 11 female volunteers. They analyzed the average absolute voltage elicited by the muscles during contraction and found a trend in which mean muscle action potential (MAP) increased with slower speeds. For three of four muscles (anterior deltoid, triceps, and biceps brachii), the moderate and slow settings elicited significantly greater MAP involvement than the fast setting. The pectoralis major, however, displayed a progressive decrease in MAP as the isokinetic speed setting increased.

Rosentwieg et al., (1980), concluded that maximum EMG output was achieved for the slower speeds and that the faster speed used in their study may have been too fast to allow for maximum resistance. The progressively significant differences in MAP encountered in pectoralis major were possibly attributed to muscle position in relation to bone, muscle action, greater stretch on muscle at starting position, and continuous movement at the shoulder (i.e., horizontal flexion) after completion of elbow extension. Rosentwieg et al., (1980) suggested that the extra movement may have caused nearby muscles to contract, which might have been picked up by the surface electrodes. Although accommodation (i.e., maximal resistance through a full range of motion) appeared to be a reasonable concept in varying isokinetic speeds (for three of the four muscles at two of the three speeds), it was not known to what extent the speed variations made a difference in the quantity of muscle activity. The fast setting may have been too rapid to provide maximum resistance, and therefore the muscles could not be maximally activated.

Hinson (1969) conducted a study on four different types of push-ups to determine which muscles (i.e., triceps brachii, deltoid, pectoralis major, trapezius, serratus anterior, rectus abdominus, and external oblique), were involved in these exercises, as well as to investigate the existence of patterns of muscular contractions among them.

Hinson (1969) found all seven muscles to be most active during the upward portion of the push-up for the full, bench and knee push-ups. Anterior deltoid was the most active muscle during all the exercises. The order of involvement from most to least was as follows: triceps brachii, trapezius, clavicular pectoralis, serratus anterior, sternal pectoralis, middle deltoid, rectus abdominus, and external oblique. Muscle involvement for the bench push-up and the full push-up did not differ substantially. Hinson (1969) stated that the let-down push-up (i.e., downward or eccentric portion of a full push-up) was somewhat comparable (but less) to that of the knee push-up. On the basis of her data, Hinson (1969) suggested an order of difficulty for the exercises: let-down, knee push-up, bench push-up, and full push-up.

EMG of pectoralis major and related shoulder muscles

Only a few EMG studies have been conducted on the pectoralis major. The first to study the pectoralis major using EMG was Inman, Saunders, Abbott (1944). Their results indicated that in abduction, no part of the pectoralis muscle was active. In forward flexion, the clavicular head was most active, reaching two peaks: one at 75° of forward flexion and the other at 115°, with the activity being greatest at 115° of forward flexion. Also, the manubrial part of pectoralis major showed slight activity during forward flexion (a peak at 110°) while the sternocostal head remained inactive (Figure 2-29).

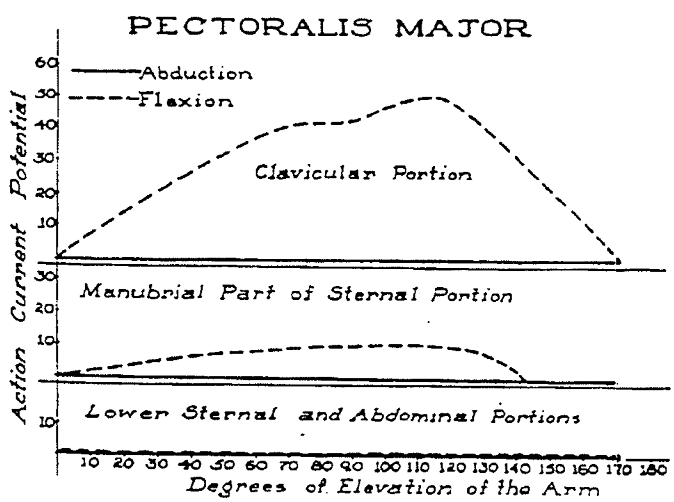

Figure 2-29 Activity of pectoralis major during forward flexion. (From Inman et al., 1944, p. 17).

Scheving and Pauly (1959) confirmed the findings of Inman, and also noted that both clavicular and sternocostal portions of pectoralis major were active during adduction. They also stated that pectoralis major did not contract in medial rotation, except when movement was performed against resistance. Conversely, Basmajian and DeLuca (1985), cited a study by DeSousa et al., (1969) that found the clavicular head of pectoralis to be active almost always during rotations without and against resistance, but agreed with the other findings regarding the sternocostal head. Using needle electrodes, Shevlin and Lucci (1969) reported functional differences between the two heads of pectoralis major. The clavicular head was most active at 0° of isometric abduction and then lost power at 90°. Finally, Zuckerman and Matsen, (1989), noted that while performing shoulder forward flexion, the pectoralis major was more active during scapular elevation than during depression.

Studies on the deltoid have found its three portions to be active in all movements of the shoulder joint (Zuckerman & Matsen, 1989; Scheving & Pauly, 1959). Scheving and Pauly (1959) also suggested that although one part acts as the agonist or prime mover, the other part stabilizes the joint by contracting. In abduction, the deltoid exhibited its greatest activity between 90-180° of abduction, while in forward flexion, it peaked and reached a plateau between 110-130° (Inman et al., 1944). Shevlin and Lucci (1969) noted that anterior deltoid showed the most action in forward flexion. Zuckerman and Matsen (1989) noted that the deltoid showed an almost linear increase in muscular activity with an increasing angle of forward flexion (up to 140°). In addition, the deltoid was much more active when in scapular elevation than when the scapulae were depressed.

Hermann's (1962) EMG shot put study revealed the anterior deltoid to be active throughout the entire skill, with the greatest contraction occurring at the thrust phase of the shoulder and arm. Kamon (1966) conducted an EMG study on static and dynamic postures of the body supported on the arms. The postures were conducted on a pommel horse; the static postures included a front support and straddled support, and the dynamic posture consisted of a continuous leg swinging under each hand in turn. Electrical activity was the greater for the dynamic movements than for the static positions of the same muscles. Also, the anterior deltoid was found to be one of the most active muscles in both the static support and the dynamic support movements.

Using power spectrum analysis, Gerdle, Eriksson, Brundin, Edstrom, (1988b), found the anterior deltoid and infraspinatus to have significantly higher mean power frequencies at 90 degrees than at 45 degrees of maximum static shoulder flexion; the mean power frequency at 65 degrees for anterior deltoid was also significantly higher than the 45 degree value. The trapezius and the long head of biceps brachii did not show any differences in mean power frequency throughout the three angles tested. (Note that the use of the mean power frequency is one of parameters from the frequency spectrum of a signal that can be used to assess changes in EMG).

In a related study, Gerdle, Eriksson, and Hagberg (1988a), compared the effect of increasing isometric shoulder forward flexion (65 degrees) on the mean power frequency of EMG signals from the trapezius, anterior deltoid, biceps brachii and the right infraspinatus. Gerdle et al., (1988) did not find a significant correlation between mean power frequency (MPF) for the four muscles and the torque. In fact, a gradual decrease in MPF was noted after the 6th second. This may have been a result of a decreased conduction velocity of the muscle fibers due to muscular fatigue.

The function of the shoulder muscles during the various phases of the tennis serve, the backhand, and the forehand ground strokes, has been studied by Ryu, McCormick, Jobe, Moynes, and Antonelli (1988). Their study reported pectoralis major along with supraspinatus, and infraspinatus to be the most active during the serve and the forehand. Both the pectoralis major and the serratus anterior reached their peak activity during acceleration (serve - 115% MVC and 113% MVC; forehand - 85% MVC and 102% MVC, respectively); infraspinatus was most active during the cocking stage (41%

MVC) of the serve and the follow-through of forehand (40% MVC). During the backhand the middle deltoid, supraspinatus, and infraspinatus were most active in the acceleration (118%, 73%, and 78% MVC, respectively) and follow-through (48%, 41%, and 48% MVC, respectively) stages of the backhand. The results indicated that the anterior muscles involved with the shoulder complex were extremely important in providing the necessary forces to accelerate the limbs as well as provide stability. Therefore, an exercise program designed to strengthen these muscles, especially the stabilizers, was recommended.

Glousman, Jobe, Tibone, Moynes, Antonelli, and Perry (1988), found different relative muscular activity, during a baseball pitch, between the shoulder muscles of healthy males and the muscles of subjects diagnosed with glenohumeral instability. The deltoid was the only muscle in which both groups compared in activity patterns. No difference, in terms of relative muscle activity, between the two groups was noted during the wind up and early-cocking events. During late cocking, the supraspinatus, pectoralis major, subscapularis, and the serratus anterior exhibited differences in muscle activation; the supraspinatus of the injured group was more active. The injured group also had greater activation in the biceps during the acceleration phase. The relative activity of subscapularis, latissimus dorsi, and serratus anterior was significantly greater in the healthy group. The pectoralis major, subscapularis, latissimus dorsi, and serratus anterior of the healthy group were significantly more active during the follow-through than the injured group. The identification of shoulder muscle imbalances suggest a strategy for postoperative rehabilitation, where the focus should be on correcting the pathology by way of strength and flexibility training.

Physical factors important to bench press performance

There are numerous studies that have been conducted to determine the effect of various anthropometric variables on bench press performance, or to determine the relationship between absolute muscular endurance and bench press strength. While a detailed review of these studies is beyond the scope of the present study, the relative importance of these results to bench press performance warrants a brief discussion.

Wagner et al., (1992a) suggested the following factors as being important for bench press performance of males: muscle mass, arm-to-trunk angle, arm-forearm angle, and training experience. In related a study, Wagner et al., (1992b), found both grip widths and forearm length as important factors in the bench press performance of experienced female weight lifters. Finally, Mayhew, Ball, Ward, and Arnold, (1991), tested upper arm and chest circumferences, upper and lower arm lengths, shoulder and hip widths, % fat, height, upper arm cross-sectional area, and drop distance (measured by taking the distance between the bar and the pectoral muscles at full elbow extension) to determine which of these characteristics best predicted bench press strength in college males. Mayhew et al., (1991), found upper arm cross-sectional area and chest circumference as the best predictors of bench press strength.

Berger and Smith (1991) conducted a study on four males and four females to determine the effect of the tonic neck reflex (i.e., head ventroflexion, neutral, and dorsiflexion) on power (Power = $f\,d\,t^{-1}$; where f is bar weight in Kg, d is 0.25 meters, and t is time in seconds) in the bench press. Analysis of the data revealed significant differences ($p<0.05$) among head positions between the sexes, and interaction. For men, ventroflexion was significantly less powerful than the neutral and dorsiflexion positions. Women, however, did not follow the same trend; the head dorsiflexion position was significantly more powerful than the ventroflexion and neutral positions.

Berger and Smith (1991) found the ventroflexion and neutral positions for women had similar power. No significant differences between the neutral position and dorsiflexed positions were recorded for the males. On the basis of their results, Berger and Smith (1991) concluded that the tonic neck reflex may augment force and power by providing afferent input by way of the neck proprioceptors. They also stated that at the same time facilitation occurred, the tonic neck reflex provided an inhibitory response in antagonistic muscles. Finally, Berger and Smith (1991) suggested that the effector muscles provided additional afferent stimulation that further enhanced the effect of the tonic neck reflex.

The results of Berger and Smith (1991) may warrant consideration, especially during a one repetition maximum. To maximize bench press performance, subjects may consider an optimal head position. Given the competitive bench press restrictions, the dorsiflexed position would not be recommended as it may raise the shoulders off the bench thereby making the lift an illegal one. For this reason the neutral head position is recommended for improved bench press performance.

Weir, Wagner, and Housh (1994) examined the effect of rest interval length on repeated one repetition maximum (1-RM) bench press performances. Once the 1-RM was determined for the bench press, the 1-RM was attempted twice during four separate testing sessions. Rest intervals of 1, 3, 5, or 10 minutes were given between attempts. Seven failed attempts were recorded: four at one minute of rest, two at five minutes, one at ten minutes, and none at three minutes of rest. Based on a Cochran Q test, rest interval length did not appear to affect the ability to repeat a successful 1-RM bench press ($p>0.05$).

The results of the Weir et al., (1994), study suggest that rest intervals as short as one minute may not impair the ability to reproduce maximum bench press performances and therefore may not compromise the validity on 1-RM values when testing for strength.

Several investigators have sought to investigate the relationship between absolute and relative muscular endurance to bench press performance. Rose and Ball (1992) found a strong positive relationship (r=0.81) between absolute muscle endurance and bench press performance in college women. Relative endurance (i.e., number of repetitions completed in one minute) was also found to be a reliable predictor of bench press performance. Mayhew, Ball, Arnold, and Bowen (1992), found very high correlations ranging from a low value of r=0.90 (women) to a high value of r=0.95 (college football players). Similarly, Prinster, Mayhew, Arabas, Ware, and Bemben (1993) found relative endurance performance as a good indicator of bench press strength. Prinster et al., (1993), found that the two equations tested produced very high validity coefficients (r=0.96 and r=0.97 for the Mayhew equation and the Welday and Lander equation, respectively); yet the Welday and Lander equation over predicted actual bench press strength by 3-5%. The Mayhew equation did not significantly under or overestimate bench press performance.

CHAPTER 3
METHODS

Introduction

The purpose of the study was to determine the effect of two different modes of bench press on muscle activity, specifically to determine if the incline bench press increased activation of the clavicular fibers of the pectoralis major more than the supine bench press. The study also attempted to determine if the incline bench press imposed a greater demand on anterior deltoid and medial deltoid than the supine bench press. A null hypothesis of no difference in muscle activation of the clavicular and sternal portion of the pectoralis major muscle, the anterior deltoid, and the posterior deltoid muscle during the three angles of the incline and the supine bench press was adopted.

Subjects

Fourteen male college volunteers (19-27 years old) served as subjects for the study. They were recruited by posters located in the University of Manitoba, Frank Kennedy Building weight room and by personal communication with the author. This number of participants was chosen to increase the external validity of the study, as well as account for an expected attrition of 10%.

The subjects met the following criteria: a minimum of two years of weight training experience and the ability to bench press at least 125% of their own body weight. These criteria defined the subjects as experienced weight trainers. These criteria were chosen because individuals who were able to press 125% of their body weight possessed the necessary technique to perform the repetitive bench press with some consistency.

After initial telephone contact, the subjects were contacted by the investigator at least two weeks prior to the initial testing date. Subjects were then given a consent form and a written description of the testing procedure. The consent form included a description of the testing procedure, the guarantee of confidentiality, and the assurance that they had the right to withdraw from the study at any time (Appendix A).

Subject characteristics such as height, weight, % fat, age, grip width, and the 1-RM maximum for the three angles (37^{o}, 43^{o}, 50^{o}) of the incline bench press and the supine bench were measured at five different testing sessions prior to the collection of the EMG data. The sessions were divided to ensure that subjects were sufficiently rested for the four maximal bench press tests.

Testing protocol

Prior to executing the incline and supine bench press exercise during the actual data collection, the subjects performed a series of bench presses to determine their 1-RM load (Kg) for the two movements. The four maximal bench press tests were performed on separate days, with a minimum of 96 hours between sessions to reduce any fatigue effect, as well as minimize the risk for injury. The subjects were instructed in the 1-RM test protocol prior to their test and were also asked to refrain from vigorous activity for at least 48 hours prior to the 1 RM tests. Subjects were scheduled in groups of three. This assisted in terms of spotting and was more time efficient. The subjects were also given as much time as required to warm up before the testing began. All subjects achieved their 1-RM in 6-7 attempts. A 5 minute rest period between lifts was given. Grip width was measured and recorded at the onset of each of the four 1-RM tests so that the same grip width was used during the submaximal tests in which the EMG data was recorded. Grip width was measured as the distance between the index fingers as the hands gripped the bar.

The sequence of events in both types bench press 1-RM tests was adapted according to the protocol described by Wagner et al., (1992). The subjects lay supine or inclined with head and trunk extended on the bench and feet on the floor. The subjects received a lift off from the weight rack by a spotter, after which the bar was slowly lowered to the chest by the lifter. The spotter followed the bar very closely to provide assistance in case of a failed attempt. When the bar was motionless on the chest (0.5 seconds), the investigator then gave the signal to start pressing. After the signal, the bar was pressed vertically and evenly to straight-arm length and held motionless (1 sec) until the investigator gave the signal to replace the bar. The bar was loaded in 2-5 kg increments until the subject failed at the attempt. If the subject failed an attempt (i.e., could not press the weight), they were given a second attempt to ensure that they had reached their maximum on the previous lift.

Once the 1-RM was established for both exercises, a 75% RM load was calculated and used for the remainder of the investigation. The load was selected because it closely resembled a typical training load employed by individuals when training for strength and it was within the recommended range for strength training (McGlynn, 1990; Schmidtbleicher, 1985; Sale & MacDougall, 1984). All 14 subjects were tested in 3 testing sessions. The order of performance in terms of inclination (i.e., 37°, 43°, and 50°) and supine bench press was randomized for each subject to help control for possible order effect.

The subjects were tested individually and performed a total of 3 repetitions of the bench press movement at each angle of inclination, at a pace dictated by a metronome which was set at 45 beats per minute (i.e., 1.333 second intervals). The metronome pace was selected to mimic an actual exercise pace. This was determined by assessing the movement pace of several subjects during a typical exercise bout. The pace dictated by the metronome was necessary to ensure that the exercises were performed consistently and to maintain a constant velocity. At the time of data collection, subjects were given an orientation to the movement pace set by the metronome and were allotted two practice trials at a light load (50% 1 RM); this also served as a warm up. Once a subject finished the first bench press condition, he performed the second protocol, then the third condition, and finally the fourth bench press movement. Two different benches were used: a supine bench and a variable incline bench. This enabled the investigator to keep the testing conditions constant for that subject. A 5 minute rest period following each test was given to ensure recovery. This amount of rest was given to minimize the effect of fatigue so as not to compromise the validity of the data (Weir, et al., 1994). Subjects were also instructed to refrain from any strenuous exercise for at least 48 hours prior to the 75% RM test.

Electromyographic recording procedures

Surface electrodes (8 mm diameter) were attached over the muscle belly and oriented parallel to the active muscle fibers (with a between electrode distance of approximately 1 cm) of the right anterior deltoid, the right middle deltoid, the right clavicular portion of pectoralis, and the right portion of sternal pectoralis (Medico test silver-silver chloride bipolar blue sensor disposable, type N-00-S). Since the subjects for the present study were experienced and healthy, it was assumed that no asymmetries in muscle activity between the left and right sides existed, hence the electrode placement on one side of the body only. The ground electrodes were placed on the inferior portion of the sternum. The electrode positions were marked on the skin using a water soluble pen, in case an electrode loosened during the test and had to be replaced.

The electrodes were connected to four small BioSys battery powered differential preamplifiers with an input impedance of 44 MOhms and a common mode rejection ratio of 95 dB at 60 Hz. The differential preamplifiers were affixed anteriorly to the right and left trapezius muscle, and laterally on the right upper arm. The preamplifiers were individually connected to the main amplifier. The EMG signal was amplified using a BioSys battery powered amplifier at a high pass frequency of 20 Hz and a low pass frequency of 3 KHz each with a roll-off of 6 dB per octave (first order RC filter). The gain of the EMG was adjusted to achieve a high signal-to-noise ratio without clipping. Gains of 500-2000 were used in all trials. The signal was processed using full wave rectification into a low pass filter of 30 Hz. (Note: equipment specifications are given in Appendix B.)

To further reduce the risk of artifact and noise pickup, the wires leading from the electrodes to the preamplifier were twisted and taped to the subject's body (Kriellaars, 1994), and approximately half of the fluorescent lights in the laboratory were turned off.

Filming procedure

Filming occurred simultaneously with the collection of EMG data and took place in the Motor Control Laboratory at the School of Medical Rehabilitation, at the University of Manitoba. All subjects were filmed using a Panasonic Digital 5100 video camera and a Sony V-801-CCD video camera at a speed of 60 frames per second, using a shutter speed of 1/500. Cameras were placed at 90 degrees to each other so that the Panasonic Digital 5100 filmed the movements in the transverse plane (i.e. behind the subject's head) and the Sony V-801-CCD filmed the movements in the sagittal plane (i.e. opposite the subject's right side). Both cameras were placed at distances that ensured full view of the entire movement. The two cameras were used to assess bar displacement (sagittal plane) and to determine that the bar was pressed evenly (transverse plane) with both arms throughout the movement protocol. The Panasonic Digital 5100 and the Sony V-801-CCD were connected separately to two video cassette recorders. The entire system and connections are illustrated in Figure 3-1.

The scaling factor for both two dimensional views was determined by digitizing a black rod with silver reflective markers placed at a known distance at each end of the rod. Two flood lights were used (one for each plane of view) to help illuminate the reflectors. The calibration rod was digitized in both camera views.

EMG data collection and processing

The myoelectric activity of the anterior deltoid, middle deltoid, sternal pectoralis, and clavicular pectoralis was collected on-line for a total duration of 25 seconds, for each three repetitions, at a sampling rate of 333 Hz per channel by a 486 IBM computer equipped with a 16 channel, 12 bit analog-to-digital converter (See Appendix B for specifications). Using Electronic Chart Recorder software (R. C., Electronics, Inc., 1992), the linear enveloped EMG was graphically displayed on a EGA graphics monitor. The EMGs were inspected on the computer screen. Trials were repeated if artifact was evident or if the subject failed to follow the movement pace set by the metronome. The signals were then stored on disk for further analysis.

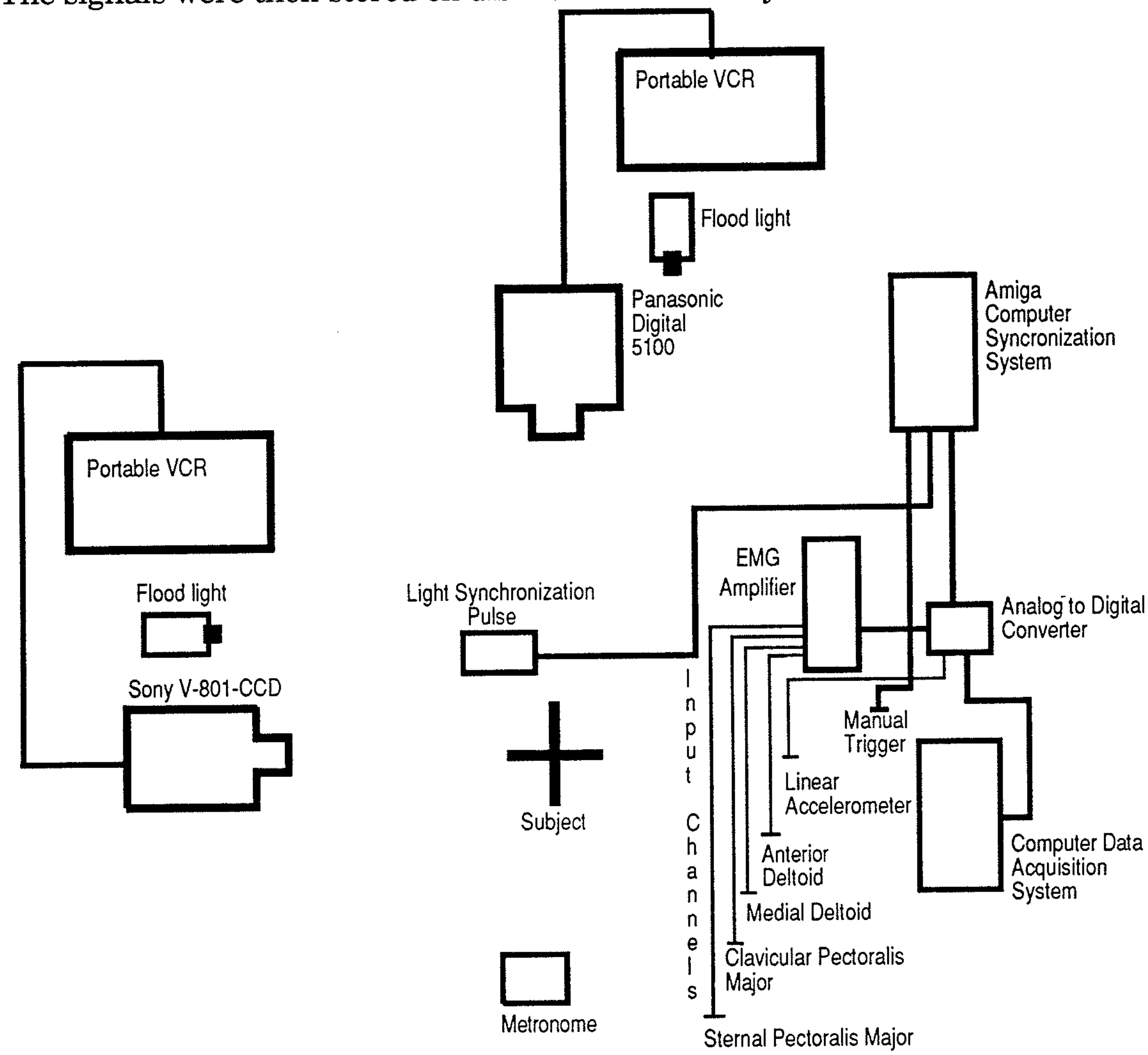

Figure 3-1 Schematic diagram of the experimental set-up

An ICSensors (type 3021) linear accelerometer was mounted to the bar and oriented perpendicular to the gravitational vector. The accelerometer was connected to the computer that was collecting the EMG data. The data was collected digitally in parallel with the EMG data at a sampling rate of 333 Hz. The accelerometer data facilitated the definition of each repetition, as well as the eccentric and concentric phases of the lift; which in turn was used to determine the onset of the ascent and descent phase.

Finally, both the EMG acquisition and video recorders were time locked. The synchronization occurred at the onset of each movement condition. Once the investigator gave the subject the verbal command "start", the synchronization signal was initiated, by a keystroke and a small trigger pulse was sent to the computer collecting the EMG data. The onset of the pulse determined the start of the data collection for the movement protocol. At the same time the pulse was generated, the synchronizing computer switched a light on that was placed in the field of both camera views. The onset of the light was used to determine the start of the video data collection. Movement on the bench press would occur about 1.0 seconds after the verbal command based on the metronome signal pacing.

EMG data analysis

Prior to further analysis, the linear enveloped EMG data was verified graphically by printing each trial on paper and examining each of them to ensure that none of the data was missing and that all data was valid. Once this was done, all the RC files generated by data acquisition software (RC Electronics Inc., 1992) were converted to ASCII text files using a customized program called Mkbai (School of Medical Rehabilitation, 1994). The raw accelerometer data (Figure 3-2) was smoothed using a 2nd order recursive Butterworth digital filter set at a frequency of 5 Hz. Data was then loaded into a second customized program (Chan3, School of Medical Rehabilitation, 1994) that separately displayed the smoothed accelerometer and linear enveloped EMG data for each muscle. The smoothed accelerometer data was then used to define start and stop times and the eccentric and concentric phases for each repetition for the four trials (Figure 3-3).

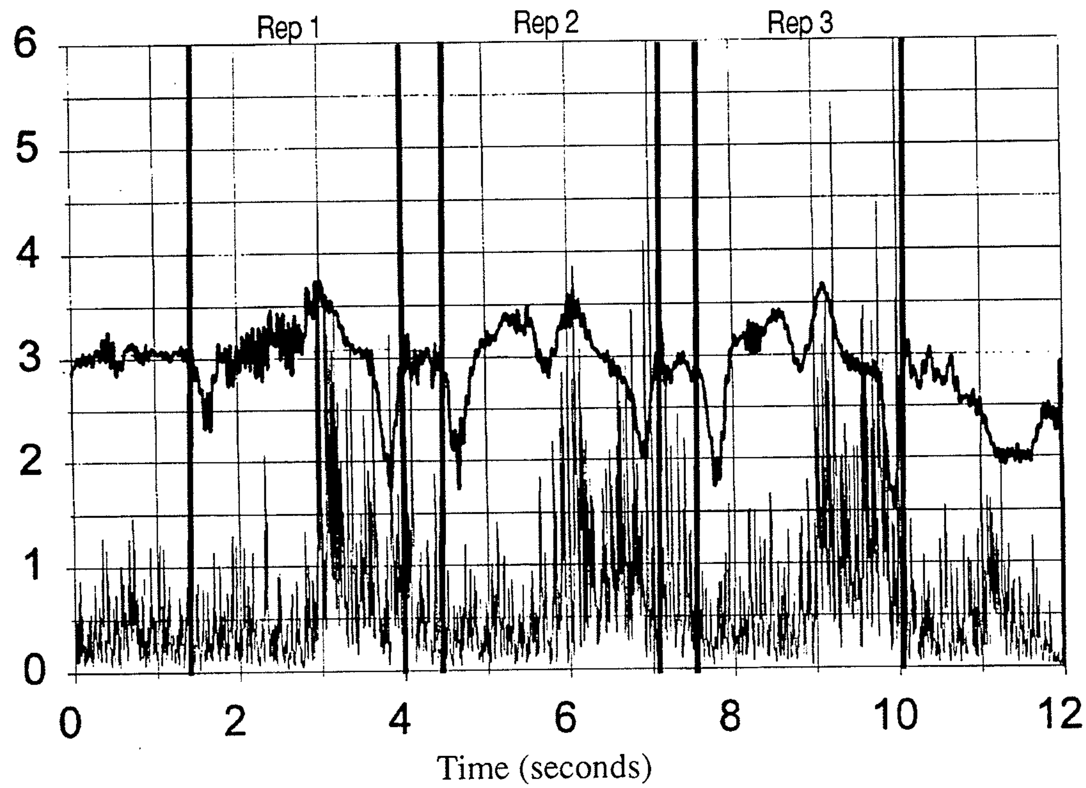

Figure 3-2 Example of enveloped EMG (raw) data of anterior deltoid along with raw accelerometer tracing (i.e., dark line above EMG tracing).

Once the start and stop times were manually identified for all repetitions by scrolling through the data (and for both ascent and descent phases of the movement), a custom batch analysis program, (School of Medical Rehabilitation, University of Manitoba), calculated the sum (for a fixed duration of approximately 0.950 seconds) of EMG activity (volts) during the eccentric and concentric phase for each repetition during all four movement conditions. This information was then compiled onto a spread sheet where the values for the sum of EMG activity (volts) in both eccentric and concentric phases (for 3 repetitions for the four conditions) were listed. The sum of EMG activity (volts) for the 3 repetitions were then combined to give an average concentric and eccentric value for each of the four movement conditions for each subject. In other words, the total activity (i.e., sum) in the eccentric and concentric phases for each repetition was calculated and then an average value (AEMG) for each of two phases for the particular movement condition was determined by adding the summed values (volts) for each eccentric and concentric portion of the repetition and dividing by the total number of repetitions (3):

<u>*Eccentric phase:*</u>

$$AEMG = (\Sigma EMG_{rep1} + \Sigma EMG_{rep2} + \Sigma EMG_{rep3}) \div 3$$

<u>*Concentric phase:*</u>

$$AEMG = (\Sigma EMG_{rep1} + \Sigma EMG_{rep2} + \Sigma EMG_{rep3}) \div 3$$

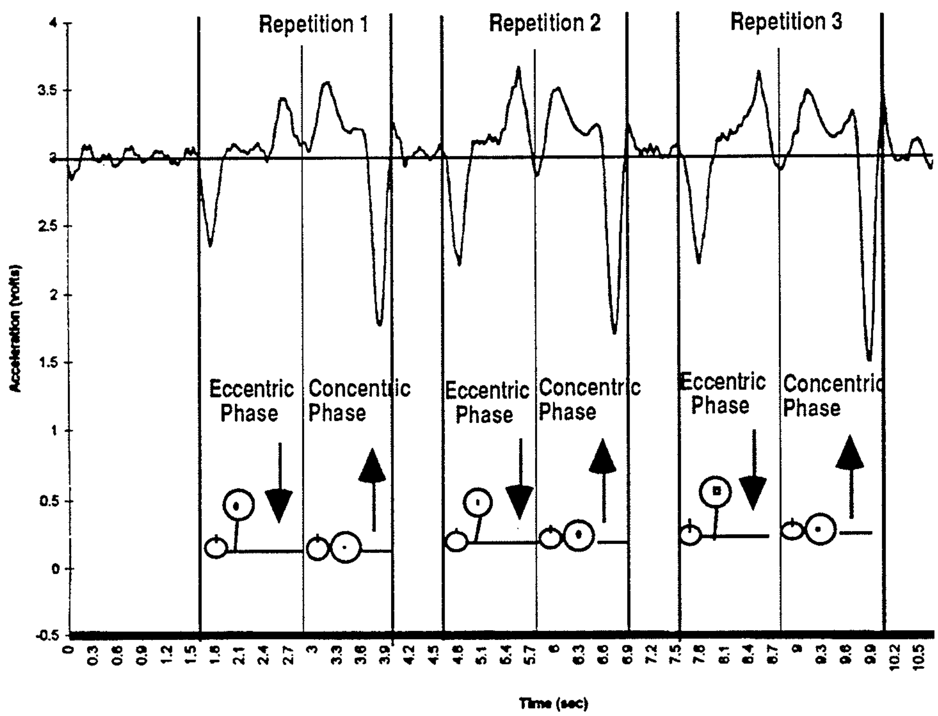

Figure 3-3 Example of smoothed accelerometer data used to define (1) start and stop times and (2) the eccentric and concentric phases for each repetition. (Note: vertical line through accelerometer tracing represents 0 acceleration.)

Finally, the mean EMG data for the four muscles (eccentric and concentric) was normalized and expressed as a ratio of the mean EMG acquired during the supine bench press (i.e., changes relative the supine bench press) :

E.g., Eccentric phase: $$\frac{AEMG_{37^{\circ}\text{ at 75\% load}}}{AEMG_{\text{supine at 75\% load}}}$$

E.g., Concentric phase: $$\frac{AEMG_{50^{\circ}\text{ at 75\% load}}}{AEMG_{\text{supine at 75\% load}}}$$

Film data analysis

The video frames of each trial were viewed using a specialized Panasonic AG-7300 video cassette recorder and a Sony Trinitron monitor. The trials were reviewed several times to determine the exact time the synchronization light turned on. A Peak Performance Technologies (1992) video digitizing system was used to acquire data for motion analysis.

The spatial model used in the present study consisted of 4 individual points which produced a 1 segment model (Figure 3-4). A spatial model can be defined as a computer simulation or representation of the subjects performing the skill. The segment masses used in the spatial model were adapted from Humanscale (Diffrient, Tilley, & Bardagjy, 1978), while the segment center of gravity distances were taken from Hinrichs (1988). Markers were placed on the right hip, the right shoulder, the bar, and on the weight rack.

Segments used for digitizing	Point(s)
Reference point	1
Bar	2
Trunk	3-4

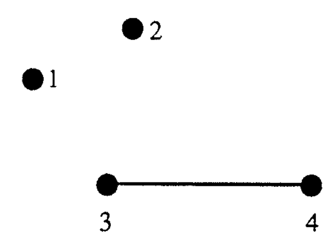

Figure 3-4 The spatial model that was used for the present study. Segments used are indicated on the left. The numbers correspond to joint locations that will be digitized.

The trials were divided into successive frames and each digitized coordinate was entered into the computer. The trials were digitized so that each repetition had a descent phase (i.e., eccentric) and an ascent phase (i.e., concentric).

The raw coordinate data for the present study was smoothed using a 2nd order recursive Butterworth digital filter at frequency of 5 Hz. Although the sampling frequencies were different for the linear accelerometer (333 Hz) and the video data (60 Hz), the acceleration data calculated from both showed very similar patterns (Appendix C).

Statistical analysis

Once all of the EMG data for the anterior deltoid, medial deltoid, clavicular pectoralis, and the sternal pectoralis was normalized with respect to the supine bench press condition for each subject, the normalized data for each muscle during both eccentric and concentric phases for the three incline bench angles and the supine bench press were compared.

All statistical analyses were performed on an Macintosh LC 475 computer using a statistical software package (StatView, version 4.0 for Macintosh). A univariate repeated measures ANOVA was used to compare the mean normalized muscle activity of the clavicular pectoralis, the sternal pectoralis, middle deltoid and the anterior deltoid during four types of bench press (Appendix E): the supine bench press and during the three different angles of the incline bench press (37^{o}, 43^{o}, 50^{o}). Significance was set at the $p < 0.05$ confidence level. A Tukey's multiple comparison test was used to determine where the differences occurred (Hassard, 1991). (Note that no comparisons between eccentric and concentric phases were done because these factors are separate types of contractions, and therefore are very different from one another).

CHAPTER 4

RESULTS

Subject characteristics are presented in Table 4-1. Mean 1-RM values for the supine bench press (Table 4-1), the 37°, 43°, 50° incline bench presses were 128 kg (± 14.39 kg), 107.76 kg (± 12.969 kg), 105.86 kg (± 12.836 kg), and 101.29 kg (± 11.232 Kg), respectively. As the angle of inclination increased, the 1-RM decreased ($p < 0.0001$). In addition, the mean maximum load lifted in the supine bench press corresponded to 156% of the average body weight of the subjects, indicating that the subjects were well trained. A sample ANOVA table may be seen in Appendix E.

Table 4-1 Subject characteristics and 1 repetition maximum (RM) for each exercise.

Subject	Age (yr.)	Height (cm)	Mass (Kg)	% Fat	1RM Flat (Kg)	1RM 37° (Kg)	1RM 43° (Kg)	1RM 50° (Kg)
1	24	186.25	86.95	6.66	115.90	106.81	102.27	93.18
2	22	172.30	93.10	12.13	135.00	115.91	113.96	102.27
3	27	164.00	66.80	5.75	110.00	93.18	93.18	90.90
4	22	166.00	73.10	16.00	117.50	90.90	88.63	88.63
5	23	166.00	74.95	7.04	132.50	109.09	106.80	109.09
6	25	172.30	81.05	17.64	132.50	111.36	107.50	106.81
7	25	173.90	88.00	11.79	132.50	111.36	105.22	104.54
8	24	181.00	80.95	9.27	117.50	102.27	100.00	95.45
9	21	176.00	74.55	10.07	110.00	97.27	93.18	90.90
10	20	178.60	89.45	7.76	120.00	100.00	100.68	95.45
11	21	168.30	92.05	15.94	130.00	125.00	122.95	115.91
12	22	183.80	92.75	8.15	162.50	136.36	129.54	127.27
13	23	171.50	71.95	9.13	145.00	115.91	125.00	106.81
14	19	178.00	86.85	13.25	132.50	93.18	93.18	90.90
Mean	22.71	174.14	82.32	10.75	128.00[a]	107.76[b]	105.86[b,c]	101.29[c]
SD	2.16	6.84	8.75	3.80	14.39	12.96	12.83	11.23
					p<0.0001	p<0.0001	p<0.0001	p<0.0001

a, b, c - means are significantly different

Grip width (Table 4-2) used by the subjects for the supine bench press, 37°, 43°, and 50° incline bench presses were 70.35 (± 5.895), 69.61 (± 4.231), 70.67 (± 4.885), and 69.24 (± 4.868), respectively. The differences were non-significant.

Table 4-2 Grip width and 75% RM values for the flat and incline bench presses.

Subject	75% RM Flat (Kg)	75% RM 37° (Kg)	75% RM 43° (Kg)	75% RM 50° (Kg)	Gripwidth Flat (cm)	Gripwidth 37° (cm)	Gripwidth 43° (cm)	Gripwidth 50° (cm)
1	86.93	80.11	76.70	69.89	79.0	76.0	78.0	77.0
2	101.25	86.93	85.40	76.70	73.0	64.5	65.0	63.5
3	82.50	69.89	69.89	68.175	62.0	64.0	63.0	63.5
4	88.13	68.18	66.47	66.47	69.0	72.0	68.7	67.4
5	99.36	81.82	80.11	81.82	71.5	71.0	72.5	74.2
6	99.38	83.52	80.63	80.12	64.5	65.0	65.0	65.5
7	99.37	83.52	78.92	78.41	73.0	73.5	73.0	74.0
8	88.13	76.70	75.00	71.59	70.0	74.8	74.8	73.0
9	82.50	72.95	69.88	68.18	65.0	65.5	64.5	64.5
10	90.00	75.00	75.51	71.59	65.0	70.5	70.5	68.5
11	97.50	93.75	92.21	86.93	71.0	70.0	71.0	68.0
12	121.88	102.27	97.16	95.45	64.0	64.5	77.5	63.0
13	108.75	86.93	93.75	80.11	81.0	74.0	75.0	74.3
14	99.36	69.88	69.88	71.76	77.0	69.3	71.0	73.0
Mean	96.07[a]	80.82[b]	79.39[b,c]	76.23[c]	70.35	69.61	70.67	69.24
SD	10.792	9.726	9.625	8.220	5.895	4.231	4.885	4.868
	$p<0.0001$	$p<0.0001$	$p<0.0001$	$p<0.0001$	N.S.	N.S.	N.S.	N.S.

N.S. - means are not significantly different
a, b, c - means are significantly different

Average duration for the ascent (concentric) and descent (eccentric) phase of the movement in each bench press condition is provided in Table 4-3. The movement pace was consistent as duration of each phase did not differ significantly between trials and within repetitions.

Table 4-3 Average duration (sec) of eccentric and concentric phases.

Eccentric phase				Concentric phase			
Flat	37°	43°	50°	Flat	37°	43°	50°
1.178 (0.07)	1.193 (0.065)	1.195 (0.106)	1.228 (0.091)	1.144 (0.087)	1.126 (0.106)	1.179 (0.125)	1.165 (0.107)

None of these values were significantly different.

Average normalized EMG activity

Average eccentric and concentric normalized muscle activity during the supine bench press, and the three angles of the incline bench press are presented in Table 4-4 and in Figures 4-1 to 4-8. Each bar graph is a ratio of average muscle activity for the fourteen subjects in each bench press condition. Standard error bars are not present at treatment one in all graphs because the average muscle activity for each bench press condition is expressed as ratio of the average muscle activity in the supine bench press. The average eccentric and concentric ratios expressed for the supine bench press were 1.0, hence no standard error bars. (Note: see Appendix E for a sample Anova table).

Table 4-4 Mean normalized EMG

Mean Normalized EMG (Expressed as a ratio of EMG activity from the supine bench press)								
Angle	Eccentric phase				Concentric phase			
	Clavicular Pectoralis	Sternal Pectoralis	Anterior Deltoid	Middle Deltoid	Clavicular Pectoralis	Sternal Pectoralis	Anterior Deltoid	Middle Deltoid
Supine	1.00	1.00[a]	1.00[a]	1.00[a]	1.00	1.00[a]	1.00[a]	1.00[a]
37°	1.00	0.62[b]	1.54[b]	1.73[b]	1.14	0.84[a,b]	1.43[b]	1.61[b]
43°	1.03	0.64[b]	1.72[b]	1.82[b]	1.17	0.73[b]	1.47[b]	1.68[b]
50°	0.98	0.46[b]	1.62[b]	1.70[b]	1.18	0.68[b]	1.56[b]	1.77[b]
	N.S.	$p < 0.002$	$p < 0.014$	$p < 0.005$	N.S.	$p < 0.02$	$p < 0.002$	$p < 0.005$

N.S. - means are not significantly different
a, b - means are significantly different

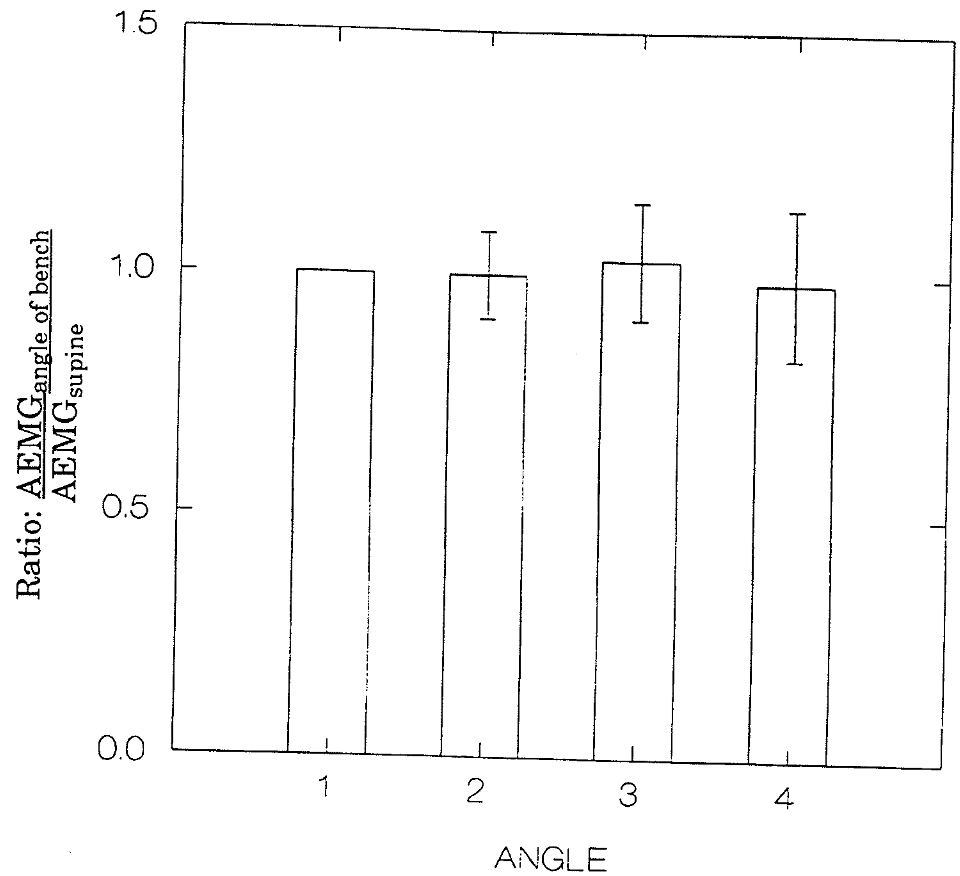

Figure 4-1 Normalized eccentric EMG for the clavicular pectoralis major muscle during 4 bench press conditions. (1) Supine bench press, (2) 37° incline bench press, (3) 43° incline bench press, (4) 50° incline bench press.

Eccentric phase

Average normalized EMG activity in the clavicular portion of the pectoralis major muscle during the eccentric phase of the movement did not change significantly between the supine bench press and the 37°, 43°, and 50° incline bench presses (Figure 4-1). In fact, the average normalized EMG activity during the eccentric phase was almost identical throughout the four bench press conditions. Conversely, average normalized EMG activity in the sternal portion of the pectoralis major (Figure 4-2) muscle was significantly less in the three incline bench conditions than the supine bench press ($p < 0.002$).

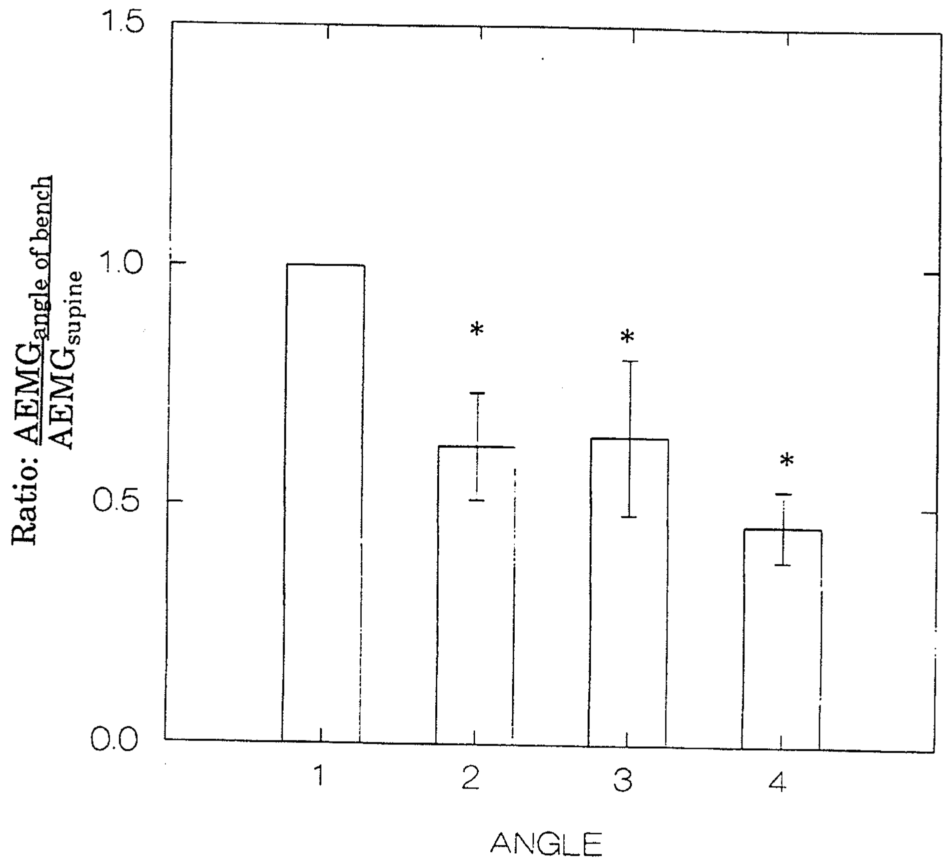

Figure 4-2 Normalized eccentric EMG for the sternal pectoralis major muscle during 4 bench press conditions. (1) Supine bench press, (2) 37^{0} incline bench press, (3) 43^{0} incline bench press, (4) 50^{0} incline bench press. *Signifantly different from supine value ($p < 0.002$).

No differences in terms of average normalized EMG activity was noted between the three incline bench presses. The trend in average normalized EMG activity for both the anterior deltoid (Figure 4-3) and middle deltoid (Figure 4-4) was opposite to that of the sternal pectoralis major, where the average normalized EMG activity for both the anterior deltoid and middle deltoid muscles was significantly greater at the three angles of the incline bench press than the supine bench press ($p < 0.01$ and $p < 0.005$, for anterior deltoid and middle deltoid, respectively).

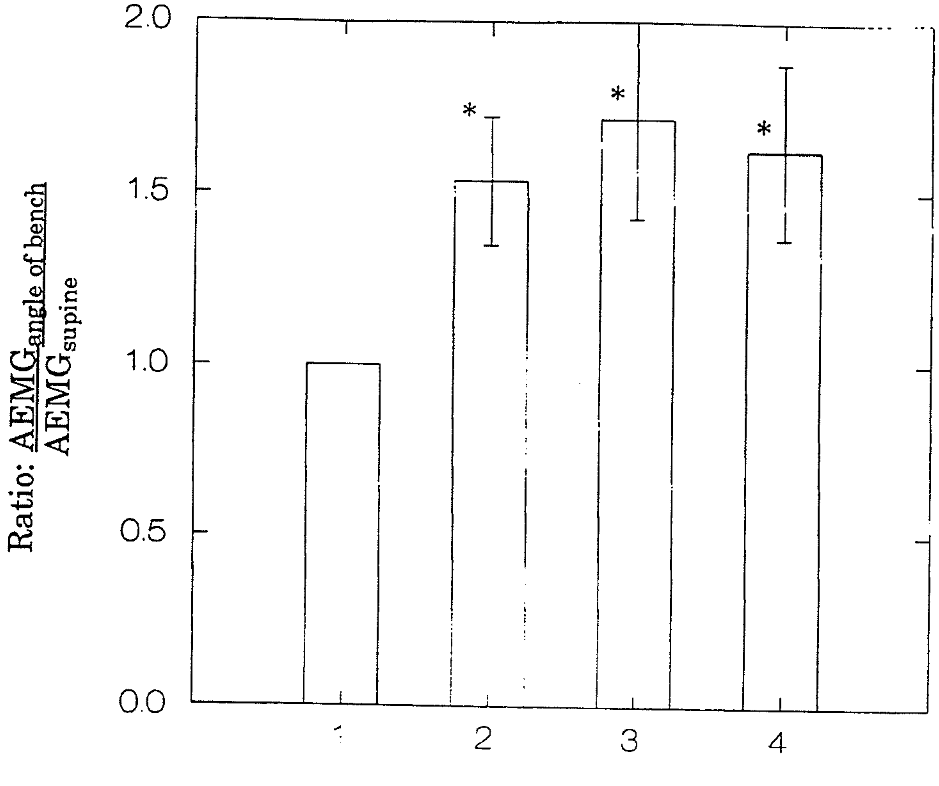

Figure 4-3 Normalized eccentric EMG for the anterior deltoid muscle during 4 bench press conditions. (1) Supine bench press, (2) 37° incline bench press, (3) 43° incline bench press, (4) 50° incline bench press. *Significantly different from the supine value ($p < 0.014$).

The pattern of activation for the middle deltoid muscle was the same as the pattern of activation for the anterior deltoid. Averaged normalized EMG activity for both the anterior deltoid and middle deltoid muscles was not different between the three angles of incline (i.e., one angle of inclination did not appear to increase or decrease the level of activation more than the other).

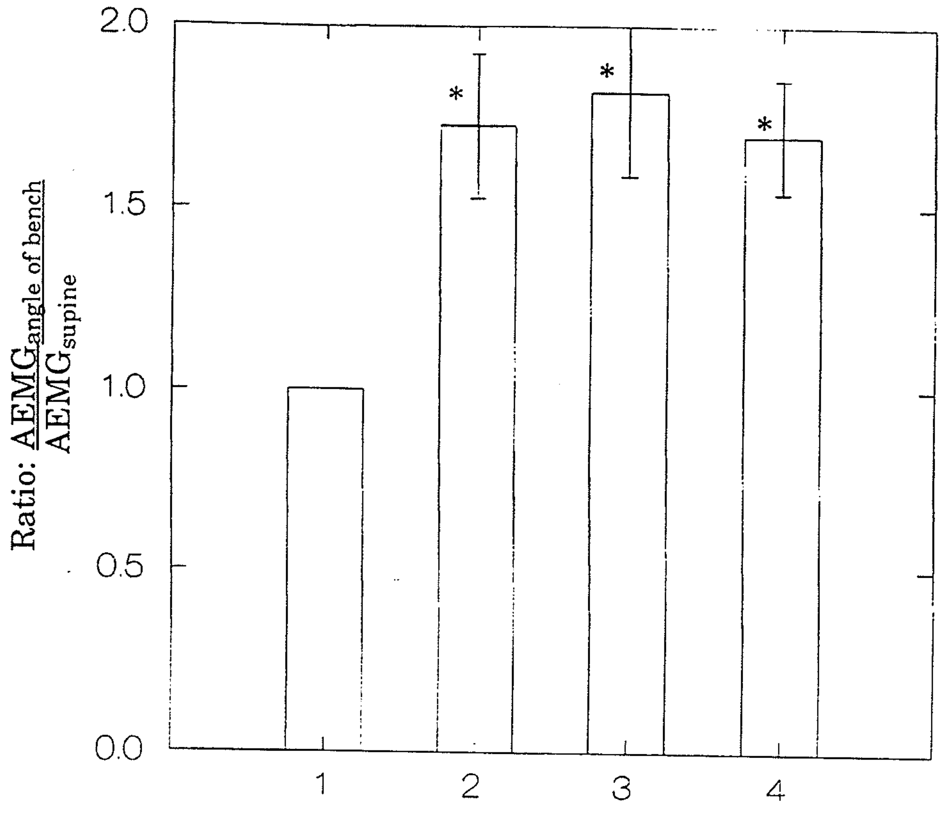

Figure 4-4 Normalized Eccentric EMG for the middle deltoid muscle during 4 bench press conditions. (1) Supine bench press, (2) 37^{o} incline bench press, (3) 43^{o} incline bench press, (4) 50^{o} incline bench press. *Significantly different from the supine value ($p < 0.005$).

Concentric phase

Similar trends in all muscles were noted in the concentric phase of the movement (Figures 4-5 to Figures 4-8). Average normalized EMG activity in the clavicular portion of the pectoralis major muscle during the concentric phase of the movement did not change significantly between the supine bench press and the 37^{o}, 43^{o}, and 50^{o} incline bench presses (Figure 4-5).

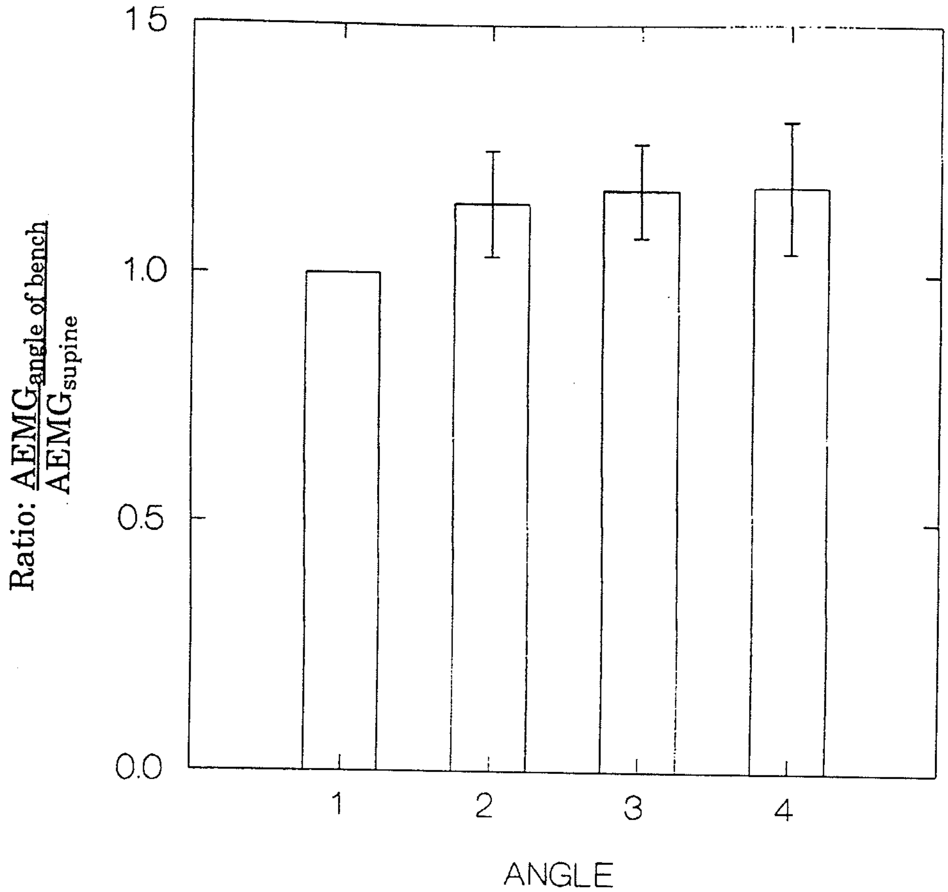

Figure 4-5 Normalized concentric EMG for the clavicular pectoralis major muscle during 4 bench press conditions. (1) Supine bench press, (2) 37^{0} incline bench press, (3) 43^{0} incline bench press, (4) 50^{0} incline bench press.

Similar to the eccentric phase (Figure 4-1), the average normalized EMG activity during the concentric phase was almost identical throughout the four bench press conditions. Average normalized EMG activity during the concentric phase for the sternal portion of the pectoralis major (Figure 4-6) muscle was significantly less in two of the three incline bench conditions than the supine bench press ($p < 0.002$). The average normalized EMG activity during the 37^{0} incline bench press was not significantly different from the supine bench press. The three incline bench presses did not appear to be any different from one another in terms of sternal pectoralis muscle activity.

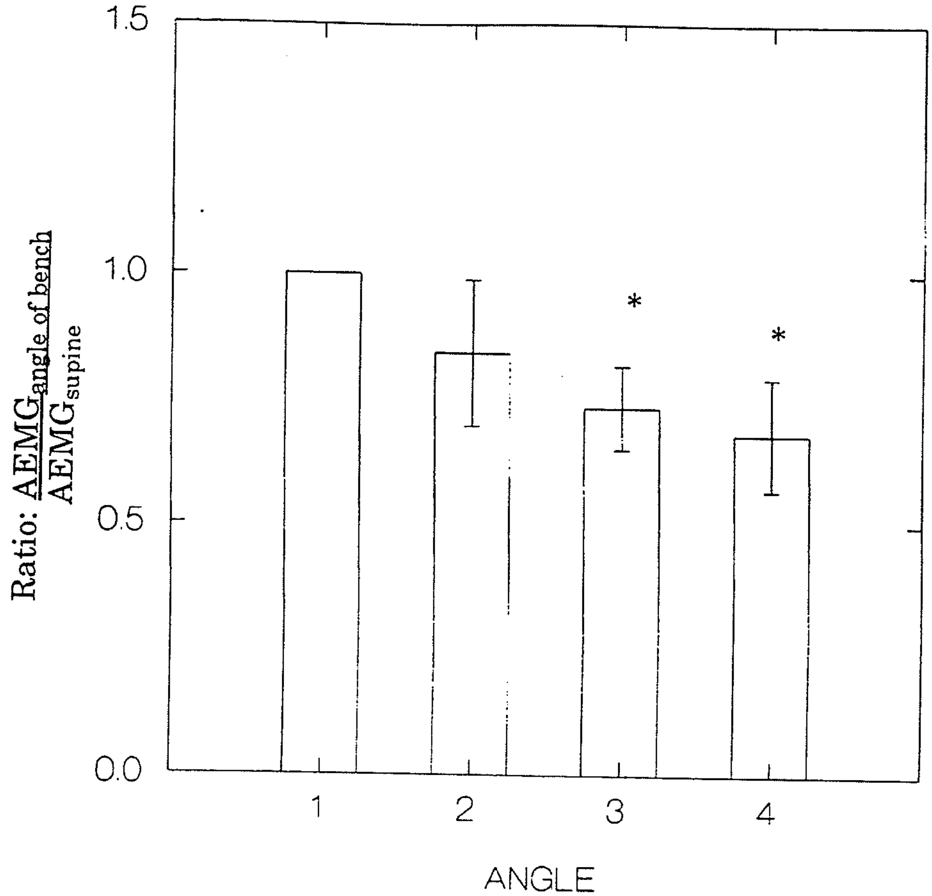

Figure 4-6 Normalized concentric EMG for the sternal pectoralis major muscle during 4 bench press conditions. (1) Supine bench press, (2) 37° incline bench press, (3) 43° incline bench press, (4) 50° incline bench press. *Significantly different from the supine value ($p < 0.02$)

Average normalized EMG activity for both the anterior deltoid (Figure 4-7) and middle deltoid (Figure 4-8) was opposite to that of the sternal pectoralis major, where the average normalized EMG activity for both the anterior deltoid and middle deltoid muscles during the concentric phase was significantly greater at the three angles of incline bench press than the supine bench press ($p < 0.002$ and $p < 0.005$, for anterior deltoid and medial deltoid, respectively).

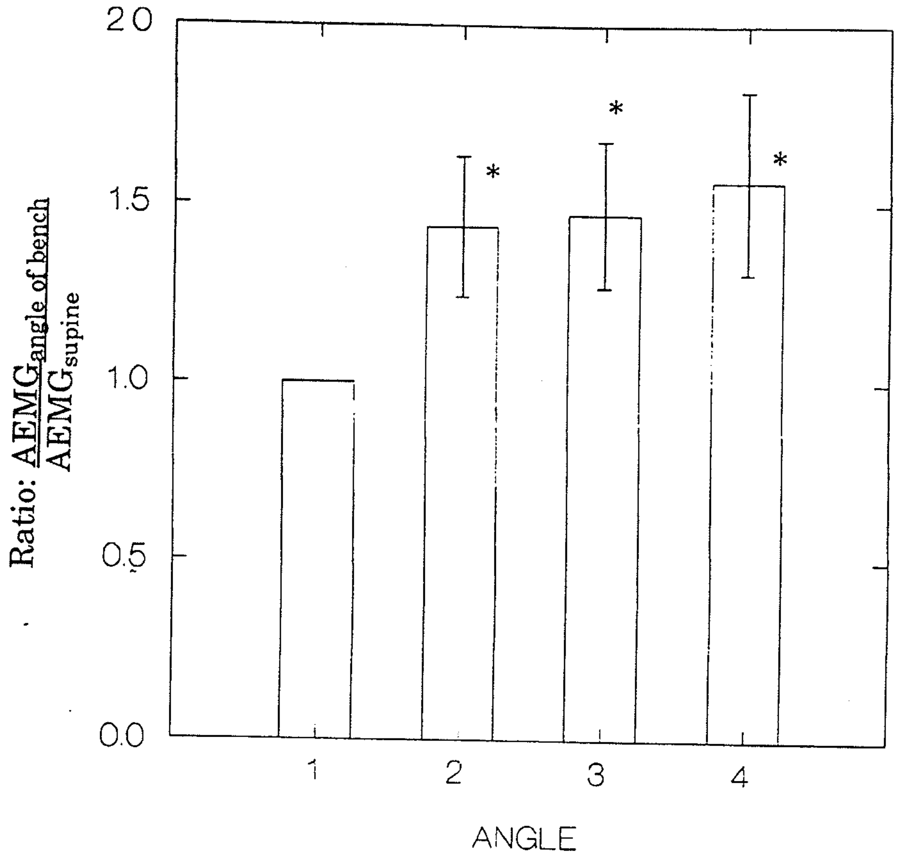

Figure 4-7 Normalized concentric EMG for the anterior deltoid muscle during 4 bench press conditions. (1) Supine bench press, (2) 37^{0} incline bench press, (3) 43^{0} incline bench press, (4) 50^{0} incline bench press. *Significantly different from the supine value ($p < 0.002$).

The averaged normalized EMG activity for both the anterior deltoid and middle deltoid muscles was not different between the three angles of the incline and therefore one angle of inclination was not any different in terms increasing or decreasing the level of activation for the anterior deltoid and middle deltoid.

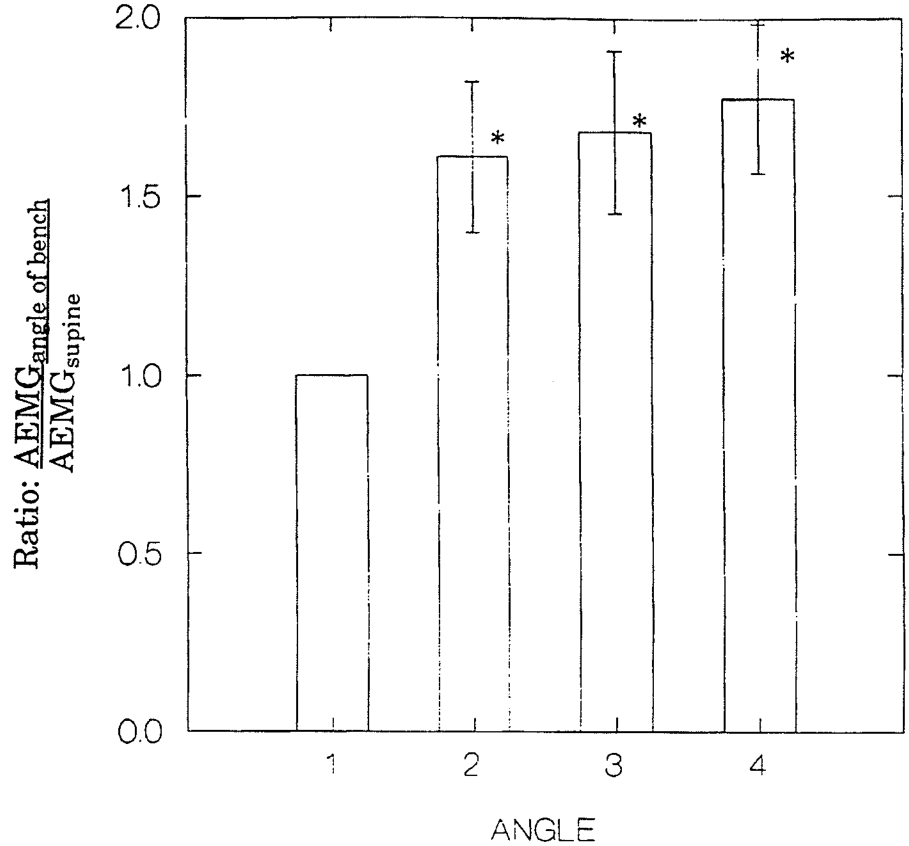

Figure 4-8 Normalized concentric EMG for the middle deltoid muscle during 4 bench press conditions. (1) Supine bench press, (2) 37° incline bench press, (3) 43° incline bench press, (4) 50° incline bench press. *Significantly different from the supine value ($p < 0.005$).

Linear kinematics of bar

Average vertical and horizontal bar displacement, peak vertical velocity, and peak vertical acceleration are presented in Tables 4-5 to 4-10. Horizontal and vertical displacement of four representative trials are also presented graphically. Note that the graphs do not represent the actual (real life) bar path as the increments used in the linear scale to depict the horizontal and vertical displacements affect the bar trajectory (i.e., producing a more variable bar path). In other words, the aspect ratio depicted in the graphs is not the same as in real life.

Bar displacement

Average bar displacement in the horizontal and vertical directions are reported in Table 4-5. The average horizontal displacement during a full repetition decreased as the angle of bench inclination increased, but the decrease was non-significant. Mean horizontal displacement (Table 4-7) was greatest in the supine bench press and least in the 50° incline condition . The 37° incline and 43° incline had mean horizontal bar displacements that were similar (Figures 4-9, 4-10, 4-11, 4-12).

Table 4-5 Average displacement of bar in the horizontal and vertical direction during the bench press (n=14).

Bench press condition	Mean horizontal displacement (m)	SD	Mean vertical displacement (m)	SD
Supine	0.139	0.051	0.500[a]	0.099
37° incline	0.129	0.028	0.562[a,b]	0.184
43° incline	0.126	0.054	0.618[b]	0.114
50° incline	0.119	0.037	0.618[b]	0.109
	N.S.		$p < 0.0005$	

N.S. - means are not significant
a, b - means are significantly different

As the angle of bench inclination increased, mean vertical displacement of the bar increased ($p < 0.0005$), yet the times to achieve these differences were not different between conditions (Table 4-3). Average bar displacement in the vertical direction (Table 4-5) was least in the supine bench press and greatest in the 43° and 50° incline, while the mean vertical bar displacement for the 37° incline was not siginificantly different from the supine bench press or the 43° and 50° inclines (Figures 4-9, 4-10, 4-11, 4-12).

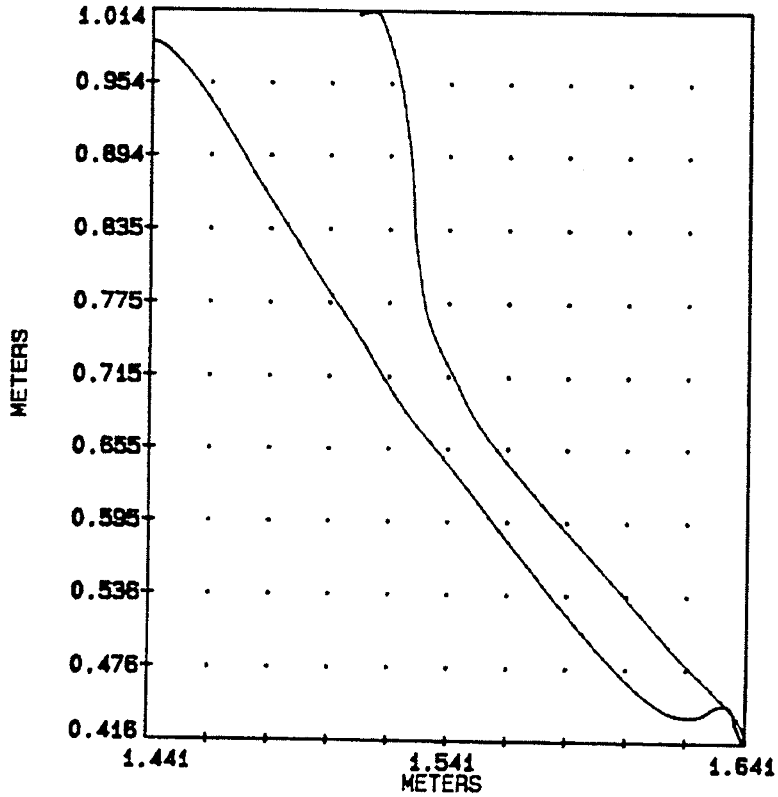

Figure 4-9 Horizontal and vertical bar displacement during the supine bench press for subject 9.

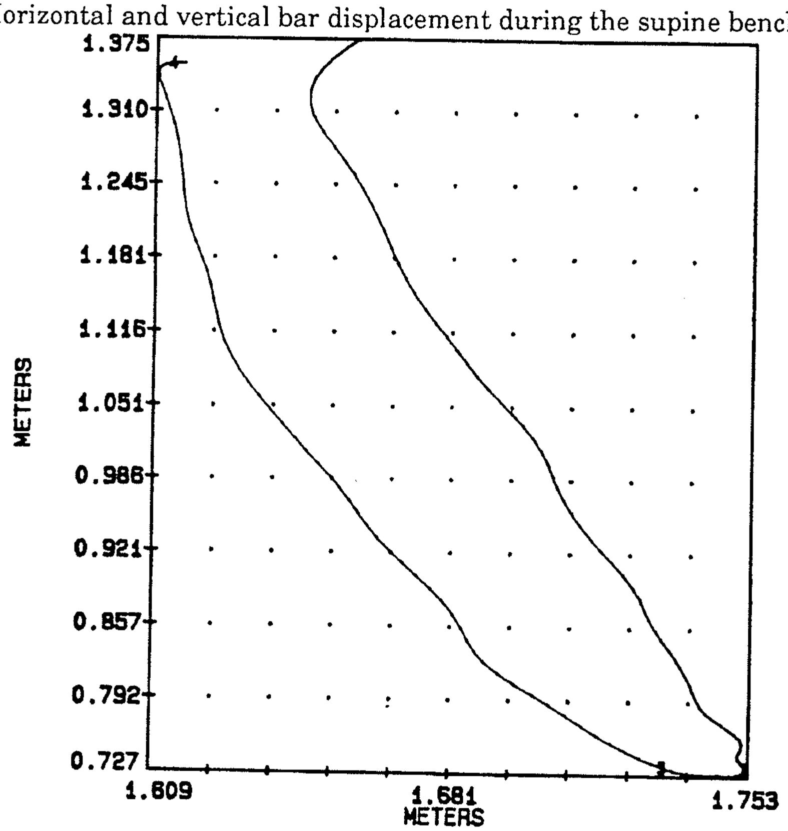

Figure 4-10 Horizontal and vertical bar displacement during the 37° incline bench press for subject 9.

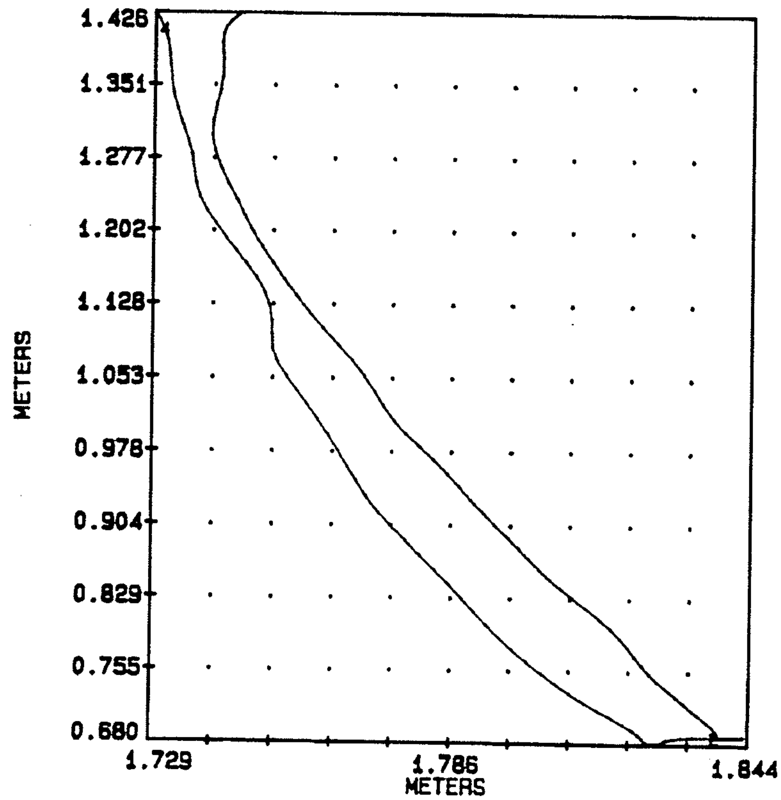

Figure 4-11 Horizontal and vertical bar displacement during the 43^{0} incline bench press for subject 9.

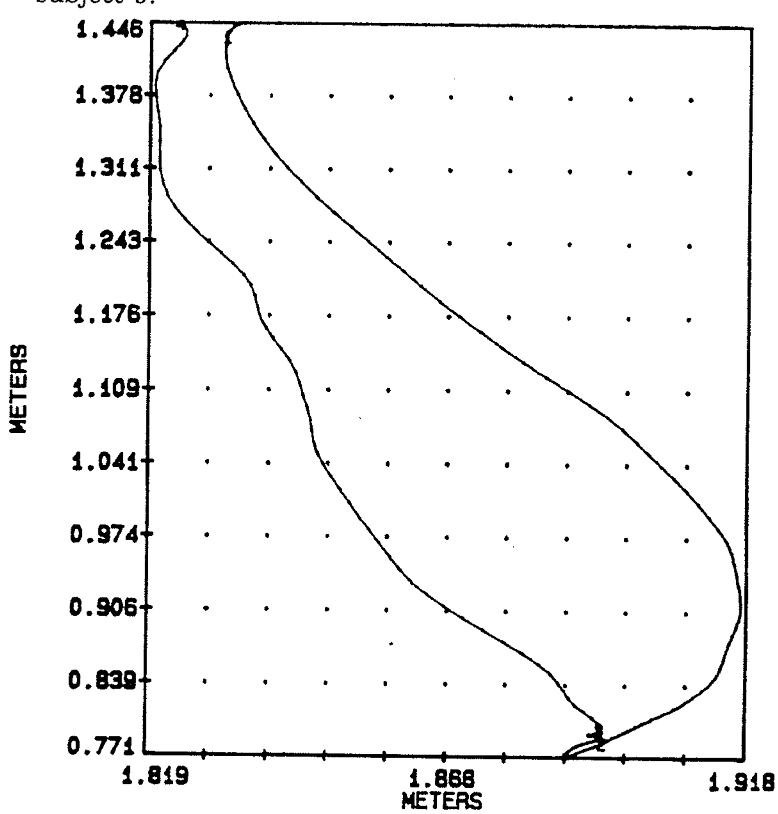

Figure 4-12 Horizontal and vertical bar displacement during the 50^{0} incline bench press for subject 9.

Bar vertical velocity

Mean peak bar velocity in the vertical direction (in both concentric and eccentric phases) is reported in Table 4-6 and Table 4-7. Mean peak downward bar velocity (Table 4-6) increased from -0.656 m/s in the supine bench press to -1.124 m/s in the 50° incline bench press. The average peak downward bar velocity for the 37° incline and 43° incline was -0.754 and -0.782 m/s, respectively. The differences in the mean peak downward velocity in the vertical direction between the four positions, during the eccentric phase were significantly different ($p < 0.0006$). The mean peak downward vertical velocity during the supine bench press was significantly less than those recorded during the three incline bench presses. No differences between the three inclines were noted.

Table 4-6 Mean peak downward velocity (m/s) in the vertical direction during the eccentric phase of the bench press exercise (n=14).

Bench press condition	Mean peak velocity (m/s)	SD
Supine	-0.656[a]	0.180
37° incline	-0.754[b]	0.209
43° incline	-0.782[b]	0.196
50° incline	-0.822[b]	1.043
	$p < 0.0006$	

a, b - means are significantly different

Mean peak upward velocity in the vertical direction (Table 4-7) was least in the supine bench press and greatest during the 50° incline bench press . Mean peak vertical velocity for the 37°, 43°, and 50° incline bench presses did not change significantly. Generally, the average peak upward vertical velocity during the supine bench press was significantly less than all three of the incline bench press conditions ($p < 0.001$). The three incline bench presses were not significantly different from one another.

Table 4-7 Mean peak upward velocity (m/s) in the vertical direction during the concentric phase of the bench press exercise (n=14).

Bench press condition	Mean peak velocity (m/s)	SD
Supine	0.699[a]	0.105
37° incline	0.824[b]	0.120
43° incline	0.800[b]	0.166
50° incline	0.835[b]	0.143
	$p < 0.001$	

a, b - means are significantly different

Vertical acceleration of bar

Mean peak downward vertical acceleration (Table 4-8) for the supine, 37°, 43°, and 50° incline bench presses was generally greater in the three incline conditions. Mean peak downward vertical acceleration was significantly greater ($p < 0.02$) in the 43° incline bench press than the supine bench press. The 37° and 50° incline did not significantly differ from the supine bench press in mean peak downward acceleration.

Table 4-8 Mean peak downward vertical acceleration ($m \cdot s^{-2}$) during eccentric phase of the bench press exercise (n=14).

Bench press condition	Mean peak acceleration ($m \cdot s^{-2}$)	SD
Supine	-3.809[a]	1.325
37° incline	-4.054[a,b]	2.502
43° incline	-4.964[b]	1.802
50° incline	-4.807[a,b]	1.696
	$p < 0.02$	

a, b - means are significantly different

Mean peak upward acceleration (Table 4-9) during the eccentric phase of the exercise (i.e., while the bar was being lowered) did not significantly change throughout the four bench press conditions. There was little variation in the mean values for this variable.

Table 4-9 Mean peak upward vertical acceleration ($m \cdot s^{-2}$) during eccentric phase of the bench press exercise (n=14).

Bench press condition	Mean peak acceleration ($m \cdot s^{-2}$)	SD
Supine	2.643	0.904
37^0 incline	2.843	1.411
43^0 incline	2.529	0.818
50^0 incline	2.586	0.867
	N.S.	

N.S. - means are not significant

Similarly, the mean peak upward vertical acceleration during the concentric phase (Table 4-10) did not change significantly and appeared to be consistent across all conditions.

Table 4-10 Mean peak upward acceleration ($m\ s^{-2}$) in the vertical direction during the concentric phase of the bench press exercise (n=14).

Bench press condition	Mean peak acceleration (m/s/s)	SD
Supine	4.311	1.520
37^0 incline	4.993	1.471
43^0 incline	4.300	1.030
50^0 incline	4.021	0.976
	N.S.	

N.S. - means are not significant

CHAPTER 5
DISCUSSION

EMG findings

The purpose of the study was to compare the effect of four imposed bench press conditions on the relative activation levels of four shoulder muscles; and to determine the validity of present weight training dogma, which suggests that the incline bench press can specifically increase strength and hypertrophy of the clavicular portion of the pectoralis major by increasing the activation level of these fibers (Baechle, Earle, Allerheiligen, 1994; Bijarnson, 1986; Moran & McGlynn, 1990; O'Shea, 1976; Pearl & Moran, 1986; Reynolds, 1991).

The results of the present study do not concur with the above statement. In fact, the activation of the clavicular pectoralis major did not change appreciably in either of the eccentric or concentric phases of the bench press (Figure 4-1 and Figure 4-5). However, the four bench press conditions had an effect on the activation level of the sternal pectoralis major, the middle deltoid and the anterior deltoid. It appears that the three different incline bench press exercises had a similar effect on the patterns of activity of the two deltoid muscles (i.e., increased) and on the sternal portion of the pectoralis major (i.e., decreased).

The trends in activation level were similar between the concentric and eccentric phases with the exception of the sternal pectoralis during the concentric phases where only two of the three angles of inclination produced significantly lower activation levels (Figure 4-6). The activation level of the sternal pectoralis major during the 37^{o} incline bench press was not different from the flat bench press condition. Upon examination of the sternal pectoralis data during the concentric phase, it was noted that the range of values was greatest for the 37^{o} incline bench press (0.272 - 2.549). The large range in values suggest that there was substantial variation from the mean value (which is used to detect significance).

A trade off occurs between the anterior and middle deltoids and the sternal pectoralis whereby the reduction in muscle activity seen in sternal pectoralis during incline bench press was matched by an increased level of activation for the anterior deltoid.

The greatest change in the level of activation was found in both the anterior deltoid and middle deltoid muscles (Figures 4-3, 4-4, 4-7, 4-8) where the average normalized EMG was significantly greater during the incline bench presses than during the supine bench press. The level of activation appeared to increase, although non-significantly, with increasing angles of inclination during the concentric phase of bench press for both muscles. The non significant changes with increasing angle of inclination was due the fact that the loads used during the three incline conditions were not that different from one another. The change in the level of activation in the anterior and middle deltoids during the eccentric phase did not follow the same pattern., although the three angles of inclination had greater changes in the level of activation during the incline than the supine.

The lack of an increase in activation level in the middle and anterior deltoid with a greater angle of inclination may be attributed to the large range of normalized values for those muscles (0.969 - 4.304 and 0.748 - 5.007). In addition the greater variation in the activation level encountered during the eccentric phase may have been caused in part by the nature of downward movement which had been constrained by a metronome (i.e., forced pace and froced load). The variability in changes in normalized EMG would also tend to suggest that some of the subjects employed different strategies for lowering the bar, or that some of them were not well trained during the eccentric phase of the movement and had greater difficulty with that portion of the exercise. (Note: according to this author, the lack of eccentric training appears to be a common occurrence among many weight lifters who neglect to concentrate on that portion of exercise.)

Some authors (Bijarnson, 1986; O'Shea, 1976) have suggested that at greater angles of inclination the level of activation of the deltoid muscle will increase, yet the present study did not demonstrate that trend. Although the mean normalized EMG increased significantly from the supine to the incline bench press, the mean normalized EMG between the three incline bench presses did not change significantly. The fact that the mean normalized EMG did not change between the three incline bench presses was due to the fact that the changes in the submaximal loads used during the three angles did not change significantly (Table 4-2).

It appears that the anterior deltoid and middle deltoid muscles play an important role during the incline bench press. McCaw and Friday (1994) suggested that the anterior deltoid tends to resist lateral rotation of the humerus, and the middle deltoid tends to resist adduction of the humerus. The increased activity observed in both these muscles may reflect their combined role in stabilizing the shoulder and manipulating the load (i.e., pressing the weight). It has been reported that all three heads of the deltoid are active in all movements of the humerus, with one head acting as a prime mover and the other heads acting to stabilize the glenohumeral joint (Scheving & Pauly, 1959). In the context of the present study, it would appear that the anterior deltoid was active as both a prime mover and a stabilizer (i.e., resisting external rotation and limit posterior translation of the humeral head on the fossa) during the incline bench press while the middle deltoid was active in stabilizing the humerus against the glenoid fossa. Since there is a tendency for the humerus to adduct and be externally rotated during the incline bench press, it is also reasonable to assume that both heads of the deltoid muscle acted to limit external rotation and humeral adduction as well as limit posterior humeral translation.

The mean normalized EMG for the clavicular pectoralis did not change significantly, (i.e., 1.0, 1.0, 1.03, 0.98, for the supine, 37^o, 43^o, and 50^o incline bench press, respectively), throughout the four bench press conditions in either phase. Rather than being the prime mover during the incline bench press (as it has been assumed), it appears that the clavicular pectoralis major plays a role in assisting the movement (i.e., synergist) during the incline bench press conditions while acting as a stabilizer during the supine bench press. Since the clavicular pectoralis has similar actions to the anterior deltoid (i.e., shoulder flexion, horizontal adduction, and medial rotation), it is not unreasonable to assume that the clavicular pectoralis functioned in the same manner as the anterior deltoid. That is, acting to manipulate the load (i.e., synergist) and provide stabilizing forces during the three incline bench presses.

Interestingly, the separate portions of the pectoralis major muscle did not exhibit the same activation pattern as the incline angle increased. Rather a separation in the neural activation pattern for the two heads of the pectoralis major was noticed as the angle of inclination changed. In other words, the sternal portion of the pectoralis major muscle did not behave in the same manner as the clavicular portion of the same muscle. The mean normalized EMG for the sternal pectoralis muscle decreased significantly from the flat to the incline bench press while the level of activation in the clavicular portion did not change throughout the four conditions. Since anatomical studies by Chaffai and Mansat (1988), Milroy and Korula (1988), and Wolfe et al., (1992), have shown that the pectoralis major is innervated by three separate nerves and the results of the present study show differential levels of activation between the clavicular and sternal portion of the muscle, it appears that these two portions of the muscle can exhibit different neural activation patterns lending credibility to the notion that the two heads can perform separate functions.

The organization of the pectoralis major is complex and since it is innervated by three separate nerves, the muscle may exist as three separate neuromuscular compartments. This may provide the muscle with the unique characteristic of possessing independent function between the three portions. The existence of distinct neuromuscular compartments in which motor units occupy specific territories within a muscle has been studied extensively in cats and rats, and now in humans (Bodine-Fowler et al., 1990; English et al., 1993; Hammond et al., 1989; Thomson et al., 1991; Tonndorf & Hannam, 1994). The significance of these neuromuscular partitions has yet to be clearly stated. It appears that the role of these compartments may depend upon the task at hand, and depending on the demands of the task, different partitions of a muscle may be recruited (English et al., 1993), which would be consistent with the different lines of muscle action of the two heads.

In the context of the present study, this differential activity suggests that the clavicular portion of the muscle acted independently from the sternal portion of the muscle, as the level of activation did not change for one muscle while it decreased for the other. As a result of demands of the movement, the clavicular portion of the muscle had to assist (i.e., as a stabilizer and synergist) the deltoid muscle with the manipulation of the loads during the incline bench press condition while the sternal portion became less involved with the movement.

It appears that the significant decrease in the mean normalized EMG that was noted for the sternal pectoralis, from the supine bench press to the three angles of the supine bench press is related to the mechanical differences in both types of movement. From a biomechanical view point, the supine and incline bench press posess different movement profiles.

The first point to consider is the moment arm of the load about the shoulder in the sagittal plane. As the angle of bench press inclination increased, the horizontal displacement of the bar (Table 4-5) towards the shoulder decreased (although non-significantly). In addition, the vertical displacement significantly increased from the supine to the 43^{o} and 50^{o} incline angles (Table 4-5), and as a result, the path of bar appeared to be relatively closer to the shoulder. Therefore, it is reasonable to assume that the moment arm of the load about the shoulder in the sagittal plane would decrease with increasing angle of inclination because the path of the bar appeared to be relatively closer to the shoulder. The moment created by the load would then tend to be less for the incline bench presses than the supine, and therefore the sternal pectoralis decreased its' level of activity because of this reduced moment. However, the extent to which the moment of the load changes with bench press angle is not known. In addition, whether the moment of the weight about the shoulder in the sagittal plane accounted for the changes in the mean normalized EMG for the sternal pectoralis, the anterior deltoid, and middle deltoid muscle is not known.

Secondly, the subtle differences in muscle actions required by each movement may alone have contributed to differences noted in the sternal pectoralis, the middle deltoid, and anterior deltoid. When examining both types of bench press movements, the supine bench press requires horizontal adduction of humerus across the middle of the pectoralis major muscle while the incline bench press horizontally adducts the humerus across the upper third of the pectoralis major (clavicular portion). In addition, the humerus is externally rotated during the incline bench press (in fact, external rotation appears to increase as the angle of inclination increased). Also, adduction of the humerus during the supine bench press occurs mainly in the transverse plane, while the incline bench press relies on a combination of humeral adduction in the transverse plane and humeral abduction in the frontal plane.

The humeral adduction and abduction between the transverse and frontal planes, along with external rotation of the humerus may alter the line of pull for the sternal pectoralis major so that it is not a prime mover during this exercise. The sternal pectoralis may also have a reduced role during this movement, because it may actually interfere with the mechanics of the exercise. In other words, the action it provides during the incline bench press may oppose the movement rather than assist in it. It has been noted by Basmajian (1982), that the pectoralis major can assist in adduction in the frontal plane when the upper arm is in 90° of abduction or more. During the incline bench press, the arm appears to be adducted at approximately 90° in the frontal plane. So if the sternal pectoralis was to act as the prime mover during the incline bench press it would adduct the humerus in the frontal plane, rather than in the transverse plane.

Finally, the position of the humerus during the incline bench press may also place the fibers of the sternal pectoralis on a greater stretch than in the supine bench press. Although muscle fiber length of the pectoralis major has not been well documented, Wolfe et al., (1992) found that at 30° of horizontal abduction (transverse plane) and 80° of adduction (frontal plane), the sternal fibers of the muscle could be stretched 40% beyond their resting length, while the remaining fibers were stretched 19% beyond their resting length, thereby placing the sternal fibers at a disadvantage for force production.

The amount of isometric force a given muscle is capable of producing is dependent on the length of that muscle, and the length changes as joint movement occurs. Muscle force is generated when the actin and myosin myofilaments form cross-bridges. A muscle can generate the most force while it is at resting length (anatomical position) because the greatest number of cross-bridges between myosin and actin exists. When the length of the muscle increases (i.e., lengthens) or decreases (i.e., shortens) beyond the resting length, the number of available binding sites is reduced, and the maximum force a muscle can produce decreases. The incline bench press may place the humerus in such a position that the sternal portion of the muscle is at such an increased length that the available actin and myosin binding sites are greatly reduced and therefore the sternal pectoralis is at a disadvantage in force production, hence the reduction in muscle activity.

Temporal aspects of movement

The metronome proved to be an effective technique to standardize the movement pace. The mean ascent and descent times across (and within) the four bench press conditions were consistent (Table 4-3). In fact, the standard deviation was small and ranged from a low of 0.065 milliseconds to high of 0.125 milliseconds. The uniformity of the movement ensured that the EMG collection period was consistent across the four conditions as well as between the eccentric and concentric phases of the movement.

The average times for the descent and ascent phases for all four bench press conditions were similar to the times reported by McCaw and Friday

(1994), yet different from the studies (Table 2-1) of McLaughlin and Madsen (1984), Elliott et al., (1989), and Madsen and McLaughlin (1984). The descent and ascent phases during supine bench press in the McCaw and Friday study consisted of 49.3% and 50.7% of the total lift time while the average descent and ascent phases for all four conditions in the present study was 50.95% and 49.05%, respectively (the supine bench press had descent and ascent values that comprised of 50.73% and 49.67% of total lift time). The differences encountered with the other reported times (Madsen & McLaughlin, 1984) were a result of loading (i.e., most loads represented a 100% RM) and the fact that movement pace was not set.

Finally, controlling the movement pace by using a metronome may have altered the technique of some of the subjects in the study as compared to the technique used in real life training. Many weight lifters make the mistake of bouncing the bar off their chests to assist in the manipulation of the load. This is a dangerous practice (i.e., potential for fracture of sternum) and also greatly reduces the tension placed on the muscle during the initial portion of the ascent phase which in turn greatly reduces the training effect of the exercise.

Bar kinematics

Bar displacement

Horizontal and vertical bar displacement (Table 4-5) appeared to be affected by the four different bench press conditions. Horizontal displacement decreased (although non-significantly) as the angle of inclination increased suggesting that the path of the bar became more vertical with increasing angles. The greatest horizontal bar displacement which occurred during the supine bench press (0.139 m) was similar to the values of 0.135- 0.155 m, 0.140 m, and 0.125m, recorded by Elliott et al., (1989), McLaughlin and

Madsen (1984), and Wagner et al., (1992), respectively. These values, however, were calculated during the performance of a 1 RM bench press exercise, and thus the horizontal displacement during a submaximal lift, (such as the one performed in the present study), may have been different. In fact, Elliott et al., (1989), recorded a horizontal displacement of 0.09 meters for an 81% RM load.

The greater displacement noted during the performance of the supine bench press using a 75% RM load may have been attributed to the difference in subjects used in both studies. The subjects used in the present study were experienced lifters who had a minimum of two years of bench pressing and had recorded maximum lifts ranging between 110-162.5 kilograms while the subjects used in the Elliott et al., study were elite bench pressers who had recorded maximum lifts of 150-245 kilograms. It has been noted by Elliott et al., (1989), McLaughlin and Madsen (1984), Wagner et al., (1992), and Wilson et al., (1989), that increased horizontal displacement towards the shoulder helps reduce the moment of the load about the shoulder in the saggital plane which in turn assists in the successful completion of the movement as the torque generated by the muscles can specifically focus on pressing the load rather than countering a secondary torque. It is possible that the subjects in the present study employed this strategy in order to perform the exercise with consistency.

Since the subjects in the Elliott et al., study were elite bench pressers (internationally competitive), the 81% RM load (which was performed for only one repetition) may not have been sufficiently heavy to require a greater horizontal displacement. In addition, elite bench pressers frequently train with loads in excess of 85-90% of 1 RM and therefore an 81% RM load may have been considered relatively light. Subjects used in the present study were experienced in weight lifting and not considered elite. These subjects usually employed loads ranging between 70-80% of 1 RM and may have found a 75% RM load more challenging than would elite bench pressers with an 81% load. It would appear that the more skilled the individual, the greater the efficacy of neuromuscular control, the greater the load that can be pressed safely. Therefore the need to reduce the moment of the weight about the shoulder in the sagittal plane during pressing of submaximal loads may not be as crucial for elite athletes as opposed to experienced weight trainers.

The significant change in vertical displacement (11.8 cm difference) noted between the supine and the 43^{0} and 50^{0} incline bench press was a result of the change in bench press inclination and not changes in grip width. Subjects were allowed to choose their preferred grip width for each of the four maximal bench press tests. These grips were then used for the submaximal test. Although grip width was not standardized between the bench press conditions, the four grips used by the subject were not significantly different. Therefore, the increased vertical bar displacement was caused the by bar location on the chest at the end of the each descent phase. As the angle of bench changed from supine to incline, the bar was lowered vertically and closer to the level of the shoulder axis in the sagittal plane. In other words, the bar is lowered to the chest (at nipple level) during the supine bench press and as the angle of inclination increased the bar was lowered to the level of the clavicles. This could have accounted for differences in vertical displacement of at least 3-4 centimetres. In addition, some subjects may have increased the vertical displacement by horizontally adducting the humerus after the elbows had been locked out. This additional anterior translation/horizontal adduction may have contributed to an increase of 4-5 centimeters.

Maximum vertical bar displacement during the supine bench press was 0.500 meters which was comparable to 0.485 (preferred grip) and 0.520 (narrow grip) reported by Wagner et al., (1992) which are very similar to those of this study. Vertical bar displacement reported by Wilson et al., (1989), and McLaughlin and Madsen (1984), were 0.380, and 0.380-0.440 meters, respectively. Since Wagner et al., (1992) found a difference in vertical bar displacement of 0.220 meters between different grip widths, the disparity between the values reported in the present study and those of Wilson et al., (1989) and McLaughlin and Madsen (1984) were probably due to the use of different grip widths. Unfortunately, grip widths used in those studies were not reported. In addition body type and arm length may have accounted for the differences reported between the present study and some of those cited in the literature. The subjects in the present study may have had longer arms than those reported by Wagner et al., (1992), Wilson et al., (1989), and McLaughlin and Madsen (1984). The extent to which subject's arm lengths differred from those reported in other studies in unknown.

Bar velocity

Peak vertical bar velocity (Tables 4-6 and 4-7) between subjects was fairly consistent as standard deviations ranged from 0.105 to 0.209, suggesting that the movement pace set by the metronome was a reliable method in controlling the upward and downward phases of the exercise. It appears that the specific bench press condition had an effect on the average peak vertical velocity during the ascent and descent phases. In both phases, the peak vertical velocity was significantly less than the peak velocities encountered during the three incline bench presses. The greater peak velocities encountered during the incline bench presses may have been a strategy employed by the subjects to keep the movement pace consistent. It was noted that the vertical displacement was significantly greater during the incline bench presses yet the average duration of the eccentric and concentric phases did not change throughout the four conditions. Therefore, it is possible that in order to keep the duration of the repetitions consistent (as set by the metronome), it was necessary to increase the vertical velocity of the bar in both the eccentric and concentric phases of the movement.

Peak vertical bar velocity during a maximal supine bench press was reported in Madsen and McLaughlin (1984) to be 0.31 m/s and 0.25 m/s in the descent and ascent phases, respectively. These velocities were much lower than the ones reported in the current study and were due to the fact that a maximal load was being manipulated in Madsen and McLaughlin (1984) study as compared to a submaximal one during the present study. It is easier to manipulate lighter loads faster than heavier ones. This remark forms the underlying principle of the force-velocity relationship which states that when a muscle develops tension (concentrically) against a heavy load, the velocity of shortening must be relatively slow; when the resistance is light, the muscles can contract faster (Hall, 1995).

Bar acceleration

Vertical bar acceleration (Tables 4-8 to 4-10) in both the eccentric and concentric phases was highly variable throughout the four bench press conditions. In fact, statistical significance in peak vertical acceleration differences was only present between the supine bench press and the 43^{0} incline bench press during the eccentric phase of the movement. Nevertheless, the peak downward acceleration during the eccentric phase was significantly greater in the 43^{0} incline bench press than the supine. Since force is directly proportional to acceleration, changes in acceleration reflect alteration in the force applied by the subject to the bar. In the context of the present study, this downward acceleration (Table 4-8) represents the portion of the exercise in which the subject is letting the force of gravity assist in the movement of the bar. The greater the downward acceleration, the less resistance the subject is applying to the bar to decrease this acceleration.

Although all the three incline bench presses had greater downward vertical accelerations than the supine, only the 43^{o} incline was significantly greater than the supine, suggesting that subjects applied more force to the bar (in an attempt to control the downward path of the load) during the supine bench press as compared to the 43^{o} incline bench press. It appears that the high variations in vertical bar acceleration may have affected the significance, yet the fact that all three inclines recorded greater downward acceleration is still worth mentioning. The higher accelerations may have been a method of ensuring that the subjects kept a consistent movement pace as the bar had to be moved through a greater vertical distance during the incline bench press. In addition, the downward accelerations imply that less vertical force was being generated during the eccentric portion of the movement.

The peak downward bar acceleration for the supine bench press (-3.809 m/s^2) recorded in the present study was much greater than the -0.696 m/s^2 reported by Madsen and McLaughlin (1984). In addition, Madsen and McLaughlin (1984) reported a high downward acceleration of 3.303 m/s^2 for a novice group of bench pressers. The greater downward acceleration noted was a function of the load being manipulated as subjects in the present study were asked to perform a bench press using 75% RM load while the Madsen and McLaughlin (1984) recorded acceleration profiles for 100% RM loads. Therefore both groups of subjects in the Madsen and McLaughlin study had to generate more force in a longer time period to safely control the downward movement of the bar.

The existence of notable individual variability in the downward accelerations should be noted. The variation for this parameter was very high and may have affected the significance between all three incline bench presses and the supine bench press. Table 4-8 clearly demonstrates the large variation within all four tests conditions, which would suggest that there exists some individual differences. To determine whether the individual differences reflect variability in movement strategies employed by subjects or training and muscle mechanics specific to an individual is beyond the scope of the present study. However, studies designed to explain the presence of individual differences represent a possible direction for research investigating muscle activity and movement kinematics during weight training exercises.

The second value recorded during the downward movement of the bar was described as the upward acceleration of the bar while the bar was still being lowered (Table 4-9). These accelerations represent the forces applied to the bar to arrest (slow down) the downward motion. In other words, these values represent the "catch" phase of the movement where the subjects had to exert forces that were greater than the actual load in an attempt to slow the bar to a stop. The upward accelerations during the eccentric phase did not change significantly throughout the four conditions, suggesting that in order to decelerate the downward movement of the bar, the strategy used was consistent throughout the four conditions.

Although the upward accelerations did not change, the level of muscle activation significantly changed between the incline press and the supine bench press. This suggests that as the sternal portion of the pectoralis major decreased activity, the anterior and middle deltoid muscles had to increase their activity which in turn kept the force profile consistent.

Madsen and McLaughlin (1984) and McLaughlin and Madsen (1984) reported small upward accelerations of 0.350 m/s^2 (elite heavy weight bench pressers) and 0.59 m/s^2 (elite light weight bench pressers) during the descent phase of supine bench press. These values were small because the load being lifted was a maximal load and thus could not be manipulated as easily as a submaximal load such as the 75% RM load used in the present study.

Peak vertical acceleration (Table 4-10) in the concentric phase of the movement did not change significantly across the four bench press conditions, again suggesting that the movement pace set by the metronome was consistent across all trials. In addition, the non-significant changes in bar acceleration suggest that the vertical force exerted on the bar was consistent across the trials. Although there was no significant change in upward acceleration, the level of muscle activation significantly changed between the three angles of the incline bench press and the supine bench press (i.e., the supine bench was different from the three inclines). This suggests that as the sternal portion of the pectoralis major decreased activity, the anterior and middle deltoid muscles increased their activity which in turn may have assisted in keeping the vertical force exerted against the bar consistent across the four trials. (Although the tension developed by the muscles would not be consistent due to changes in moment arm and changes in the length tension relationship of the muscles used in the movement.)

Limitations

A limitation of the study was the fact that only one camera view was used for the analysis. Although the transverse view was filmed, it was only used to ensure that the bar was evenly pressed during each trial. Analysis of the transverse view may have provided additional kinematic information regarding anterior and posterior translation of the shoulder during the eccentric and concentric phases of the movement. (Note: additional anterior translation/horizontal adduction of the humerus may have increased the vertical displacement of bar). In addition, this view can also provide data regarding changes in upper-forearm angle and can also be used to estimate moments occurring about elbow and shoulder in the transverse plane. Therefore this may be an area that warrants consideration for further studies.

Practical application

The non significant changes observed in the average normalized EMG for the clavicular pectoralis suggests that the incline bench press does not have the effect of specifically increasing muscle activity as previously stated (Baechle, Earle, Allerheiligen, 1994; Reynolds, 1991; Moran & McGlynn, 1990; Bijarnson, 1986; Pearl & Moran, 1986; O'Shea, 1976). Using this type of exercise to specifically increase strength and hypertrophy in the clavicular portion of the pectoralis major may not have that specific effect. The higher normalized EMG values for the anterior deltoid and middle deltoid muscles suggest that these muscles are more active during the incline bench press than the supine bench press. Within the context of the variables studied, the incline bench press does not appear to provide an advantage over the supine bench press in terms of increasing the activation of the clavicular portion of the pectoralis major muscle. It appears that the incline bench press is a better shoulder exercise than it is an "upper chest" exercise.

The results of the study provide some insight into the partitioning hypothesis which suggests that certain muscles exist as separate neuromuscular compartments. The results of the present study have considerable implications for other weight training exercises that make the claim of increasing the level of activation of specific fibers within a muscle. Trainers, coaches, and educators in the weight training community have a responsibility to provide accurate information. Therefore it is imperative that more research should be conducted on various exercises that are believed to offer specific advantages over others. For example, the decline bench press is often believed to increase the activity of the lower third of the pectoralis major fibers. Different abdominal exercises such as the reverse crunch have been associated with improving the strength in lower abdominal muscles. Changing the orientation of the heels during the standing heel raise so that toes are pointing more laterally or medially is often believed to change the level of activation of the medial or lateral head of the gastrocnemius. Hypotheses of this type require an EMG study similar to the present study, to examine muscle activity in different positions of a particular exercise. With the current interest in resistance training by subjects of all age levels and both sexes, information of this type would be useful and appreciated by participants.

CHAPTER 6

SUMMARY, CONCLUSIONS, AND RECOMMENDATIONS

Summary

The purpose of the study was to compare the effect of four imposed bench press conditions (i.e., supine, 37^{o}, 43^{o}, and 50^{o} incline bench presses) on the normalized EMG levels of sternal pectoralis major, clavicular pectoralis major, anterior deltoid, and middle deltoid muscles. This was done to assess the accuracy of weight training dogma which suggests that the incline bench press can specifically increase the activation level of the clavicular fibers of the pectoralis major. It was hypothesized that the relative muscle activity in all four muscles would not change across all four trials tested.

The data was collected on 14 male subjects while performing four different bench press exercises (i.e., supine, 37^{o}, 43^{o}, and 50^{o} incline), using a 75% RM load which had been separately determined for all four conditions prior to the data collection. Using the normalized EMG data, it was noted that the four bench press conditions had an effect on the level of activation of three of the four muscles tested. No significant changes in the level of activation of the clavicular pectoralis major occurred across all four trials, thereby refuting the suggestion that the incline bench press can elicit greater muscle activity from that portion of the muscle. Activation of the sternal pectoralis muscle during the supine bench press was significantly greater than during the 43^{o} and 50^{o} incline, but not during the 37^{o} incline bench press. Both the middle and anterior deltoid muscles had significantly greater levels of activation during the three incline bench press exercises than the supine bench press, suggesting that these muscles play an important stabilizing role in the incline bench press.

Mean ascent and descent times were consistent across all trials as well as within each repetition. Horizontal and vertical displacement of the bar was affected by the bench press condition. Although horizontal displacement decreased (0.139, 0.129, 0.126, 0.119 meters) with increasing angle of inclination (suggesting that the bar path became more vertical), the changes were not significant. Conversely, vertical displacement increased as the angle of inclination increased (the mean vertical displacements were 0.500, 0.562, 0.618, 0.618 meters). The vertical displacement during the 43^o and 50^o incline was greater than the supine bench press, yet no different form each other and the 37^o incline bench press.

Peak vertical bar velocity for the supine, 37^o, 43^o, and 50^o incline bench press conditions during the eccentric phase were 0.656, 0.754, 0.782, and 0.822 m/s, and 0.699, 0.824, 0.800, and 0.835 m/s for the concentric phases. In all cases, the incline bench press trials had greater peak velocities than the supine bench press, suggesting that the subject may have increased the velocity of the bar to keep the total movement pace constant throughout the four conditions. Vertical bar velocity during the supine bench press was greater than the values reported in the literature because the load being manipulated was submaximal. Values given previously were for maximal loads.

Peak vertical bar acceleration did not significantly change across the four trials with the exception of one condition in which peak downward acceleration during the flat bench press (-3.809 m/s^2) was significantly less than the 43^o incline bench press (-4.964 m/s^2). In addition, the peak vertical downward accelerations of the 37^o and 50^o incline bench press were greater (-4.054 and -4.807 m/s^2) than the supine bench press (although non significantly). There appeared to be excessive variation in the peak downward acceleration which may have affected the ability to detect statistical significance in the 37° and 50° incline bench press conditions. The variation in acceleration found within all four conditions may have been a result of digitizing error; or it suggested that there existed variability in the movement strategy used by the subjects. Furthermore, the greater downward acceleration noted during the incline bench press suggested that less vertical force was being applied against the bar during the eccentric phase.

Finally, the peak upward velocity during the eccentric and concentric phases did not change significantly across the four bench press trials. This indicated that the strategy used to decelerate the load during the eccentric phase and to accelerate the load in the concentric phase was relatively consistent throughout the four bench press conditions. The first upward peak in vertical acceleration represented the "catch" phase of the movement, where subjects had to exert a force greater than the actual load in order to slow the weight down.

Although the upward accelerations did not change appreciably, the level of muscle activity in the sternal pectoralis, anterior deltoid, and middle deltoid significantly changed. This suggested that as the level of activation in the sternal pectoralis major decreased from supine to incline bench presses, the anterior deltoid and middle deltoid increased their activity to keep the upward accelerations consistent. This alteration in muscle activation produced a more constant application of vertical force across all four conditions.

It appears that subtle differences in the movement profile between the two types of bench press exercise likely contributed to the changes in the activation levels of the sternal pectoralis major, anterior deltoid, and middle deltoid muscle. During the supine bench press, the humerus is pulled directly across the center of the pectoralis in a movement of horizontal shoulder adduction. In the incline bench press, the humerus is externally rotated and is pulled across the upper third of the pectoralis. This simple change in humeral orientation may have changed the line of pull of sternal pectoralis so that it was not at an optimal force producing angle and therefore the incline bench press required greater activation from the anterior deltoid and middle deltoid muscle to perform the exercise. In addition, the line of pull for the sternal pectoralis major may have been in such a position that it may have interfered with the movement during the incline bench press. If it were to contract it may have pulled the upper arm down toward the side of trunk (i.e., humeral adduction in the frontal plane). Finally, the change in humeral orientation may have also affected the length tension relationship of the sternal pectoralis major muscle so that it was at a disavantage in terms of force production.

Conclusions

Based on the results of the present study, the following conclusions are justified:

1. The incline bench press did not increase the level of activation in the clavicular portion of the pectoralis major over the supine bench press.
2. The incline bench press decreased the level of activation in the sternal pectoralis major. Only two of the three angles (43^{0} and 50^{0}) had a significant effect on decreasing the muscle activity in the sternal pectoralis muscle.
3. The anterior and middle deltoid were more active during the three incline bench presses than during the supine bench press. No difference was found in normalized EMG between the three incline angles.
4. The clavicular pectoralis and sternal pectoralis major can act independently. The level of activation in the clavicular pectoralis major did

not change across all four trials, while muscle activity in the sternal portion decreased significantly.

5. The ascent and descent times were not significantly different during the four bench press conditions.
6. Vertical bar displacement was significantly greater in the 43^{0} and 50^{0} incline bench press than the supine bench press.
7. Peak vertical bar velocity was significantly greater during all three incline bench presses than the supine bench press.
8. Peak vertical downward acceleration was only significant between the supine bench press and the 43^{0} incline bench press. The greater downward acceleration noted during the 43^{0} incline suggested that subjects applied less force against the bar during the eccentric portion of the movement.
9. Changes in peak upward vertical acceleration during the eccentric phase between bench press conditions were not significantly different.
12. Peak upward vertical acceleration during the concentric phase of the movement did not significantly change across the four bench press conditions.

Recommendations

Based on the present study, the following recommendations are made for future studies that intend on using a similar methodology:

1. The filming rate used in the present study may have been too slow. A faster filming rate would give more accurate acceleration values. The faster filming rate would provide more frames to digitize, which in turn would give more vertical acceleration values and the acceleration curves would then have a smoother transition from one value to the next.

2. A more detailed stick figure representation of the human body should be considered for this type of study, especially if one of the purposes of the study is to compare angular changes between segments.
3. Since the level of inclination affected the activation level of three of four muscles, other angles of inclination could be tested such as a 30^{o} incline bench press, or a decline bench press to assess their effects on the activation level of the four muscles.
4. Since muscles such as the triceps brachii, subscapularis, supraspinatus, posterior deltoid are also involved during the bench press exercises, EMG analysis could be conducted to determine their contribution (as the incline bench press may actually be changing their level of activation).
5. Further analysis of the movements occurring in the transverse plane could be used to determine changes in upper-arm angle and could be used to estimate moments occurring about the elbow and shoulder.
6. A third video camera could have been used to film movements in the frontal plane. This would help provide information regarding amount of external rotation occurring as the angle of bench press inclination changed. This camera view could also determine how much humeral adduction is occurring during the bench press movements.

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APPENDIX A

Personal Consent Form

PERSONAL CONSENT FORM

You have volunteered to participate in a study entitled "An EMG analysis of the muscles involved during the incline and supine bench press".

The purpose of this study is to quantify the muscle activity for each muscle during both the incline and supine bench press. These values will then be compared to determine if there are any differences between the contribution of the muscles in the incline and supine bench press.

In the present study, you will first be weighed on a digital scale, and then have your height measured. Your shoulder width will also be measured. Finally, your body characteristics will be assessed using skinfold calipers.

You will be tested on separate days for your one repetition maximum (1-RM) on both the three angles incline (37^o, 43^o, 50^o) and supine bench press (the four tests will be performed on four separate days). The max-test protocol will be as follows: Two spotters will help you lower the bar to your chest. Once the bar is still (for 1 second), you will forcefully push the bar upward until your arms are fully extended. Your feet must remain in contact with the floor during the lift and your buttocks must remain in contact with the bench. This procedure will be repeated until your maximum bench press is determined. The above protocol will also be repeated for the incline bench press. These results will then be used to determine the value of your 75% maximum for both the incline and supine bench press.

The actual testing will require you to perform a three (3) repetition lift using 75% of your one repetition maximum. Four sets of electrodes will be placed on your right shoulder and chest. They will be placed on your anterior deltoid, middle deltoid, sternal pectoralis major, and clavicular pectoralis major. You will be required to perform this movement at pace that is similar to a normal training pace, as determined by a metronome. You will be given an orientation to the use of metronome. The movement will be repeated at three different angles on the incline, and once on the supine bench press.

During the incline and supine bench press, a video camera will film your performance, and the video tapes will used to determine the path of the bar as well as to calculate vertical force exerted on the bar. You will not be identified in any publication: your name, height, weight, biacromial breadth, skinfolds, and bench press performance, will be recorded for identification by the investigator, however, your data will remain confidential. The recorded films and recorded EMG values will not be redistributed or used for any purpose other than this biomechanical research study.

Since you are an experienced and trained weight lifter it is assumed you are capable of performing the one repetition maximum tests as well as the 75% repetition maximum tests on both the incline and supine bench press test, and the risk of injury is low.

I, __________________________, have read the above information and understand the testing procedure, the risks involved, and I agree to participate at my own risk. I acknowledge that the one repetition maximum and the 75% maximum tests are well within my capability and I can successfully perform these tests on a regular basis. I also understand that I have the right to withdraw at any time. In case of injury, I relieve the University of Manitoba and the Investigator of any liability that may arise as the result of my participation.

Signature of the Investigator | Date

Signature of Subject (Parent/Guardian) | Date

Witness | Date
Witness | Date

APPENDIX B

Equipment Specifications

EMG SYSTEM SPECIFICATIONS

II. System Specifications

Preamplifiers

Input:	Differential Two active leads and signal ground Microtech connector used for input
Output:	Single-ended Telephone headset cable
Gain:	100X
Input Impedance:	44 MOhms
CMRR:	> 85 db (typically 95 db)
Sensitivity:	3 uV RMS
Isolation:	Optically coupled to main amplifier Separate power supply

Main Amplifier

Gain:	Continuously adjustable to 100X maximum
System Gain:	10,000 X maximum
Filters:	Low Pass 0.3,1,3,10 kHz High Pass 1,3,10,30,100 Hz Adjustments ganged for channel pairs
Linear Envelope Processing:	Full wave rectification into low pass filter Low pass filter settings. 30 Hz, 50 Hz, bypass (full wave rectified) Adjustable Gain
Audio Output:	Channel selectable Adjustable Gain (Max power output of 200 mW)
LED Bar Graph:	Channel Selectable Full wave rectified signal is displayed 0-5 Volt display range for gain adjustment
Output:	All outputs limited to +/- 5 Volts
Battery:	Rechargeable Gel Cell Batteries Battery Recharger Supplied Charge cycle - 12 hours Separate Battery supply for Preamplifier and Main Amplifier Battery Low Indicators for each supply Charge Position on Power Switch
Fuse:	1 Amp, 14 Volt in all fuse receptacles

ANALOG TO DIGITAL CONVERTER SPECIFICATIONS

ISC-16 Technical Specifications

Analog to Digital Converter

Channels:	1, 2, 4, 8 or 16 single ended
Sampling Speed:	1 MHz maximum, user selectable
Resolution:	12 bit
Relative Accuracy:	0.025%
Absolute Accuracy:	0.1%
Input Range:	±10 volts or ±2.5 volts
Bandwidth:	250 KHz
S/H Aperture Uncertainty:	100 PS
Sampling Clock:	8 MHz Crystal Clock or External Clock
Trigger Logic:	± slope, level, voltage threshold, external trigger
Input Capacitance:	35 pF
Input Impedance:	20K ohms (1M optional)
A/D Memory:	64K double buffered

System Requirements

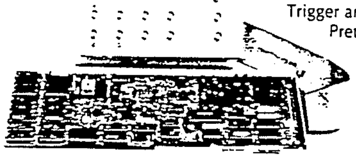
Computer:	PC, XT, AT, 386 or PS/2 Model 25/30
Graphics:	EGA or VGA color graphics card with 256K memory
Monitor:	EGA Graphics Monitor
Printer:	IBM Graphics Dot Matrix, HP Laser Jet II
Plotter:	HP 7470, 7475
Memory:	640K RAM
Operating System:	PC-DOS 2.1 through DOS 3.3

Optional Sample and Hold

Channels:	2, 4, 8 or 16
Percent Accuracy:	0.05%
Aperture Jitter:	0.25 nanosecond
Aperture Delay:	20 nanoseconds
Settling Time:	0.3 microsecond

State-of-the-Art Converter

Faced with continuing advances in technology, you need an acquisition system to keep pace with your imagination and creativity. R.C. Electronics designed the ISC-16 acquisition array processor with the flexibility you require for programming. Unlike other A/D boards, the ISC-16 operates independently of the PC. You are no longer required to set up DMA channels, timing loops, or assembly subroutines to run your system efficiently. Commands in C, BASIC, FORTRAN, and a variety of other languages can be used to define the sample rate, active channels, number of data points, and trigger conditions or directly addressing the hardware through I/O ports. Data handling flexibility starts with a double buffered memory configuration which allows computer access to one memory bank while the other bank is acquiring new data. Allowing direct access to the on-board memory eliminates the need for data transfer between the A/D and computer memories. Other unique features, such as External Trigger and Sample Clock inputs, Pretrigger capability, and optional Sample and Hold are available to work for you regardless of how challenging your project becomes.



Waveform Synthesis and Control

R.C. Electronics provides a full line of both digital I/O and analog output products to complement the ISC-16 data acquisition system. Our RC-200 line consists of an RC-202 base-board with optional piggyback modules designed to meet a variety of stimulus generation and control requirements. Each RC-202 contains four 12-bit DACs along with 8 digital inputs and 8 digital outputs. With this hardware you can respond to external control inputs while generating both digital and analog control signals. Adding the RC-204 timer module provides 4 independent pulse train generators plus a master synchronization clock for triggering other instruments. RC-216 modules can be added for ultra-high accuracy 16-bit arbitrary waveform generation with output rates of up to 1 MHz from the 64K on-board memory. Please contact us for more information on these products.

RC Electronics Inc.

R. C. Electronics, Inc.
5386-D Hollister Avenue
Santa Barbara, CA 93111
Phone: (805) 964-6708
FAX: (805) 964-4906

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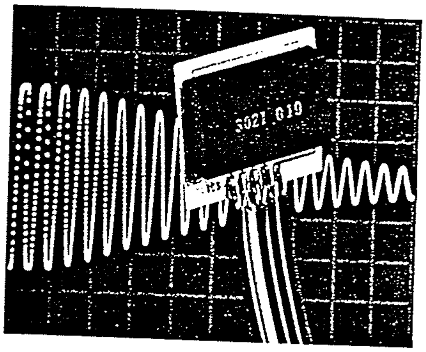
LINEAR ACCELEROMETER SPECIFICATIONS

Technical Note TN-008

IC SENSORS **Silicon Accelerometers**



Introduction

Batch-fabricated silicon accelerometers open up a wide variety of applications because of small size, low mass, and low cost. The unique features of the devices include the ability to precisely control mass, spring thickness, damping, and overforce stops without the need to individually hand-assemble the accelerometers. One of a series of accelerometers from IC Sensors, the Model 3021, is shown above.

Integrated Accelerometer Designs

The integrated, batch-fabricated accelerometer is the product of evolutionary silicon sensor development efforts. The first reported silicon accelerometer was made by Roylance in 1976. This device was simply a silicon spring with a silicon mass attached to it. The device featured diffused piezoresistors to allow batch fabrication of the device. Acceleration causes the mass to move with respect to the frame, creating stress in the piezoresistor, which changes its resistor value. A conceptual model of this early device is shown in Figure 1. This accelerometer was configurable as a half bridge with only one of the two resistors being sensitive to stress.

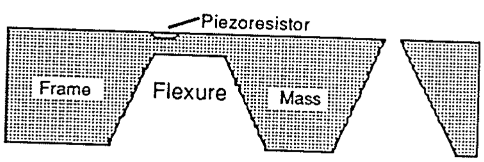


Figure 1
Simple Cantilever Beam Accelerometer

The device has evolved from these early beginnings to the structure, developed by IC Sensors, shown in Figure 2. Here the device still features the silicon mass and integrated piezoresistors but instead of only a half bridge, the device now incorporates a full bridge design. Further, the device has been upgraded from a single cantilever to a double cantilever structure. This substantially reduces off-axis sensitivity. Further improvements, including ion-implanted resistors and a unique process sequence, result in precisely positioned and matched piezoresistors. Thus, optimum stress concentration is achieved.

The resultant electrical structure is a Wheatstone bridge where two resistors increase with downward acceleration and two decrease with the same force. This is shown schematically in Figure 3. One advantage of the piezoresistive bridge is that the device can measure true DC response, unlike several other types of accelerometers; this enables measurements of slow transients and low frequency vibration. Further, the device can be directly connected to other electronic equipment without concern for electrical loading of the output or possible damage due to static discharge.

A key feature of the IC Sensors' piezoresistive bridge accelerometer is that it can be easily adapted to precision trimming for temperature, gain and offset correction, just as the more conventional piezoresistive pressure sensors are. The three layer sandwich used by IC Sensors and shown in Figure 2 also provides mechanical stops to prevent damage to the device in handling and shipping.

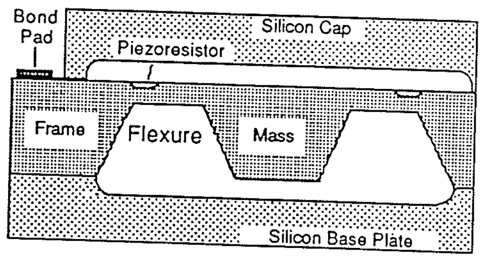


Figure 2
Double Cantilever Silicon Accelerometer
with Overforce Protection

Comparison With Existing Technologies

There are presently three main types of accelerometers which offer reasonably small size. These are the capacitive, piezoelectric, and piezoresistive types. In each of these classes, there are both integrated and non-integrated structures. The integrated units are ones which lend themselves to batch-fabrication on silicon or other substrates. The three general classes are compared in Table 1.

This table is representative of the trade-offs in selecting a device, although there are exceptions to this list depending on acceleration range and application. The strengths of the integrated silicon accelerometer using piezoresistive transduction are highlighted in the table. While size is usually a key advantage for the piezoelectrics, the piezoresistive device, for high (≥ 100g) acceleration ranges, can be just as small, or smaller, with built-in damping, with the ability to shunt calibrate, and with the low impedance output.

Because the device is responsive down to DC, the piezoresistive accelerometer can be used to show orientation of the mounting structure (i.e. for use in inclinometry). As shown in Figure 4, the structure can be rotated through 180° and shows plus and minus one g acceleration due to the earth's gravity. The added advantage of this, as noted in Table 1, is that the calibration of the device can be verified by rotating the device through 360°, recording the high and low readings, taking the difference, and dividing by 2 to give the sensitivity. Unlike the piezoelectric device, the piezoresistive device responds to both slow and fast changes in acceleration.

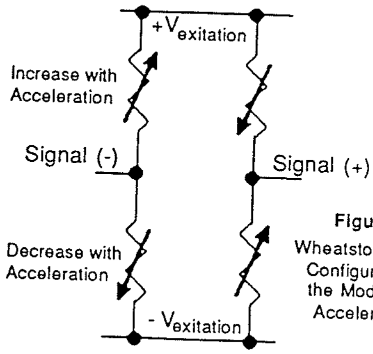


Figure 3
Wheatstone Bridge Configuration for the Model 3021 Accelerometer

Damping in Silicon Accelerometers

The silicon accelerometer is a nearly ideal mechanical structure. Because the loss factors in the silicon accelerometer are so low, the inherent damping of the device is also low. As a result, the accelerometer can have significant gain at resonance. Typical undamped gains of 30 to 200 have been reported at resonance. The gain of the Model 3021 accelerometer with damping factors of 31, 0.707, and 0.05 is shown in Figure 5. Higher damping factors result in lower gain at resonance.

The key problem with undamped units is that the device must be able to withstand the high gain at resonance, and, further, the electronics which interfaces with the device must also be able to filter out this resonance and any harmonic distortion

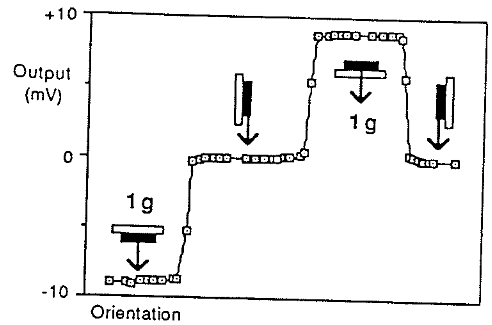


Figure 4
Effect of Rotation on the Accelerometer Output

due to cross-products of the resonance and the signal. This is a technologically difficult problem and as a result, controlling the magnitude of this resonance in the structure is critical.

IC Sensors controls the damping factor of the accelerometers to provide devices with critical or near critical damping. This prevents over-force damage and further reduces the possibility that the device will hit the mechanical overstop due to high amplitude gain at resonance. The design of the accelerometer is such that damping factors can be specified during wafer fabrication and lower damping is achievable, although it is recommended that the minimum damping be limited to result in a gain of 10 at resonance. This results in the top curve shown in Figure 5. These under-damped devices are substantially more prone to output saturation and distortion than are the critically damped accelerometers.

Off-axis Sensitivity

All accelerometers are sensitive to forces other than in the principle sensing direction. Three fundamental reasons exist for this non-ideal sensitivity. The first is that the accelerometer, even if ideal, is still sensitive to angular errors when mounted

Table 1
Comparison of Sensing Technologies

	Capacitive	Piezoelectric	Piezoresistive
Impedance	High	High	Low
Electrical Loading Effects	Very High	High	Low
Size	Large	Small	Moderate
Temperature Range	Very wide	Wide	Moderate
Linearity Error	High	Medium	Low
DC Response	Yes	No	Yes
AC Response	Wide	Wide	Less Wide
Damping Available	Yes	No	Yes
Sensitivity	High	Moderate	Moderate
Zero Shifts due to shock	No	Yes	No
Turn-over or Shunt Calibration	Yes	No	Yes
Electronics Required	Yes	Yes	No
Cost	High	High	Low
Cross-axis Sensitivity	Primarily dependent on mechanical design, not transduction		

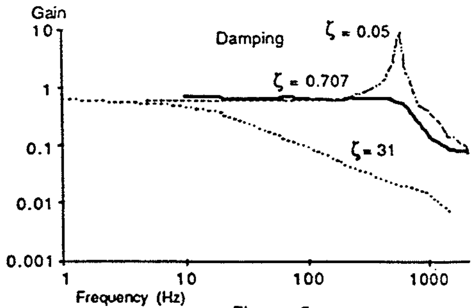


Figure 5
Effect of Damping on Frequency Response

in a system. The second is that the center of mass and the center of bending are not co-planar in all axes. The final limitation is in the matching of the piezoresistors.

With respect to mounting angle sensitivity, the effects can be quite dramatic. For a 1° error, the sensitivity is only degraded by 0.02%; for a 5° error, the effect will be a 0.4% gain error. If the device is mis-mounted by 8°, then the accelerometer will register a loss in gain of 1% in the principle axis and will have a net increase in sensitivity in the direction of mounting to 1%. This means that over the length of the Model 3021, which is 15.4 mm, the surface must be flat to better than 1.8 mm. Note that for shorter devices or if the device has a post to allow bolting down, it is sometimes difficult to assure flatness; the flat ceramic substrate used in the Model 3021 thereby helps in minimizing the off-axis sensitivity. Nonetheless, care must be used when the accelerometer is mounted in a system.

The more fundamental problem is one of having the center of mass and the center of bending non-coplanar. This is shown in Figure 6. In the conventional force direction, the beams bend vertically and the center of mass moves downward. If an acceleration is applied from the left to the right, then the mass will tend to rotate about the center of mass and one spring will shorten and torque upward while the other will lengthen as it is torqued downward. If the center of mass were in the plane of the flexures, then an off-axis acceleration would result in pure shortening of one flexure and pure lengthening of the second. The difference in bending moments in these two cases results in a minimal signal when the piezoresistors are well matched.

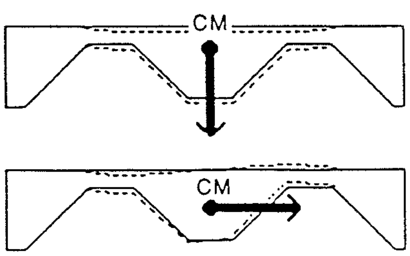


Figure 6
Vertical and Horizontal Loading of the Accelerometer

The third area which contributes off-axis sensitivity is in resistor placement and matching. Ideally, two resistors increase and two resistors decrease by the same amount with applied acceleration; slight mis-alignments will produce one or more resistors which are less sensitive than the others, and, therefore, torquing of the mass, as will happen due to off-axis loading, will produce a smaller change in those resistors than will be produced by the corresponding resistors on the opposite side of the mass. Note that these effects are set by the efficiency of the stress collection and not by the resistor matching. The piezoresistive bridge can be perfectly balanced at zero g's and can show significant off-axis sensitivity. Processing tolerances become the dominant issue in setting off-axis limits.

It should also be noted that the off-axis sensitivities can be minimized by careful mounting of the devices; Figure 7 shows a typical nulling curve. The procedure to provide this null is time consuming and if the device can be used without resorting to nulling, then it is recommended that technique not be attempted.

Accelerometer Mounting Considerations

The accelerometer measures motion of a surface. Consequently, coupling between the accelerometer and the surface is critical to assure a high fidelity signal. The accelerometer is a simple spring/mass system. The insertion of a weak coupling

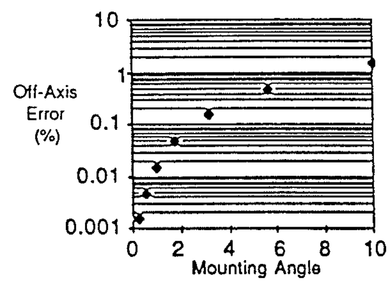


Figure 7
Off-Axis Sensitivity Vs Mounting Uncertainty

media, such as an RTV, results in a secondary spring/mass system related to the mass of the total package and the elastic nature of the coupling agent. This effect is shown in Figure 8. The 3000 Series accelerometers are housed in a silicon casing with the mass representing less than 12% of the total housing weight. For the Model 3021 device, the casing is mounted on a ceramic substrate and the mass of the housing is then further increased by a factor of 7. Thus the total mass of the package is 60 times the seismic mass and thus a spring interface which is 60 times stiffer than the accelerometer spring will have the same resonance. Further, this parasitic spring may be relatively low-loss and result in a high Q, low damping subsystem. As a result, the very low overall mass of the 3000 Series accelerometers directly reduces the constraint on mounting.

Because of these factors, care in mounting should be exercised. It is recommended that a hard epoxy be used to mount the accelerometer to the surface under study. Thick glues should generally be avoided as the coupling spring increases linearly with layer thickness. Cyanoacrylate base cements (instant bonding)

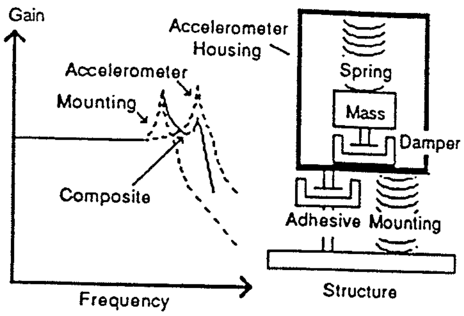


Figure 8
Effect of Secondary Spring-Mass Mounting Resonance on Composite Frequency Response

offer excellent adhesion while minimizing the thickness of the glue.

Prior to application of the sensor to the surface, both surfaces should be degreased. Acetone or TCA are suggested cleaning solvents. Both surfaces should be allowed to dry and then a drop of adhesive should be applied to the mounting surface. The sensor substrate should be pressed against the surface; pressing lightly with fingers on the top cap normally provides adequate force.

To remove the sensor after the application is complete, it is suggested that a bead of acetone be run around the substrate, allowed to soak for 5 to 15 minutes and then torque should be applied using a crescent wrench or similar tool to apply rotation to the lower edges of the ceramic case. Do not torque by attaching to the black cap.

As a temporary mounting approach, the device can be attached using RTV. While such a technique is not generally recommended both because of the mounting resonance and because of the mounting angle uncertainty, the approach may be viable for critically damped accelerometers having the low mass of the 3021. For a 1/4 inch thick layer of RTV, the effective resonance should be in the 100 to 300 Hz range, below the seismic resonance and within the useful range of the accelerometer. A thin layer of RTV (<0.06 mm) should result in a resonance in the 10 to 30 kHz range. Thus, as can be seen by this example, careful selection of mounting materials and thicknesses is critical for proper operation of even an extremely light accelerometer such as the Model 3021.

UNDER NO CIRCUMSTANCES SHOULD AN ACCELEROMETER BE STRUCK TO FREE IT NOR SHOULD THE DEVICE BE PRIED UP. BOTH APPROACHES CAN DAMAGE THE UNIT.

Cable Considerations

The forces exerted on this series of accelerometers due to cable whip and torque can be significant. Unlike the more conventional accelerometers which are relatively heavy and attached to the measurement surface with mounting studs, an advantage of these ultra-light accelerometers is that they can be epoxy mounted and demounted by torquing the device. The cable can provide one source of torque. Further, the sensor / cable system can effectively dampen or alter the movement of the structure under test. For this reason, it is recommended that the cable selected for the accelerometers be extremely flexible, especially in the direction of sensing.

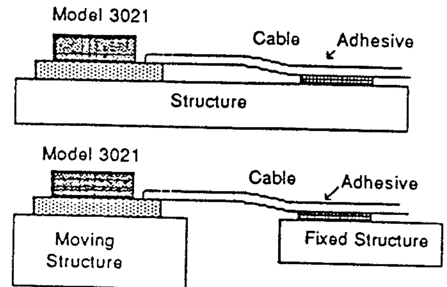


Figure 9
Mounting Examples for Accelerometers

It is recommended that the cable be taped down in the vicinity of the accelerometer but that the area nearest the sensor for the first few centimeters be allowed to be free and not glued or taped down. This approach will tend to reduce coupling and torque problems but will still allow relative ease of mounting and use. Care should be taken not to clamp or cement the cable near the accelerometer as this will tend to localize stress in the cable and may lead to breakage of the cable near the sensor. Examples of mounting are shown in Figure 9.

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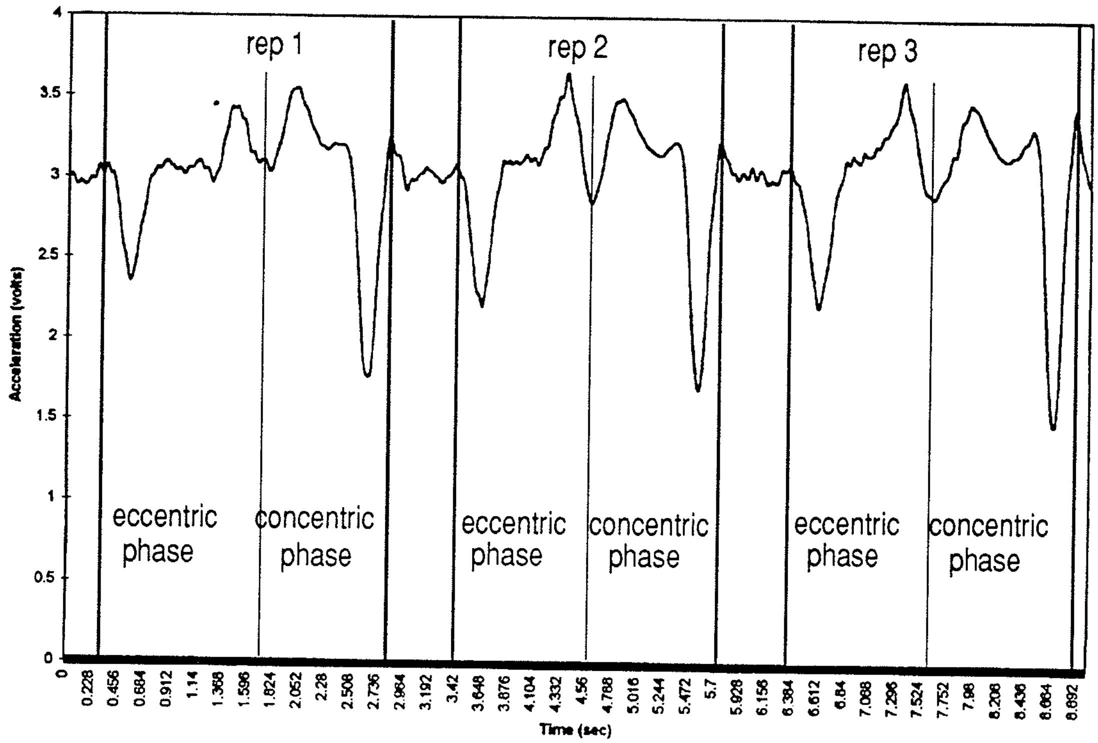
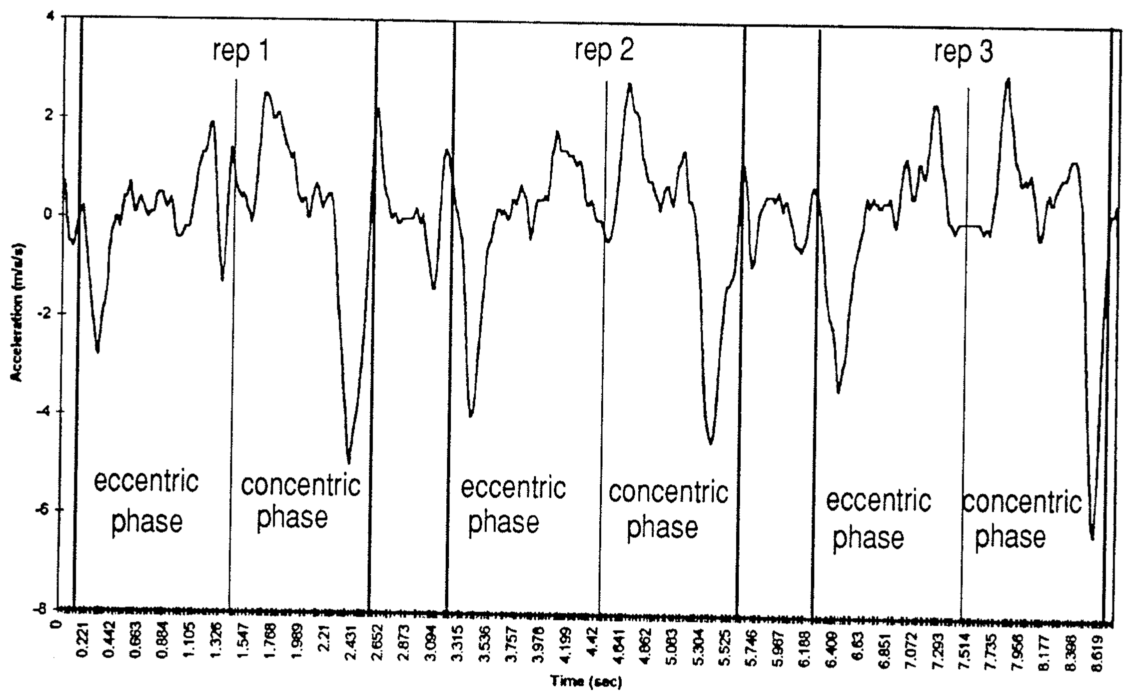
APPENDIX C

Comparison of Acceleration Between The Linear Accelerometer and Video Film Data

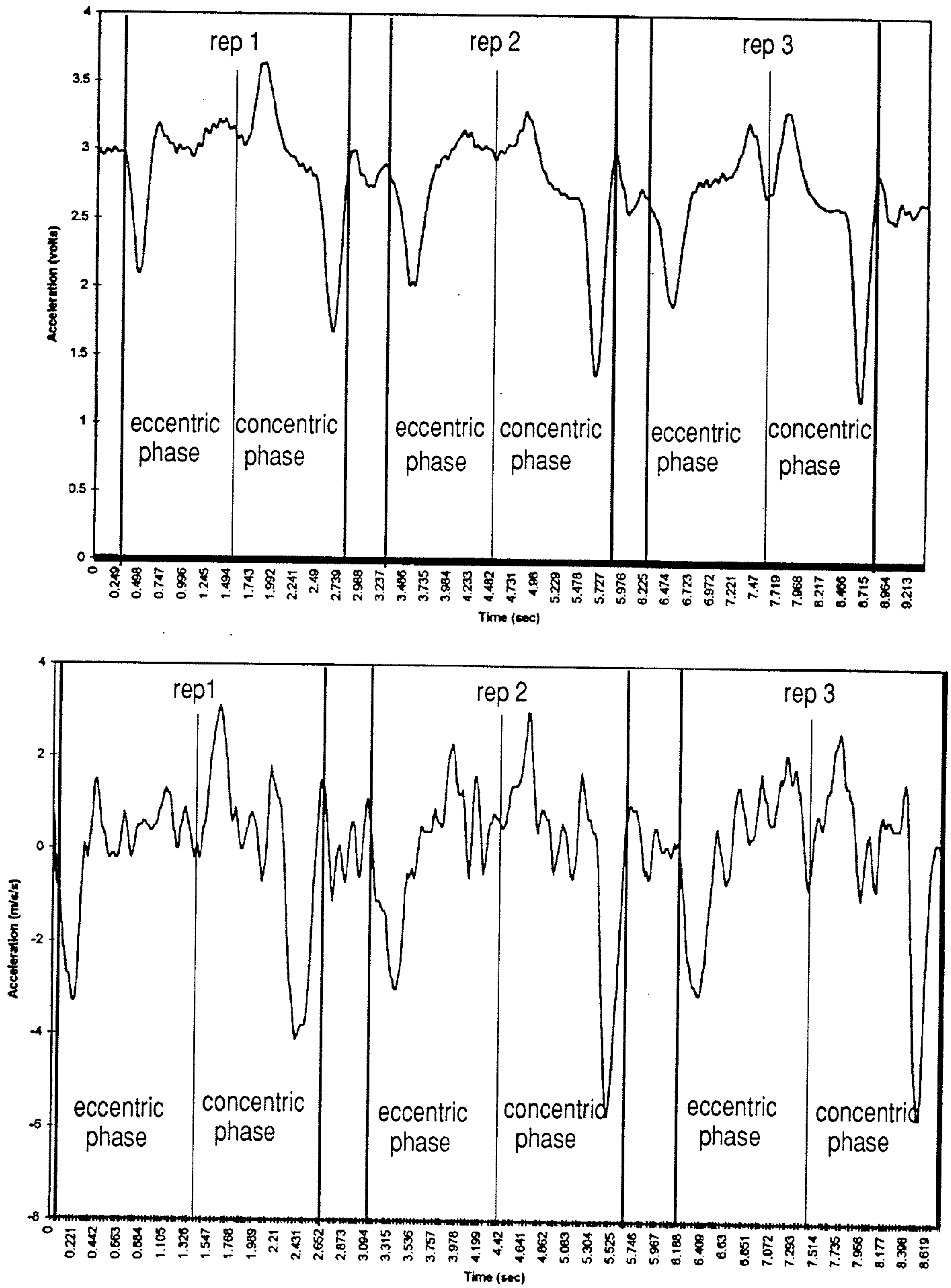
ACCELERATION PROFILE COMPARISON





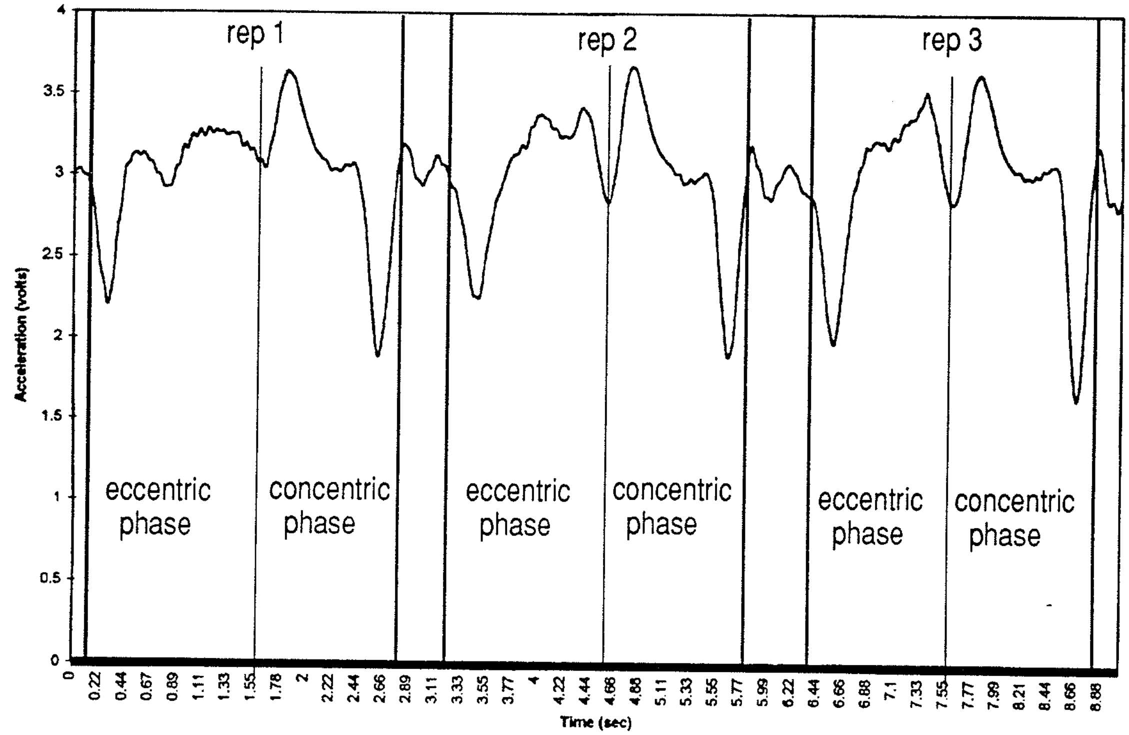
Acceleration profile for the supine bench press recorded from: (a) linear accelerometer (sampled at 333 Hz), and (b) video film data (sampled at 60 Hz). Both were smoothed using a 5 Hz Butterworth digital filter.

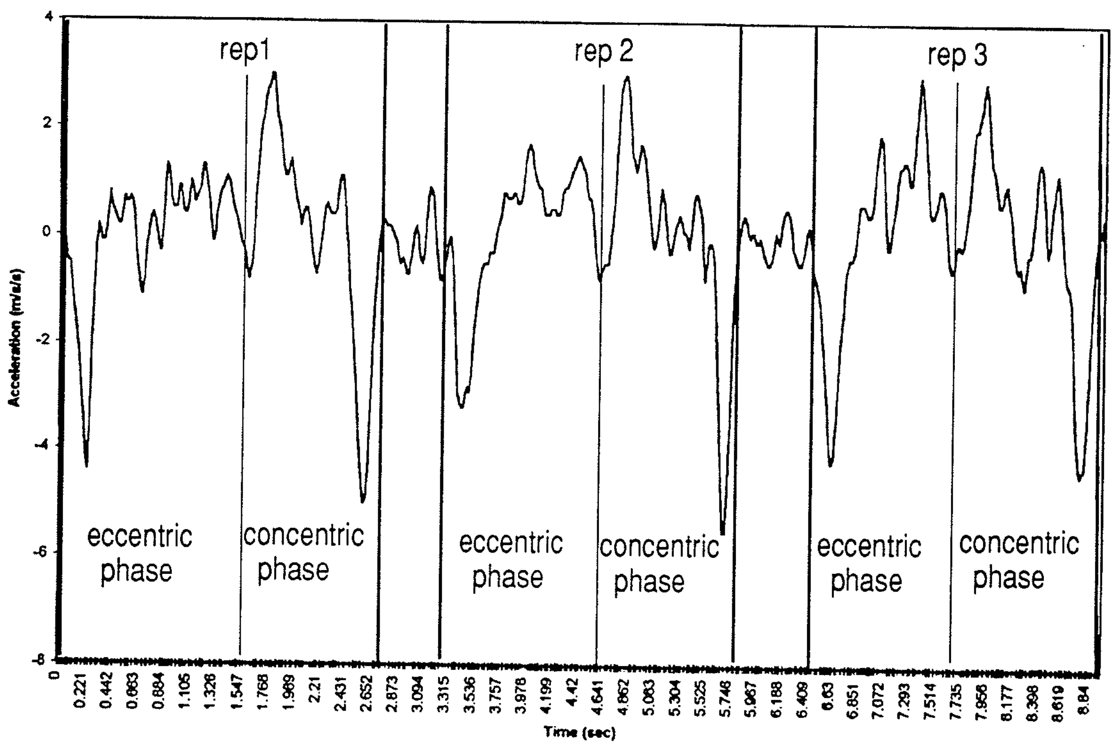
ACCELERATION PROFILE COMPARISON



Acceleration profile for 37^0 bench press recorded from: (a) linear accelerometer (sampled at 333 Hz), and (b) video film data (sampled at 60 Hz). Both were smoothed using a 5 Hz Butterworth digital filter.

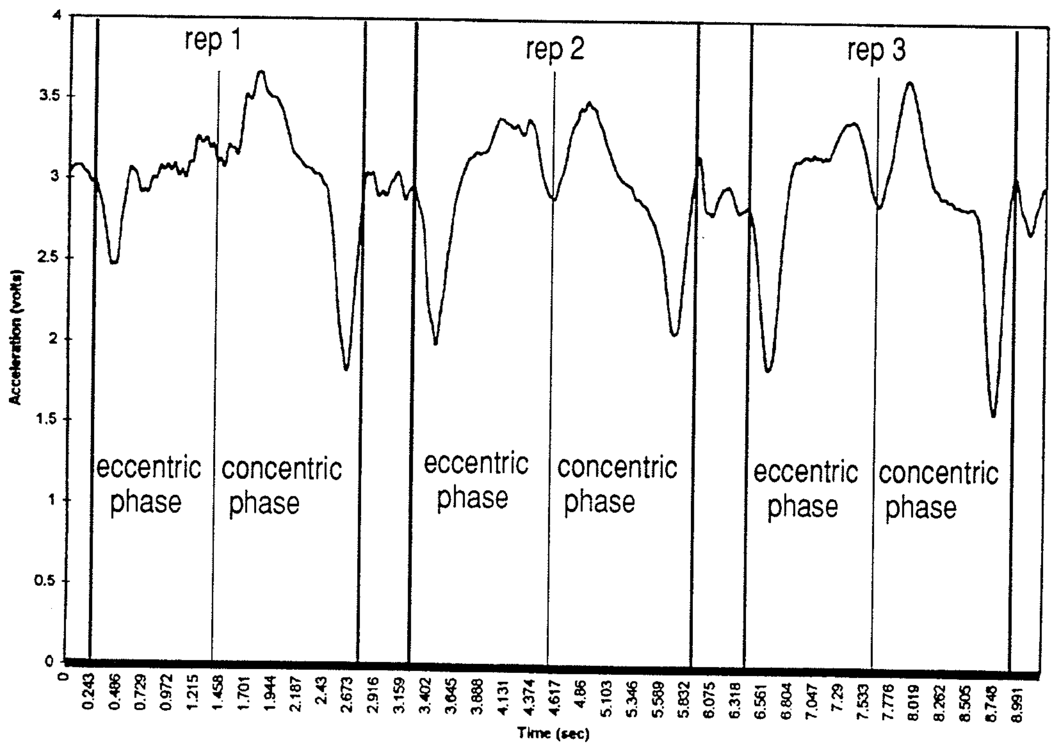
ACCELERATION PROFILE COMPARISON

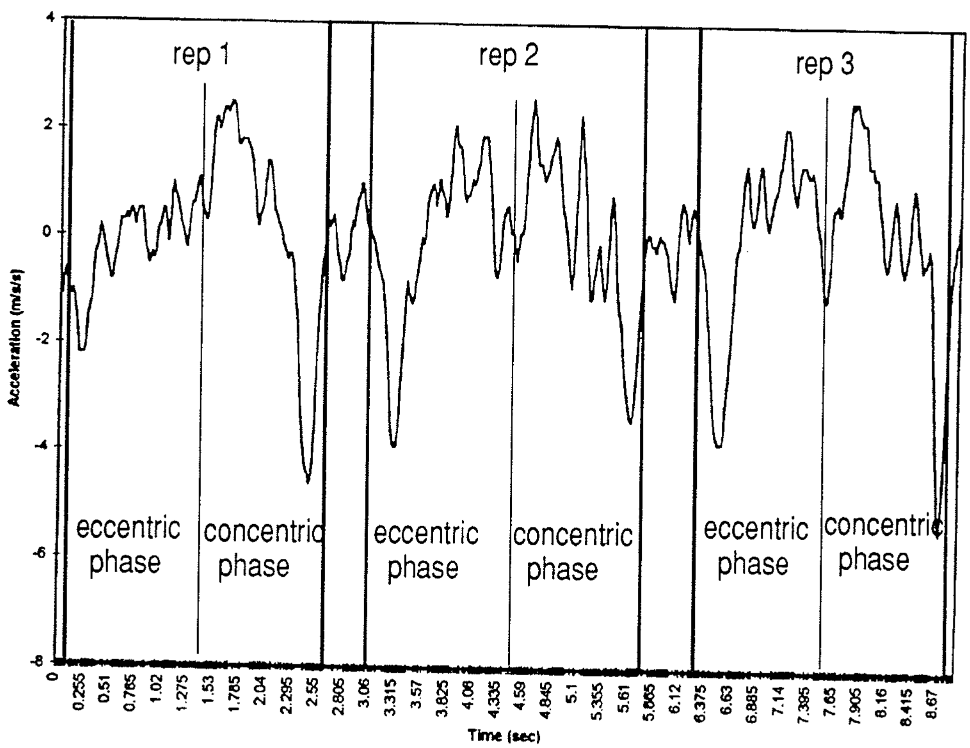




Acceleration profile for 43^{0} bench press recorded from: (a) linear accelerometer (sampled at 333 Hz), and (b) video film data (sampled at 60 Hz). Both were smoothed using a 5 Hz Butterworth digital filter.

ACCELERATION PROFILE COMPARISON



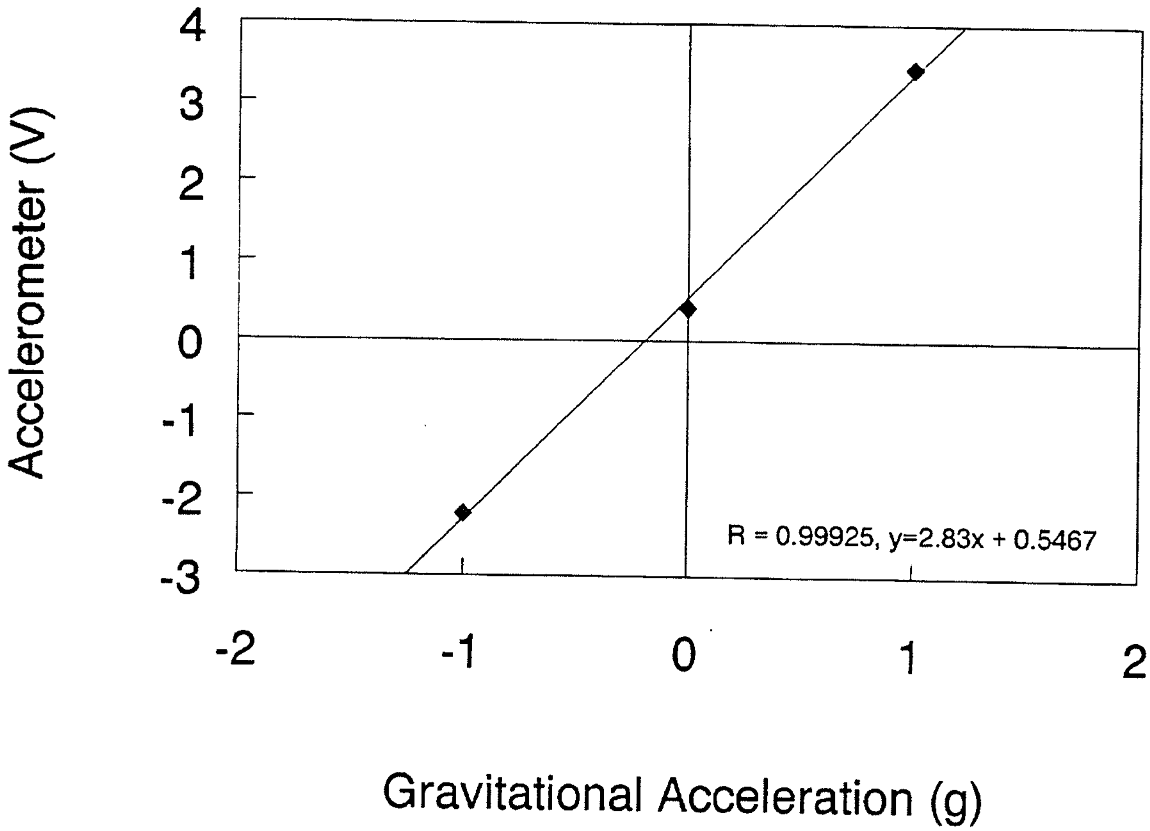


Acceleration profile for 50⁰ bench press recorded from: (a) linear accelerometer (sampled at 333 Hz), and (b) video film data (sampled at 60 Hz). Both were smoothed using a 5 Hz Butterworth digital filter.

APPENDIX D

Accelerometer Calibration

ACCELEROMETER CALIBRATION



APPENDIX E

Sample Statistics

ANOVA Table for Normalized Eccentric Clavicular pectoralis

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	5.683	.437		
Category for Normalized Ec...	3	.015	5.101E-3	.071	.9750
Category for Normalized Ec...	39	2.793	.072		

Reliability Estimates - All Treatments: .847; Single Treatment: .581

ANOVA Table for Normalized eccentric sternal pectoralis

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	5.433	.418		
Category for Normalized ec...	3	2.181	.727	10.188	<.0001
Category for Normalized ec...	39	2.783	.071		

Reliability Estimates - All Treatments: .717; Single Treatment: .388

ANOVA Table for Normalized eccentric medial deltoid

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	14.410	1.108		
Category for Normalized ec...	3	5.970	1.990	11.124	<.0001
Category for Normalized ec...	39	6.977	.179		

Reliability Estimates - All Treatments: .722; Single Treatment: .394

ANOVA Table for Normalized eccentric anterior deltoid

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	23.749	1.827		
Category for Normalized ec...	3	4.345	1.448	5.757	.0023
Category for Normalized ec...	39	9.812	.252		

Reliability Estimates - All Treatments: .815; Single Treatment: .525

ANOVA Table for Concentric normalized calvicular pectoralis

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	4.795	.369		
Category for Concentric no...	3	.284	.095	1.762	.1705
Category for Concentric no...	39	2.097	.054		

Reliability Estimates - All Treatments: .846; Single Treatment: .579

ANOVA Table for Concentric normalized sternal pectoralis

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	5.062	.389		
Category for Concentric no...	3	.832	.277	4.590	.0076
Category for Concentric no...	39	2.358	.060		

Reliability Estimates - All Treatments: .805; Single Treatment: .508

ANOVA Table for Concentric normalized middle deltoid

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	16.060	1.235		
Category for Concentric no...	3	5.155	1.718	6.903	.0008
Category for Concentric no...	39	9.708	.249		

Reliability Estimates - All Treatments: .714; Single Treatment: .384

ANOVA Table for Concentric normalized anterior deltoid

	DF	Sum of Squares	Mean Square	F-Value	P-Value
Subject	13	18.875	1.452		
Category for Concentric no...	3	2.629	.876	4.565	.0078
Category for Concentric no...	39	7.486	.192		

Reliability Estimates - All Treatments: .834; Single Treatment: .557

APPENDIX F

Pilot Study

PILOT STUDY

METHODS

A pilot study was conducted on April 21, 1993, at the School of Medical Rehabilitation at the University of Manitoba. Prior to the pilot study, the subject was required to perform a series of preliminary tests to determine the 1 RM for the four bench presses. The subject's characteristics, as well as his 1 RM for both movement conditions are listed in Table F-1. The goal of the pilot study was: (1) to collect data that would provide the investigator with the opportunity to gain some practical experience in the analysis of data, (2) to make comparisons between the four movement protocols, (3) to determine if the proposed study was a feasible undertaking, and to collect and analyze preliminary EMG data.

Table F-1 Subject characteristics

Age (yrs)	Mass (Kg)	Supine 1 RM (Kg)	75% RM (Kg)	43° Incline RM (Kg)	75% RM (Kg)
25	86.4	125	93.75 (93.18)*	93.18	69.88 (70.45)*

* Actual load used for the pilot study.

EMG procedures

Three sets of Beckman silver-silver chloride (type 650 418) bi-polar surface electrodes (8 mm diameter) were attached over the muscle belly of the left anterior deltoid, the left clavicular pectoralis major, and the left sternal pectoralis major. The electrodes were placed as described previously, with the exception that the electrodes were placed directly on the muscle belly (the approximate location of the motor point where the EMG signal would be extremely strong). The preamplifiers were affixed to the bench, directly behind the subject's left trapezius.

The EMG signal was differentially amplified using a BioSys battery powered amplifier and band passed at (10- 1000 Hz, $\pm$ 3 dB) with a roll-off of 6 dB per octave. The EMG signal was amplified with overall system gains ranging between 500-2000. The signal was processed using full wave rectification into a low pass filter of 30 Hz.

The data was collected on-line for the duration of the 4 repetitions at a sampling rate of 128 Hz per channel by a 486 IBM computer equipped with a 16 channel, 12 bit analog-to-digital converter. The EMG data was then stored on disk for later analysis.

Video analysis

All movements were filmed at a nominal rate of 60 Hz by a Panasonic Omnimovie VHS camcorder (PV-460-K) which was mounted to a rigid tripod. The camera was positioned in line with the bar's plane of motion (i.e., at 90 degrees to the subject's sagittal plane).

The weight rack was measured and filmed prior to the experiment to provide the necessary scaling factor for positional data. The following landmarks were digitized for the video analysis (Figure 3-4): the center of the bar; the axis of rotation for the hip, the glenohumeral joint, and one reference point (located on the weight rack).

Prior to digitizing, the movements were reviewed to determine a starting point; after which every other frame of the movement was digitized. The downward and upward phases of the lift were digitized together to provide a complete movement profile. The movement was digitized from just prior to the downward movement of the bar and until the bar had reached the maximum upward distance from the chest.

EMG analysis

In order to analyze the EMG data, it was first necessary to digitize the video data to determine the onset of the descent and the end of the ascent phases for each repetition. Since every second frame was digitized, the corresponding time per frame was 0.033 seconds. Once the total number of frames for each repetition was determined, the total was multiplied by the time elapsed per frame (0.033). Therefore, the total time of a given repetition was then used to determine a repetition time for the EMG data.

RESULTS

Since only one subject was used in the pilot study, the data was of a descriptive nature. The EMG data was limited in terms of generalizability and comparability, so the data is presented descriptively. Figure F-1 and Table F-2 represent the ratio of mean EMG activity (expressed as a proportion of the mean supine bench press EMG activity) for a typical repetition during four angles of the bench press. Anterior deltoid was most active during the 43 degree incline (1.04), and least active during the 37 degree incline (0.748). The ratio of mean EMG activity for anterior deltoid was similar between the flat (1.0) and the 50 degree angle (0.994). The clavicular pectoralis muscle was most active during the flat bench press (1.0) and least active during the 37 degree incline. The ratio of mean EMG activity for clavicular pectoralis was similar in both the 43 and 50 degree inclines (0.774 and 0.784, respectively). The order of the ratio of mean EMG activity, from least to most, for the sternal pectoralis was as follows: flat bench (1.0), 37 degree incline (2.266), 50 degree incline (2.368), and the 43 degree incline (2.509).

Table F-2 Mean normalized EMG activity for 1 subject in pilot study.

Bench press angle	Clavicular pectoralis	Sternal pectoralis	Anterior deltoid
Flat	1	1	1
37^0	0.548	2.266	0.748
43^0	0.774	2.509	1.04
50^0	0.784	2.368	0.994

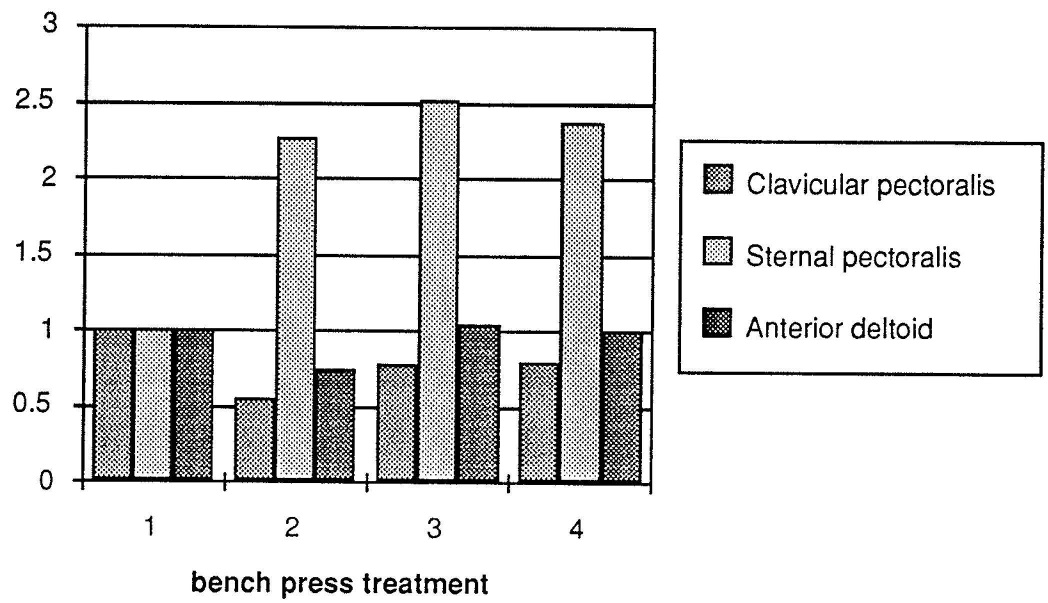


Figure F-1 Representative mean normalized EMG activity for the supine bench press (1), and three angles of the incline bench press (2). Note the incline angles are as follows: 2 - 37^0, 3 - 43^0, and 4 - 50^0 (measured to the horizontal).

Since the video film data was limited in terms of generalizability and comparibility (due to small n=1), the data is presented descriptively. Table 4-2 is a compilation of data collected using video images and digitized data. The most interesting results noted in Table F-3 are the horizontal and vertical displacement of the bar and mean ascent and descent times.

Table F-3 Selected kinematic parameters of four representative bench presses

Bench press angle (degrees)*	Mean repetition time (sec)	Mean descent time (sec)	Mean ascent time (sec)	Mean horizontal displacement (m)	Mean vertical displacement (m)
flat	2.16	1.06	1.10	0.086	0.577
37	2.18	1.12	1.06	0.084	0.633
43	2.36	1.17	1.19	0.066	0.666
50	2.34	1.20	1.14	0.066	0.660

*angles are reported relative the horizontal

The angle of the bench relative to the horizontal appeared to affect the mean ascent path of the bar as the horizontal displacement decreased with increasing angle of bench, while the vertical displacement of the bar increased as the angle of bench press increased (Figures F-2, F-3, F-4, F-5). A more vertical bar path was noted for the three angles of the incline than the supine bench press. Ascent and descent times for all of the bench press conditions appeared to be relatively consistent.

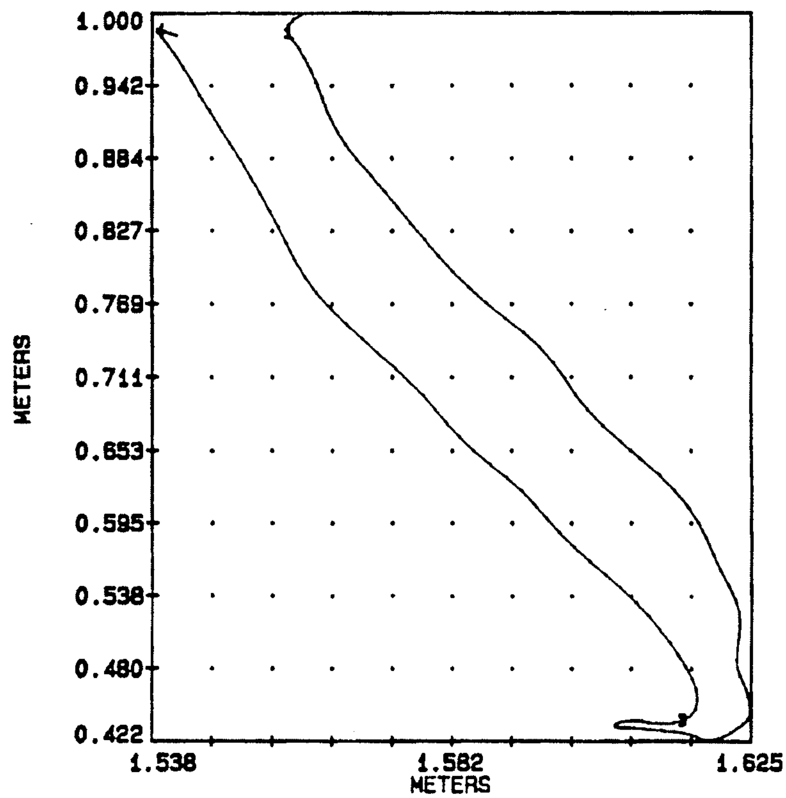


Figure F-2 Horizontal and vertical bar displacement for supine bench press.

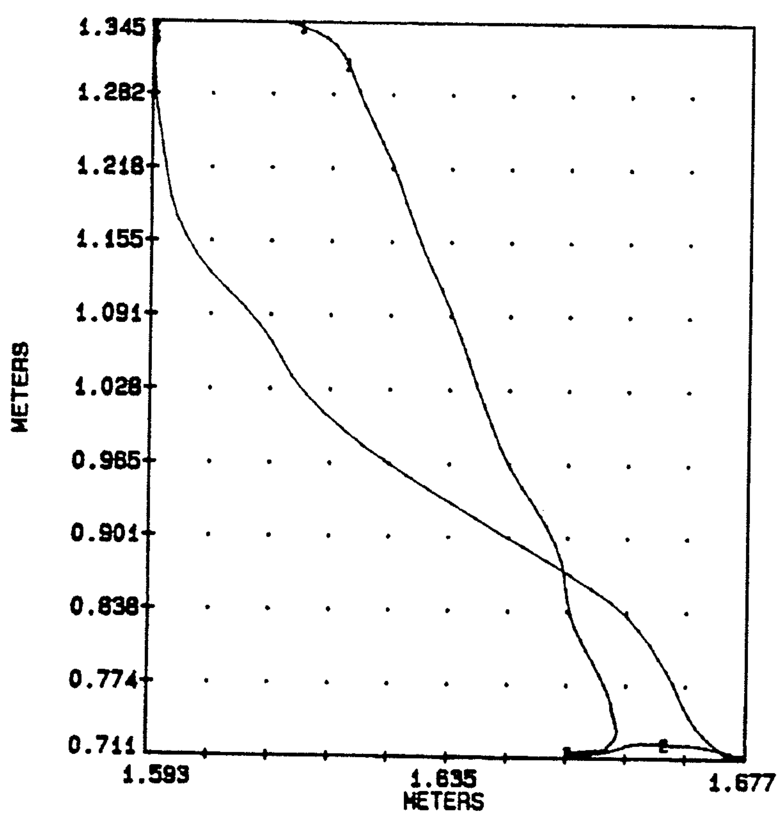


Figure F-3 Horizontal and vertical bar displacement for 37° incline bench press.

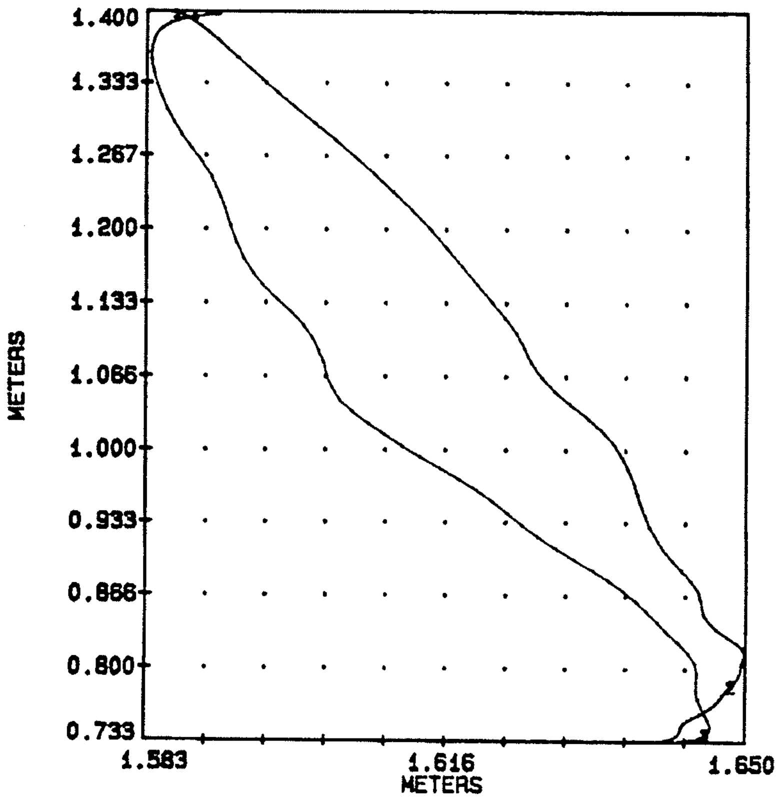


Figure F-4 Horizontal and vertical bar displacement for 43° incline bench press.

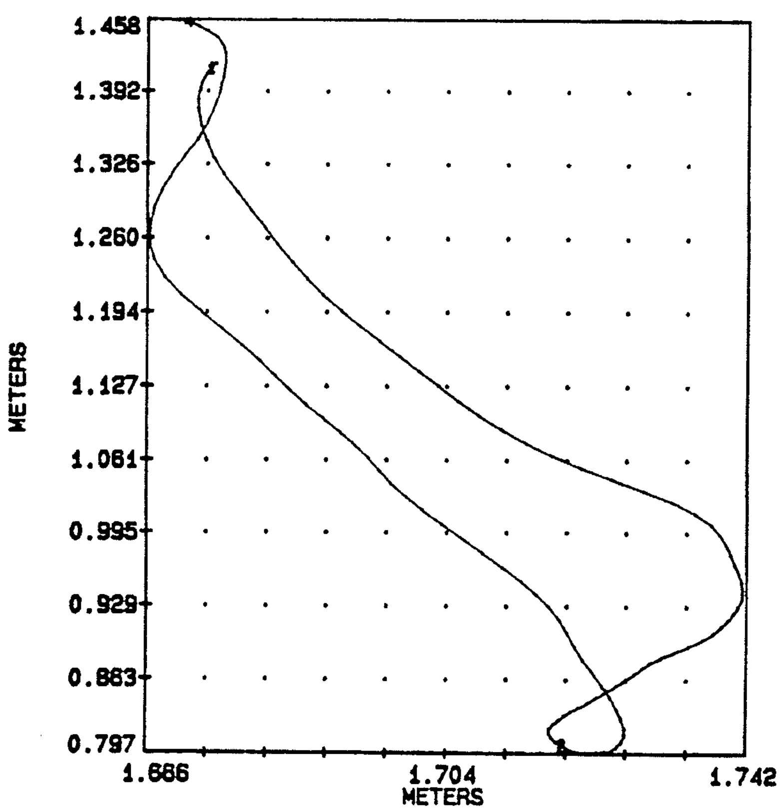


Figure F-5 Horizontal and vertical bar displacement for 50° incline bench press.

DISCUSSION

The results of the pilot study must be viewed with caution due to the use of only one subject. However, based on the results from a single subject, the activity of clavicular pectoralis was least in the incline position and greatest in the supine position. This was not in agreement with the hypothesis that no significant differences in muscle activity would exist between the supine bench press and the three angles of the incline bench press. As well, sternal pectoralis was least active during the supine bench press and most active in the angles of incline bench press. The results provide insight into, and disagree with the current assumption suggesting that the incline bench press is better for imposing a training effect on the clavicular pectoralis muscle. The results of the pilot study also suggest that the supine bench press is not as effective in eliciting greater muscular activity in the sternal pectoralis.

The results noted in the pilot study were not consistent with the results of the present study. In fact, the sternal portion of the pectoralis major muscle in the pilot study was the most active while it was the least active in the present study. In addition, the level of activation of anterior deltoid did not change significantly from the supine to the incline bench press conditions. This was not observed in the present study where the level of activation was significantly greater at all three incline angles than the bench press. Finally, the level of activation for the clavicular portion of the pectoralis muscle was less during the three angles of the incline bench press than the supine bench press. The level of activation for this muscle did not change significantly during the four bench press conditions.

The differences noted between the pilot study and the present study were likely due to differences in data collection and lack of standardizing certain test protocols during the pilot study. For example, the subject in the pilot study was instructed to perform the repetitions without the use of a metronome and thus the movement pace was subject to greater variation. The method used to determine the onset of each repetition was determined by examining digitized video data which may have not been as accurate as using a linear accelerometer mounted to the bar. Some of the data may have been excluded as a result. The EMG data was only assessed for one repetition at each condition which likely increased the variability of the results, whereas the mean sum of EMG over three repetitions at each condition was calculated for the present study. In addition, the EMG data during the pauses in the transition from descent to ascent and vice versa for the present study were not calculated . Therefore, any variability within the repetitions was minimized by averaging all three repetitions and limiting data that was not pertinent to the two phases of the movement. Finally, the pilot study calculated the average EMG for all the raw data points collected which would tend to bias the results towards the higher values. The present study calculated the sum of each individual point of data and then gave a mean value based on three repetitions. This again helped to reduce any variation within the three repetitions.

Although the movement pace was not constrained during the pilot study, the ascent and descent times across the four conditions were similar to those used in the present study, indicating that the subject used in the pilot study was skilled and was able to perform the required movements with consistency.