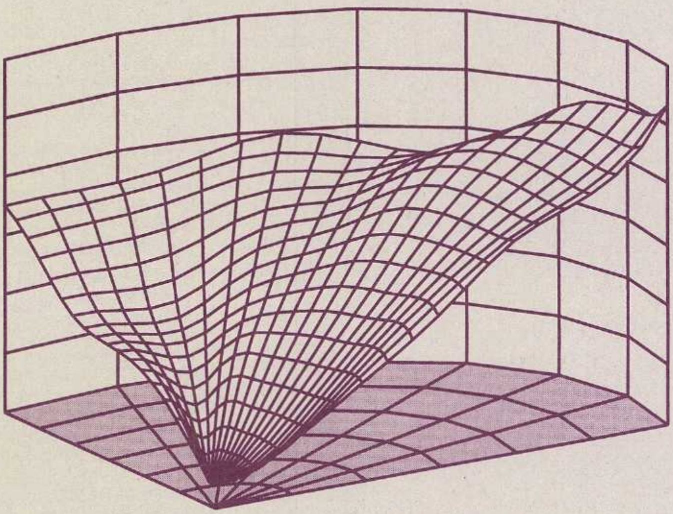
Biomechanical and Ergonomic Considerations on Shoulder Muscle Load

Experimental studies



Gunnar Palmerud

Göteborg 1998

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Experimental Studies

Akademisk avhandling
som för avläggande av doktorsexamen i medicinsk vetenskap vid
Göteborgs Universitet kommer att offentligt försvaras i
Aulan, Centralkliniken, Sahlgrenska Universitetssjukhuset/Östra, Göteborg
kl. 13.00, onsdagen den 27 maj 1998.

av

Gunnar Palmerud

Forskningsingenjör

Avhandlingen baseras på följande arbeten:

I Järvholm U, Palmerud G, Karlsson D, Herberts P & Kadefors R: **Intramuscular Pressure and EMG in Four Shoulder Muscles**. *J Orthop Res*, 9(4), 609-619, 1991.

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Abstract

Palmerud G.: *Biomechanical and Ergonomic Considerations on Shoulder Muscle Load - Experimental Studies*. Department of Orthopaedics, Institute of Surgical Sciences, Göteborg University, Göteborg, Sweden, 1998.

Work-related shoulder disorders represent a significant subset of work-related musculoskeletal disorders. Shoulder tendinitis has a high prevalence in major populations of employees. There is an urgent need for hints and directions on how to evaluate risks for developing chronic work-related shoulder disorders, as well as for methods of assessing hazardous working conditions.

The aims of this work were to assess usability and accuracy in techniques and methods for exploring arm positions and shoulder muscle activity and to investigate the pattern of shoulder muscle interplay regarding voluntary control. The aims were also to elucidate the relations between the external physical demands originating from working tasks and internal load on different structures in the shoulder and to propose a model for estimating the risk of developing acute and chronic work-related musculoskeletal disorders of the shoulder, by using observable data which characterize the exposure.

Electromyography (EMG) and intramuscular pressure (IMP) measurements were used to assess shoulder muscle activity. Bipolar electrodes were used for acquisition of EMG from the superficial muscles and for deep lying muscles indwelling wire electrodes were used. Acquisition of IMP was accomplished by microcapillary infusion technique. A 3D motion analysis system (MacReflex) has been used for control and registration of arm postures.

The results indicate that IMP and EMG measurements produce similar estimates of muscle load and that there were substantial differences in IMP generation between shoulder muscles. Extremely high IMPs were recorded in the supraspinatus muscle. The MacReflex motion analysis system was found to produce position data with an angle error less than 1.8° in biomechanical and ergonomic application. It was demonstrated that the activity of the descending part of the trapezius muscle could be voluntarily reduced to 56% of the initial activity, without altering external conditions, and that the main part of the trapezius load was transferred to the rhomboid major and minor and the transverse part of the trapezius. IMP measurements showed that accumulated muscle fatigue might occur in the infra- and/or supraspinatus, if the upper arm exceeds 30° of arm elevation and that impaired muscle blood flow might occur if the upper arm exceeds 50°. In the validation of the biomechanical shoulder model, it was established that the shoulder muscle forces estimated by the model acceptably agree with the shoulder muscle forces estimated by the EMG measurements, except for the levator scapulae and the supraspinatus, where estimations from EMG measurements were 53% and 207%, respectively, of the activity calculated by the model.

In conclusion: feedback assisted reduction of trapezius activity will affect other shoulder muscles, MacReflex motion analysis system possesses adequate accuracy for biomechanical and ergonomic purposes and IMP recordings from the infra- and supraspinatus indicate the necessity of restricted arm positions in static or repetitive work.

Key words: Biofeedback, biomechanics, electromyography, ergonomics, intramuscular pressure, motion analysis, shoulder model, shoulder muscles, voluntary control.

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ISBN 91-628-2879-7

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Experimental Studies

by

Gunnar Palmerud

Göteborg

1998

Since fitting the task to man,

Is the ergonomist's plan,

Make Changes easy,

So that employers aren't queasy

And your report won't end up in the can.

Kathleen Welsh
Canada

Printed in Sweden by
Chalmers Reproservice
Göteborg 1998

ISBN 91-628-2879-7

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ABBREVIATIONS

3D	three dimensional
A/D	analogue to digital
AEA	arm elevation angle
AEP	arm elevation plane
AMF	Arbetsmarknadsförsäkringar (AMF insurance)
ASS	National Board of Occupational Safety and Health in Sweden
BFI	blood flow impairment
CEN	European Committee for Standardization
DC	direct-current
EMG	Electromyography
FM	frequency modulated
FMA	final muscle activity
IMA	initial muscle activity
IMP	intramuscular pressure
ISA	Arbetarskyddsstyrelsens Informationssystem om Arbetsskador
MBF	muscular blood flow
MCI	microcapillary infusion
MVC	maximal voluntary contraction
MPF	mean power frequency
NIOSH	National Institute for Occupational Safety and Health
NR	non-recovery
RMS	root mean square
RWL	Recommended Weight Limit
VDU	video display unit
WMSD	work-related musculoskeletal disorder

List of Papers

This thesis is based on the following papers, which will be referred to in the text by their Roman numerals:

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CONTENTS

INTRODUCTION

Occupational injuries
In the Swedish system for collecting information on occupational injuries (ISA), all reported accidents leading to at least one day of absence from work after the day of the accident, and all diseases are included. An occupational injury that can be assigned to a certain event or a specific point in time, e.g. in connection with lifting a heavy load, is defined as an occupational accident. It is distinguished from an occupational disease, which is expected to develop over a certain length of time. Occupational diseases affecting body structures of mobility and support, are referred to as work-related musculoskeletal disorders (WMSDs, Kuorinka and Forcier, 1995).

The work injury insurance scheme presents a general description of what is considered an occupational injury. The description includes injuries resulting from accidents or other harmful influences at work. The term "harmful influence at work" refers to factors in the working environment that, with a high degree of probability, can be the cause of the type of injury acquired. The requirement for a high degree of probability was introduced in the beginning of 1993. At the same time, the terms of the work injury insurance became more stringent in an additional aspect. If it is certain that the insured person has suffered an accident or some harmful exposure at work, his or her injury may presumably be the result of a harmful influence, but only if there are stronger grounds for such a presumption than the contrary. The modified definition influenced the outcome of reported injuries, as could be expected (see Figure 1). Before 1993 this rule of evidence was inversely formulated, i.e. the presumption was made if there was no stronger evidence against it.

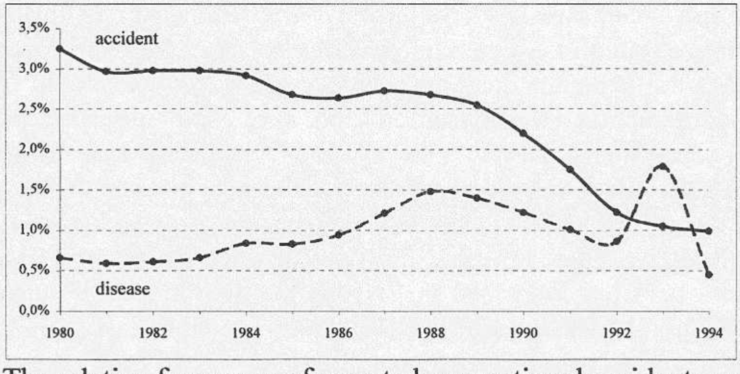

Figure 1. The relative frequency of reported occupational accidents and occupational diseases among Swedish employees during 1980-1994 (Broberg and Eklund, 1996)

The most frequently reported WMSDs are attributed to ergonomic factors, such as monotonous or unusually strenuous movements or working postures, and affect the musculoskeletal system. Several epidemiological studies have shown that there is a maximal prevalence for musculoskeletal disorders for the neck and shoulder in the population between 45-65 years of age, and that women have a higher prevalence compared to men. In the general population, every third woman and every fourth man report pain from the neck and shoulder region (Östergren 1997). WMSDs often result in long rehabilitation periods, which increase in length with age. The mean rehabilitation times for WMSDs are 117 days (Broberg and Eklund, 1996).

The work-related musculoskeletal injuries affect different parts of the body, with back disorders being predominant among musculoskeletal accidents, and neck and shoulder disorders among musculoskeletal diseases. The most common locations of WMSDs are the neck and shoulder. The AMF insurance (the public labor market insurance agency in Sweden) stated in 1996 that the majority of paid worker's compensation involves musculoskeletal disorders, and that neck and shoulder problems have increased during the last decade (Wennström 1996).

Work-related shoulder disorders

Work-related shoulder disorders represent a significant subset of WMSDs. Shoulder tendinitis, like shoulder myalgia, has a high prevalence in major populations of employees. There is an apparent gender-related distribution of such diagnises. Women often suffer from shoulder myalgia and men from subacromial pain (Bergenudd et al. 1988). Biomechanical studies have revealed that there is a difference in muscular control between men and women (Shklar and Dvir 1995). Myalgia frequently occurs in populations working with material handling demanding light to moderate force and in populations with monotonous working postures, especially in combination with mental stress (Veiersted and Westgaard 1994, Veiersted 1995, Holmström et al. 1992). These populations are found in e.g., assembly industry, retail trade and dental service. In branches with a higher demand on muscle force, for example the welding industry, construction, transportation and the food industry, rotator cuff tendinitis is prevalent (Herberts et al. 1984, Hagberg and Wegman 1987, Dimberg 1991). Work-related shoulder disorders do not show any decreasing tendency, in spite of the fact that most of the heavy and demanding working tasks have been eliminated in industry. On the contrary, the prevalence of neck and shoulder pain has increased in Sweden during the last decades (Allander 1974, Hagberg et al. 1992, Ekholm 1995). New groups of workers, performing working tasks characterized by low but sustained shoulder muscle contractions, have been affected. Retirement because of work-related shoulder disorders is frequent.

Contributing factors to WMSDs

Work-related diseases have been characterized as multifactorial by the WHO indicating that a number of factors are involved. Physical, work organizational, psychosocial, individual and sociocultural factors contribute to these diseases. A factor in this context can be defined as "an event, a condition or a characteristic, that plays an essential role in producing an occurrence of a disease" (Wiesel et al., 1985). Evidence for work-relatedness with respect to musculoskeletal disorders is partly based on epidemiological studies, where workplace factors and their relationship to symptoms and physical examination findings are investigated. Work factors include exposure to awkward postures, forceful exertions, repetitive or static exertions and absence of rest (e.g., Putz-Anderson 1988; Kuorinka and Forcier 1995; Kumar 1994). However, a prime cause of musculoskeletal disorders is exposure to load, as concluded in a recent report from NIOSH (National Institute for Occupational Safety and Health, USA, 1997), where it was stated that "a large body of credible epidemiological research exists that shows a consistent relationship between musculoskeletal disorders and certain physical factors, especially at higher exposure levels". It is difficult to find epidemiological evidence for a single factor because many work tasks involve combined exposure to two or more factors. The epidemiological evidence showing an association between highly repetitive work and shoulder musculoskeletal disorders (Bjelle et al., 1981; Vihman et at., 1982; Kilbom, 1994). Repeated or sustained shoulder postures with a shoulder elevation greater than 60° have been related to an increased risk for occurrence of shoulder musculoskeletal disorders (Hagberg and Wegman 1987).

Affected structures

WMSDs affect several structures of the shoulder, particularly the joints, the tendons and the muscles.

Joints

The acromioclavicular joint is the most affected joint of the shoulder (Stenlund, 1992). Occupational disorders of the actual joint of the shoulder, the glenohumeral joint, are rare. The pressure in the subacromial bursa has been thoroughly investigated by Sigholm (1987). He did not find any significant differences in the bursa pressure in a group of subjects with chronic shoulder pain compared to a reference group. The bursa pressure, increasing with increasing humeral torque, could however adversely affect adjacent structures.

Tendons

Shoulder tendinitis is primarily associated with an inflammatory reaction in the long tendons of the infra- and supraspinatus muscles. Hypovascularity of the tendon (Neer 1972, Herberts and Kadefors 1976) and mechanical compression of the tendons in the subacromial outlet (Herberts et al. 1984) are two factors, which may underlie various clinical conditions. Also, sustained high tension and repetitive strain can lead to shoulder tendinitis (Rahne 1994).

Muscles

Work-related muscle pain in the neck and shoulder region are often localized to areas corresponding to the descending part of the trapezius, the levator scapulae, supra- and infraspinatus and the rhomboid muscles. The correlation between chronic myalgia and work place factors is unclear. Several hypotheses have been suggested concerning the etiology. There are some evidence that myalgic muscle differs from normal muscle, for instance concerning impaired microcirculation, damaged mitochondria and a reduced prevalence of high energy phosphates (Bengtsson and Henriksson 1989). Morphological changes (so-called "moth eaten fibres" and "red ragged fibres") were reported by Larsson and co-workers (1988), and larger cross-sectional area and reduced capillary supply in Type 1 fibers by Lindman and co-workers (1992). It has been hypothesized that it is the low threshold motor units that are subject to irreversible degenerative processes, causing myalgic pain (Hägg 1991). It is unclear if there are stereotypic motor unit recruitment and firing patterns in occupational work situations. Observations by Søgaard (1995) in the biceps muscle indicate however that this may be the case. Trapezius muscle tension may also be evoked by experimental stress (e.g., Lundberg et al. 1994). The low threshold units recruited in stressful situations may be the same as those recruited voluntarily (Wærsted et al. 1996).

In the last decades, the multifactoriality of musculoskeletal disorders has become well known and accepted. A cause-and-effect model for musculoskeletal disorders should be based on the hypothesis that there exists an interaction between mechanical and psychological factors at work, as well as outside work (Östergren 1997).

The necessity of prevention

Chronic work-related shoulder disorders induce pain, fatigue, loss of muscle energy and restricted function to the individual. Long term absence from work will furthermore undermine personal economy and occupational competence. Industry and society are heavily affected by the economic burden of sick-listing, medical care and early retirements, which is the case in all industrialized countries. The comprehensive intention for research in the field of WMSDs must be to generate knowledge, which makes it possible to prevent the negative effects for the individual as well as for industry and society at large.

REVIEW OF THE LITERATURE

Ergonomic standards and guidelines

Swedish law and regulations
Regulations, as well as hints and directions for health and safety in working life, originate from the National Board of Occupational Safety and Health in Sweden (ASS). Revised regulations in the field of *working postures and physical loads* were recently referred to different authorities for consideration, both professional and industrial organizations, and in January 1998 they were ratified. The regulations cover work postures and movements, manual handling, monotonous repetitive work, strongly controlled or restricted work, decision latitude and knowledge, abilities and information. The responsibilities for the employer concerning the work environment were settled, as well as the responsibilities for importers, producers, proprietors, etc. and for the employees. In the statute AFS1998:1 published by ASS, it is stipulated that:

The employer should make sure that the employee is informed about:
appropriate working postures and movements
how to use technical aids and devices
the risks involved with inappropriate working postures, movements and manual handling
- early signs on overload of joints and muscles

The hints and directions in the new regulations harmonize well with the state of the art on working postures and movements, manual handling and force exertion, as well as monotonous repetitive or static work, psychosocial and individual aspects.

European standards
In the European Community, laws and regulations are drawn up by Technical Committees and submitted to CEN members for formal approval. If one of these drafts become an European standard, the CEN members are bound to give this European standard the status of national standard, without any alteration. Sweden is a member of CEN.

In the European standard, it is established that the ergonomic principles should be followed and that the term "ergonomics" refers to the multidisciplinary interpretation of the concept. Most of the regulations concerning working postures, working movements and physical loads are found in publications under the "Safety of machinery" directive (European Committee for Standardization, 1992; 1994; 1995). Regarding working postures, recommendations are expressed in general terms such as, "awkward postures leading to body fatigue shall be avoided", "sitting should be preferred to standing", "suitable body postures and appropriate support for the body shall be ensured". Regarding working movements, equally general expressions are used, such as, " the body or parts of the body shall be allowed to move in accordance with their natural paths and rhythms of motion", "repetitive movements that lead to impairment, illness or injury shall be avoided" and regarding physical strength "the demands on physical strength during work shall be kept to an acceptable level". No specific recommendations expressed in entities like joint angle, height, weight, force duration or velocity is provided.

NIOSH guidelines

The most generally adopted recommendations on load exposure are perhaps the NIOSH (National Institute for Occupational Safety and Health) guidelines for manual lifting. The guidelines were published 1981 and revised and expanded 1993 (Waters TR et al., 1993). The NIOSH guidelines focus on manual lifting exposure and the development of lifting-related low back pain (LBP), but they also have a potential to reduce other musculoskeletal disorders or injuries associated with some lifting tasks, such as shoulder or arm pain (Chaffin et al., 1976). The guidelines were developed from biomechanical, work physiological and psychophysical knowledge derived from the scientific literature and expressed as an equation for calculation of a *recommended weight limit* (RWL), where horizontal and vertical displacement, lifting distance, asymmetrical lifting, couplings equipment and lifting frequency are taking into account. The development of the lifting equation was based on three criteria derived from the scientific literature: 1) manual lifting poses a risk of LBP to many workers; 2) LBP is more likely to occur when workers lift loads that exceed their physical capacity; and 3) the physical capacity of workers varies substantially. A committee of experts from the fields of biomechanics, work physiology and psychophysics agreed upon: 1) a maximum disc compression force of 3.4 kN, 2) a maximum energy expenditure of 2.2-4.7 kcal/min. depending on duration and location of lifting and 3) a maximal acceptable weight, implying that 75% of female workers and 99% of male workers perceive the weight acceptable.

The lifting equation is a tool for assessing risk of low back pain or injury and is limited to conditions it was designed for. This implies that the equation is applicable only to working situations were non-lifting activities or environmental factors are negligible. One-handed lifting or lifting while seated or kneeling is not included in the range of application.

Electromyography

Techniques

Electromyographic studies of the human shoulder have been carried out by several investigators, employing surface as well as intramuscular techniques. The choice of electrodes generally depends on a) the requirements of selectivity, and b) the accessibility of the muscle under study.

Intramuscular fine wire electrodes, insulated except for at the tip, have been used particularly in functional anatomy studies (Basmajian 1985). The wire diameter is usually 25-75µm and the wire material of platinum, stainless steel, copper, nickel-chromium etc. has been used (Németh et al., 1990, Järvholm et al., 1989, Kadaba et al., 1992, Sporrong et al., 1995, Krivickas et al., 1996;). It is inserted by means of a cannula, which is withdrawn after insertion, allowing recording of EMG during body movements. Needle electrodes have also been used for acquisition of intramuscular EMG (Søgaard, 1994).

Surface electrodes affixed on the skin over the muscle under investigation have been widely used. For optimal EMG acquisition, the bipolar surface electrodes are orientated along the direction of the muscle fibres, with an interelectrode distance of 10-30 mm (De Luca, 1992; Örtengren, 1996). A decrease of the interelectrode distance will result in increasing the bandwidth of the detected signal and the spatial resolution, while the amplitude of the signal will decrease (De Luca, 1992). The innervation zone impairs signal amplitude and frequency properties and should be avoided (Lindström, 1974). The dominating electrode material is silver/silver-chloride. The electro-chemical potential generated between the skin and the electrode is low for this material compared to other materials. Use of electrode gel and skin preparation are parts of a routine procedure, to reduce the cross-over impedance, which should be no more than a few kilo-ohms (Örtengren, 1996). Generally the detection surface of the electrode is approximately 20 mm^2.

EMG to force relationship

One major reason for the importance of EMG recordings in the field of ergonomics and biomechanics is the correlation between the EMG activity and the force output. While the characteristics of the EMG-force relationship are discussed, a monotonous increasing correlation is generally accepted for a constant muscle length. In a critical review of the literature concerning the EMG-force relationship Perry and Bekay (1981) concluded that "during isometric effort a linear relationship exists between an appropriately quantified EMG measure and a registered force, at least near the middle of the operating force range, and for some electrode configurations". Körner (1984) studied intramuscular EMG and force output of the biceps brachii during static isometric contraction of 23 subjects. He found that there was an almost linear relationship between EMG and force output with a high correlation coefficient. In a study concerning EMG-force relationship in different muscles Lawrence and De Luca (1983) found that the relationship was primarily determined by the muscle under investigation but appeared to be independent of the force rate of contraction from 10 to 40%.

Electromyography in the study of the shoulder

In the last decades, the awareness of the increasing occurrence of musculoskeletal disorders of the shoulder has lead to several studies and a number of papers and theses concerning the muscles of the shoulder have been published.

Six shoulder muscles were investigated by Sigholm et al. (1984). EMG from the trapezius descendens, all three parts of the deltoid, the infra- and supraspinatus, were picked up by means of single fine wires. The recorded data on shoulder muscle activity was later used for a modelling of the shoulder. Järvholm (1990) investigated shoulder muscle activity with recordings of intramuscular electromyography and intramuscular pressure, and found that they were highly correlated during isometric contractions.

The importance of electrode placement in EMG studies was showed by Mathiassen (1993). He investigated the significance of changes in external load and arm position with four surface electrodes affixed on the area covered by the descending part of the trapezius muscle. He found that the relationship between glenohumeral torque and EMG depends on the location of the electrodes, and concluded, that signals from one pair of electrodes may not be representative of the whole upper trapezius. In the same study he also noted a coactivation in the trapezius muscle on the "non-active" body side – contralateral coactivation.

"Zero crossing", a new method for detection of electromyographic signs of muscle fatigue was developed by Hägg (1991), and applied to the trapezius and the infraspinatus muscles. He showed that zero crossing was comparable with other methods for estimating local muscle fatigue. The method was applied in a crossectional and a longitudinal study to test a hypothesis on EMG signs of muscle fatigue during work and the risk for developing WMSDs in the neck and shoulder. The results indicated that the method had a potential application as a diagnostic tool, but not as a predictive tool. Also Weiersted (1995) estimated the risk of developing trapezius myalgia from the activity pattern of trapezius EMG. In a prospective study on female packers, he found that future patients showed an activity with low frequency of EMG gaps during work and a higher static level in stressful situations compared to the non-patients. In the work of

Sundelin (1992) the effects of repetitive working cycles and pauses on the some shoulder muscles were evaluated with electromyography.

To investigate the influence of repetitive work on the descending part of the trapezius muscle, the middle part of the deltoid muscle and the infraspinatus muscle, Søgaard (1994) conducted an EMG study on cleaners. Conclusions on different methods of floor cleaning was presented.

Nieminen (1994) conducted several studies on shoulder muscle load and developed new signal processing and modelling methods for the EMG signal, especially methods for analysis of the temporal pattern of the EMG. The validity and reliability of the EMG mean power frequency as an estimator of local muscle fatigue in the trapezius muscle, was systematically studied by Öberg (1992). He found several confounding factors to MPF changes, e.g. glenohumeral elevation, shoulder torque and external hand load, and no correlation between MPF and subjective muscle fatigue at low load level. Another methodological question was addressed by Bao (1995). He investigated the consequences of different electromyographic normalization procedures and found that different upper trapezius EMG normalization procedures might result in a substantial variation in load estimations.

The additive effects of static and intermittent hand activity, as well as hand activities with high demands on precision has an additive effect to the postural shoulder muscle activity. This was demonstrated by Sporrong (1997) in several EMG studies on the shoulder. Also mental stress induces muscular tension. The effect of mental stress as well as physical load, separate and in combination, on muscular tension as reflected in the EMG activity, was examined in a study by Lundberg et al. (1994). Also Wærsted (1997) addressed the influence of psychological factors on trapezius muscle activity. In experimental studies he examined the pattern of attention-related trapezius surface EMG during VDU work. He found that psychological factors may give rise to sustained activity in low-threshold motor units.

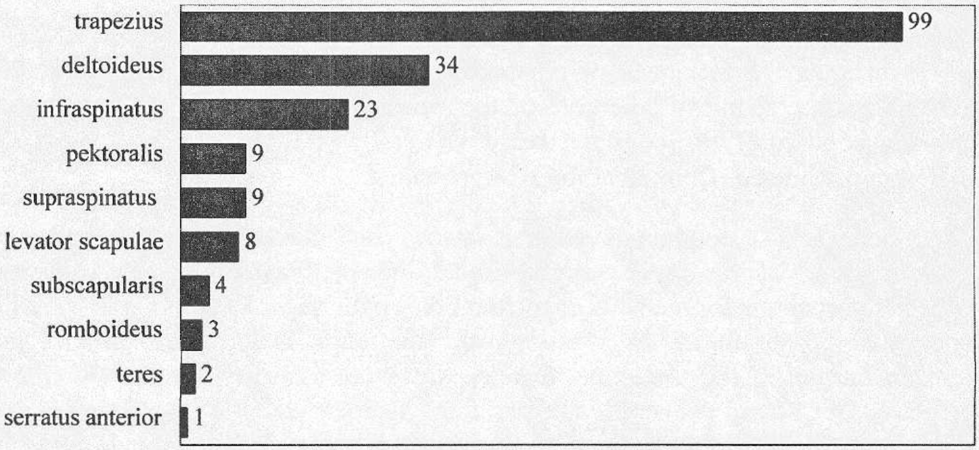

Figure 2. The distribution of papers published 1990-1996 on different shoulder muscles. Source: Ergonomics Abstracts 1996.

In fact, the vast majority of papers on shoulder muscles published the 1990s deal with the trapezius muscle, see Figure 2. The main explanation for the large number of papers concerning the trapezius muscle is the fact that shoulder pain and disorders are associated with this muscle, and perhaps that there exists a relationship between mental stress, trapezius muscle activity and chronic pain. The number of papers on the rotator cuff muscles are remarkably few, considering their intriguing and complex function and the frequency of occurring pain and dysfunction from the rotator cuff tendons. A possible explanation might be found in methodological difficulties in investigating deeply located muscles.

Intramuscular pressure

Tissue pressure has been monitored in physiological research since Leanderer performed interstitial tissue pressure measurement with a needle manometer in 1884. Guyton described the total tissue pressure as a combination of interstitial fluid pressure and solid tissue pressure (Guyton et al. 1971).

Methods for tissue pressure recording

Different techniques have been developed for tissue pressure measurements. They can be divided in two categories, extracorporal transducer techniques and intracorporal transducer techniques, referring to the location of the pressure sensitive device. The methods can also be divided into infusion and noninfusion techniques (see Table 1).

Table 1. A summary of different tissue pressure measurement techniques. Some of the techniques are only of historical value.

Tissue pressure measurement techniques	Extracorporal methods	Intracorporal methods
Infusion technique	MCI technique Needle technique	Microsemiconductor transducer techniques
Noninfusion technique	Balloon technique Needle manometer technique Wick catheter method Slit catheter method	Microsemiconductor transducer techniques Fiberoptic transducer technique

Intramuscular pressure recordings

Diagnosis of compartment syndrome is an example of a clinical application of tissue pressure registration. Compartment syndrome is a serious condition arising from elevated tissue pressure in low compliance compartments compromising the circulation (Matsen, 1975). If unattended, it might lead to necrosis in the affected area. Tissue pressure measurements in muscles have been applied in the fields of biomechanics and ergonomics (Körner et al., 1984; Sejerstedt at al., 1984; Sjøgaard et al., 1986; Järvholm et al., 1988a). Intramuscular pressure measurement is a method for assessing information on muscle engagement and muscle physiology during working conditions. Several of the methods listed in Table 1 have been thoroughly evaluated and accepted as suitable for intramuscular pressure measurements. Disadvantages with these methods may include poor dynamic properties, easily affected by occlusions, trauma for the subject, easy breakable and extreme cost.

Microcapillary infusion (MCI) technique

The MCI technique has been evaluated by Styf and Körner and found suitable for intramuscular pressure measurements during static as well as dynamic muscle contractions (Styf and Körner, 1986). They investigated the dynamic properties and estimated the rise time to be 35 – 70 ms. If a sinusoidal time function of the muscle pressure is assumed, this would indicate an upper frequency of 3-6 Hz.

Fluid infusion is a method to prevent occlusion of the catheter openings. However infusion introduces a volume load of the tissue, which might give false elevated pressure levels. Errors due to the fluid injected into the muscle have been investigated. A substantial amount of fluid could be absorbed by the muscle tissue at rest, without influencing the recorded muscle pressure (Guyton, 1965). Järvholm and co-workers (1988a) found that the intramuscular pressure in the supraspinatus muscle increased by 0.5 mmHg per ml/h of infusion rate at rest and 1 mmHg per ml/h during contraction. Since the MCI technique is a non-constant infusion method, where the infusion rate is determined by the pressure difference across the microcapillary, infusion rate will adapt to the intramuscular pressure level, i.e. when the contrapressure in the muscle is high due to muscle contraction, the infusion rate decreases, and when it is low due to muscle inactivity, it increases. This is in agreement with the fact that a resting muscle is more tolerant to volume load than a contracting muscle. However the muscle contra pressure must not exceed the driving pressure, because this will cause a retroflux and most certainly occlude the catheter. A methodological error introduced by an infusion rate less than 6.3 ml/h is regarded as acceptable in this context.

The occasionally occurring occlusion of the catheter is a generally recognized methodological disadvantage in fluid-filled systems with extracorporal transducers. The MCI technique is nevertheless considered the most reliable system of this type, partly because of the continuous infusion of saline and partly because of the design of the Myopress catheter tip (Styf et al. 1989).

However, if catheter occlusion occurs, the detection of this condition is trivial, provided a chart recorder is used for supervision. The character of the signal from an occluded catheter with the MCI technique differs clearly from a normal IMP signal by the way the monotonous buildup of the pressure corresponds to the current infusion rate and by the way the pressure asymptotically approaches the driving pressure. If only partial catheter occlusion is at hand, depression of the frequency response of the measurement system will occur with all fluid-filled systems, implying low sensitivity to rapid changes in pressure. With a careful supervision of the IMP signal characteristics, this artifact is also easily detectable.

Intramuscular pressure variations

Most studies on intramuscular pressure have revealed a considerable variation of the IMP between individuals (Sadamoto et al., 1983; Körner et al., 1984; Sejersted et al., 1984; Sjøgaard et al., 1986). The variation is due to methodological as well as physiological reasons. Järvholm and co-workers (1988a) found that the interindividual variation of IMP during contraction was large and suggested that this could either be due to methodological reasons, for instance variation in the catheter location, different relative lengths of the muscle, different or manipulated muscle activation patterns and poor dynamic properties due to occlusion of the catheter tip or a reflection of genuine individual differences.

The sensitivity of the IMP to variation in catheter location depends on the IMP gradient of the muscle, e.g. the variation of the IMP level across the muscle. The

IMP gradient varies substantially, e.g. a bulky muscle surrounded by high compliance structures versus a thin muscle surrounded by low compliance structures such as bone and fascia (Sejersted et al., 1984)

Biomechanics

Biomechanical analysis of the shoulder is a challenge, because of their complexity. The total shoulder torque can be estimated if the magnitude and direction of the force and the lever arm are known (and eventually also the weight and the center of gravity of the arm). However, the internal distribution of forces on different structures of the shoulder remains unknown. To make it possible to prevent musculoskeletal disorders, it is crucial to identify local overloading of muscles. Since there exist no direct methods for assessing single shoulder muscle forces in vivo, there is a need for a comprehensive shoulder model, with the ability to predict the internal force distribution from external parameters. To develop a shoulder model it is necessary to have sufficient data concerning the shoulder anatomy, the relative movements of the shoulder bones ("the shoulder rhythm") and the muscular recruitment pattern. Data on the shoulder rhythm as well as on the muscular recruitment pattern are however very limited, which hampers the development of a shoulder model. In addition to these difficulties, the interindividual differences are large.

A three-dimensional biomechanical shoulder model was developed by Högfors and colleagues in Göteborg (Högfors et al., 1987; Högfors et al., 1991; Karlsson and Peterson, 1992; Högfors et al., 1995). It comprises 38 muscles or muscle parts, four joints and one ligament of the human shoulder. The model is based upon geometric data (the shoulder anatomy), kinematic data (the relative movements of the shoulder bones, called "the shoulder rhythm") and anthropometric data (the length, weight and center of gravity for the body segments involved). It allows for arbitrary body postures. The shoulder model is based on assumptions of the muscle forces being strictly correlated to the physiological cross-sectional areas of the muscles and a criterion of optimization. The shoulder model is directed towards the analysis of static working situations where the load is low to moderate. It can estimate the force contribution from each muscle acting across the shoulder joint, to the external moment of shoulder torque.

To validate the model, strength profiles of the shoulder joint were measured by a force device and compared to the strength profiles calculated by the shoulder model (Maksous, 1996).

Other research groups have also presented three-dimensional shoulder models, for instance the Delft group in the Netherlands, who has presented a dynamic model for the whole shoulder (Pronk, 1991, van der Helm, 1991).

Laursen (1996) used a different approach to solve the problem with the optimization criterion in developing a model for predicting shoulder muscle forces. An EMG based shoulder model was developed from the results of experimental studies. The EMG-force relationship between force exerted by the hand and EMG activity of 13 shoulder muscles was established for six female subjects. A similar muscle activity pattern was found for all the subjects, which justified the use of a standard muscle activity pattern for all individuals. The model was verified by comparing joint moments calculated from EMG with moments from the external force. When the model was restricted to low muscle forces (≤20% of MVC) a correlation coefficient of 0.65 - 0.95 was found for shoulder abduction-adduction moments and 0.70 - 0.93 for shoulder flexion-extension moments.

Aware of the time-dependent muscle rotation phenomenon occurring in the muscle activity pattern of the shoulder, Nieminen (1994) developed a model including a load-sharing principle. This implies that the time elapsed from the start of the activity decreases the allowable muscle stress levels on the basis of the stress-endurance curves. The stress endurance curves of the muscles varies according to the amount of slow-twitch fibers in the muscle. He also introduced the shoulder stiffness effect of cocontraction. This effect may occur for instance from working tasks with high demands on precision.

Aims of the Thesis

- to elucidate the relations between the external physical demands originating from working tasks and internal load on different structures in the shoulder (I, V)
- to assess usability and accuracy in techniques and methods for exploring shoulder movement and muscle activity (I, II, VI)
- to investigate the pattern of shoulder muscle interplay regarding voluntary control, its interindividual variation and reproducibility (III, IV)
- to propose a model for estimating the risk of developing acute and chronic work-related musculoskeletal disorders of the shoulder, by using observable data which characterize the exposure (V)

Material and Methods

Subjects

The subjects in all the studies were healthy young people with no previous history of shoulder pain. In all studies, the composition of subjects was either males or females, except for the study on the supraspinatus muscle in Paper I. The mean group ages were about 30 years of age and the oldest participant was 45. In Paper I, a total of 20 male subjects participated. In Paper II, there was only one male subject and in Paper VI, the subjects were identical to those participating in Papers III and IV (see table 2). The subjects were mainly students from the universities in Göteborg or medical staff from the hospitals. In Paper V, the subjects were athletes recruited from a sports team. An experimental session lasted approximately two hours. The subjects were economically compensated for participating.

Table 2. Summary of data for the groups of subjects participating.

Group of subjects	No. of subjects	Gender Male/female	Age (year)	Weight (kg)	Length (cm)	Dominant hand right/left	Occurring in paper
A	22	19/3	24	73	180	21/0	I
B	1	1/0	42	88	189	1/0	II
C	6	0/6	30	64	169	6/0	III+VI
D	11	0/11	33	64	167	11/0	IV+VI
E	11	11/0	26	80	182	10/1	V

Before giving their consent to participate in a study, the subjects were thoroughly informed both verbally and in writing about the duration of the experiment, the number of intramuscular electrodes or catheters, the pain and discomfort expected and their right to terminate their participation at any time during the experiment, without giving any motivation. All protocols were submitted to and approved by the Research Ethics Committee at the Medical Faculty, University of Göteborg before the studies were executed.

Electromyography

Surface electromyography

Bipolar surface electrodes were used for acquisition of EMG activity from the superficial muscles. A commercially available, pregelled and self-adhesive electrode was chosen (Blue Sensor N disposable electrodes from Medicotest A/S, Denmark), with a pick-up area of approximately 20 mm^2. The skin preparation of the electrode site included cleaning and degreasing with Klorhexidin (0.5mg/ml) and shaving if necessary. The electrodes were attached to the skin, edge to edge along the direction of the muscle fibers, implying an interelectrode distance of approximately 20 mm.

Electrode placement was done in accordance with the recommendation found in the works of Basmajian (Basmajian et al. 1983, Basmajian et al. 1985). For the muscles not described by Basmajian the pick-up area was chosen above the bulky central part of the muscle. The pick-up area of the *descending part of the trapezius* was chosen to midway between the acromial angle and the 7th cervical vertebra. This is in accordance with the Basmajian recommendations.

The elongated oval area below the lateral end of the clavicle was chosen for EMG registration of the *anterior part of the deltoid muscle.* For the *medial part of the deltoid muscle*, the elongated oval area on the midline of the lateral surface of the arm and approximately ¼ of the distance from the acromion to the elbow, distal to the lateral margin of the acromion, was chosen. In Paper IV, EMG registration was made from *the transverse part of the trapezius.* This segment of the trapezius muscle was defined as the part of the muscle, which originates from the spinous processes of the first to fourth thoracic vertebrae and attaches to the spine of the scapula. This definition follows the representation of force vectors in the shoulder model (Högfors et al. 1987). The electrodes were attached to the skin over the trapezius muscle, halfway on the horizontal line between the midpoint of the spine of the scapula and the vertebral column.

The electrode set for *the middle part of the serratus anterior muscle* was found on the midaxillary line over the fourth rib from the top. Also, the definition of the participation of the serratus anterior follows the vector representation of the muscles in the shoulder model.

The reference electrode was an Ag/AgCl surface electrode, taped to the skin above the processus prominens (C7).

Intramuscular electromyography

Acquisition of EMG activity from the deep lying muscles, for instance the *supraspinatus* and the *infraspinatus* muscles were accomplished by indwelling single wire electrodes. (Stabilohm 110, with a diameter of 70 μm, polyurethane isolated nickel chromium alloy; Johnsson Matthey Metals, London, UK). Different electrode diameters have been used for intramuscular EMG recordings. A diameter of 25-50 μm is frequently used, but our experience is that this wire can easily become entangled. The 70 μm wire is easier to handle, especially when surgical gloves are used, and also easier to see. The tip of the wire was deinsulated and then formed like a hook. The wire electrode was introduced to the proper position in the muscle by means of a 22-gauge hypodermic needle. The needle was withdrawn, leaving the electrode in place.

To access EMG from and the *levator scapulae* muscle, fine wire electrodes must be employed, since it is covered by the trapezius muscle. The electrode was introduced approximately 2 cm mediallyand ½ cm cranial to the superior angle of the scapula, penetrating the trapezius muscle. Also, the rhomboid muscles are only accessible through the trapezius muscle. The *rhomboid major* muscle was reached from a position approximately 1 cm medial of the medial border of the scapula and midway between the base of the spine of the scapula and the inferior angle of the scapula. EMG from the *rhomboid minor* muscle was picked up from a point approximately 1 cm medial of the medial border of the scapula, close to the base of the spine of the scapula.

The correct locations of surface electrodes, as well as indwelling electrodes, were verified by adequate muscle provocations like shoulder elevation for the upper trapezius, arm flexion for the anterior part of the deltoid and external upper arm rotation for the infraspinatus.

Intramuscular pressure

Intramuscular pressure was measured with the microcapillary infusion (MCI) technique as described by Styf and Körner (1986). The MCI technique is an extracorporal method, with a non-constant infusion rate. The intramuscular pressure is transferred to the pressure transducer through a fluid filled system (0.9% solution of NaCl). The infusion rate is depending on the resistance of the microcapillary, the driving pressure in the pressurized reservoir and the counter pressure in the muscle. A nominal infusion rate of 1.5 ml/h was used, following a driving pressure of 150 mmHg (20 kPa). In cases where the intramuscular pressure exceeds this pressure, the driving pressure was elevated to 300 mmHg (40 kPa), to avoid a reflux and most certainly occlusion of the catheter.

Pressure catheter

A teflon catheter with a diameter of 1.05 mm was used for the intramuscular pressure recording (Myopress®, Atos Medical, Hörby, Sweden). This catheter is provided with 4 side holes close to the open tip, to increase the contact area between the fluid of the system and the interstitial fluid. The saline filled catheter was introduced into the center of the muscle by means of an intravenous infusion cannula (Vasculon®, Viggo, Helsingborg, Sweden). The cannula comprises an outer plastic tube and an inner steel needle. After local anesthesia of the skin and the subcutis, the intravenous infusion cannula was inserted as parallel to the muscle fibers as possible. The final millimeters of penetration were performed with the sharp steel needle retracted, to minimize tissue trauma. The catheter then replaced the needle, and finally the plastic tube was removed.

Pressure transducer

The pressure transducer used in Paper I (Bentley Trantec, Irvine, Valifornia, U.S.A.) had a displacement of $0.4 \cdot 10^{-3}$ $mm^3/mmHg$ ($3.0 \cdot 10^{-3}$ mm^3/kPa). A disposable pressure dome was screwed on to the pressure transducer, leaving the flexible membrane of the dome in close contact with the membrane of the transducer. The pressurized reservoir and the pressure catheter were connected to the dome by low compliance tubes.

In Paper V a disposable pressure transducer for invasive pressure measurements was used (System DPT-6000, Peter von Berg GmbH, München, Germany). According to the technical specification from the manufacturer the accuracy of the transducer was ±1.5 mmHg in the operating range (–50 to +300 mmHg). The transducer possesses low temporal and thermal drift (±1mmHg/8 hours and ±0.2 mmHg/°C) and good dynamic properties (800Hz natural frequency).

Before the start of each experiment, the pressure transducer was zeroed to the ambient atmospheric pressure and vertical adjusted to the same level as the catheter tip to avoid a systematic hydrostatic pressure bias. The horizontal deviation of the catheter tip during humeral elevation amount to 50 mm, implying a hydrostatic pressure bias of approximately 4 mmHg. This deviation was not compensated for.

Body posture registration

Two methods have been used to control and/or record the arm postures during EMG and/or IMP registration. A *guiding frame* was used in Paper I for positioning of the upper arm and to control for discrepancies during the registration. The guiding frame could be adjusted for different elevation angles (vertical angles) as well as different elevation planes (horizontal angles) The guiding frame was equipped with two indicators, which the shoulder joint and the elbow were oriented to. The length of the guiding frame was adjustable to fit different arm lengths. The elbow angle was adjusted with an ordinary protractor.

Position registration of the upper extremity in Paper V, was accomplished by utilizing a non-contact three-dimensional *motion analysis system* (MacReflex®, Qualisys AB, Partille, Sweden). The basic principle of the system is to record the positions of a number of well-defined points in space. The points to be measured are marked with reflective markers. The system comprises of cameras (version NP-1) in combination with infrared flash lights, video processors (version VP-II), and a Macintosh computer (see Figure 7); the monitors are practical but not necessary. In this study, two cameras were used, but the system is expandable up to seven cameras. MacReflex 3D Software for 50 Hz was used for coordinate calculations and a calibration fixture (type CAB-800) for establishing a laboratory coordinate system.

The torso, the upper arm and the forearm are considered rigid body segments and three spherical light reflective markers with a diameter of 20 mm were attached to each body segment using an exoskeleton. The exoskeleton consisted of two cuffs and a cuirass, each part carrying three markers.

A reference position was needed for calibrating the position data, recorded with the motion analysis system. 90° of forward flexion with a straight arm was chosen for a reference position. The position of the arm was first carefully adjusted, using a protractor, a water-level and a spirit bubble goniometer and then recorded by the motion analysis system. All registrations were then referred to this position.

Statistics

The large amount of variation in biological material necessitates the use of descriptive statistics for organization and presentation of data as well as statistical inference for drawing reasonable conclusions, especially in studies where the number of subjects is small. Different statistical methods have been used in the different papers. For statistical evaluation of the mean values in Paper I (series B), Paper III and Paper IV, a two-sided t-test was used. A probability level of 5 % was used, but in a few cases, other probability levels were used. When more than two groups were compared, the significant level was corrected with Bonferoni inequality. In Paper V a general linear model was applied to estimate the distribution of the total variance and the significance of the independent factors. The model was applied to each combination of hand load and shoulder muscle (4 combinations). In describing the variance of the means, standard deviation (SD), as well as standard error of the means (SEM) were used.

Summary of Papers

Paper I

Aims

- to simultaneous evaluate IMP and EMG in the trapezius, deltoid, infra- and supraspinatus muscles in standardized arm positions and with various hand loads

- to compare IMP and EMG as descriptors of muscle load during isometric conditions

Material and Methods

In this study a total of 25 subjects participated in the IMP measurements. IMP was measured in four shoulder muscles: the supraspinatus, the infraspinatus, the descending part of the trapezius and the anterior part of the deltoid. Seven subjects participated in the supraspinatus measurements, three females and four males, with a mean age of 25 years (20-36 years). In the infraspinatus group there were eight male subjects with a mean age of 23 years (20-25 years). In the trapezius and deltoid groups there were six male subjects in each, with mean ages of 24 years (21-26 years) and 24 years (23-26 years) respectively.

IMP was measured with a microcapillary infusion technique (MCI), which is a non-constant infusion technique, with a flow rate depending on the pressure difference between the driving pressure in the saline reservoir and the counter pressure in the muscle. The driving pressure was set to 150 mmHg or 300 mmHg depending on the IMP. The driving pressure always had to be higher than the intramuscular counter pressure in order to avoid a reflux in the catheter. The microcapillary device was calibrated to give the flow rate of 1,5ml/h for a pressure gradient of 150 mmHg or 3,0 ml/h for a pressure gradient of 300 mmHg. Before the start of each experiment, the external pressure transducer was adjusted to the same horizontal level as the catheter tip in order to avoid a hydrostatic pressure bias.

The pressure-recording catheter was introduced into the muscle by means of a intravenous cannula.

The EMG activity was pick-up by means of bipolar fine wire electrodes. The electrodes were introduced through the same cannula as the pressure-recording catheter. This made it possible to record the EMG activity at the same location in the muscle as the IMP.

The external force output was registered using a force transducer connected between a fixed point and the appropriate location on the subject.

Experimental Procedure

Two different experimental series were conducted. In the first experiment, EMG and IMP in four shoulder muscles and external force was recorded during isometric contractions. Arm positions and force measurement locations were chosen depending on the muscle under study. The supraspinatus measurements were performed during arm elevation, with a straight arm and 45° of shoulder abduction in a vertical plane 45° to the frontal plane. The force transducer was attached to a strap around the elbow. The infraspinatus measurements were performed during external rotation of the upper arm, with the upper arm in a vertical position close to the body and with the elbow flexed 90°. The force transducer was attached to a strap around the wrist. The measurements on the anterior part of the deltoid muscle were performed during arm elevation, with a flexed elbow and 45° of shoulder flexion. The force transducer was attached to a strap around the elbow. The measurements on the descending part of the trapezius muscle were performed during shoulder elevation, with the arm hanging along the body. The force transducer was connected to a strap over the acromion.

All measurements were performed with the subject sitting in a chair and the recorded external force was displayed to the subject, to make it possible for an equi-incremental force generation. Each experimental session was completed with a MVC in the same position as described above and IMP was recorded for each of the four muscles.

In the second experiment, IMP and EMG were registered in different arm positions with and without hand load. Ten standard arm positions were chosen: 0°, 30°, 60°, 90° and 135° of arm elevation in the vertical plane 45° to the frontal plane and 0°, 30°, 60°, 90° and 135° of arm elevation in the vertical plane parallel to the sagittal plane. The initial five arm positions were performed with a straight elbow and the last five with a flexed elbow. In all arm positions, the hand was loaded with 0, 1 and 2 kg. To assist the subject to achieve and maintain the right arm positions, a specially designed guide frame was used. Each arm position was maintained for 10-45 seconds; in between, the arm was relaxed on the lap.

Results

In the first experiment, an almost linear relationship found between the external force and the recorded IMP in all four muscles. Also, between the external force and the EMG recording there was an almost linear relationship. The individual correlation coefficients for the linear regressions were calculated for the force versus IMP or EMG (see Table 3).

Table 3. Mean correlation coefficients of the linear regressions of force versus IMP or EMG for four muscles. SD within parentheses.

Muscle	IMP vs. external force	EMG vs. External force
Supraspinatus	0,95 (0,02)	0,97 (0,02)
Infraspinatus	0,98 (0,01)	0,93 (0,11)
Trapezius descendens	0,95 (0,03)	0,95 (0,03)
Deltoid anterior	0,90 (0,12)	0,90 (0,08)

The mean of the IMP at MVC differed substantially between the four shoulder muscles. The supraspinatus muscle generated the highest IMP of 524 mmHg, followed by the infraspinatus muscle with an IMP of 439 mmHg. The descending part of the trapezius muscle and the anterior part of the deltoid muscle generated IMPs of 86 mmHg and 146 mmHg, respectively.

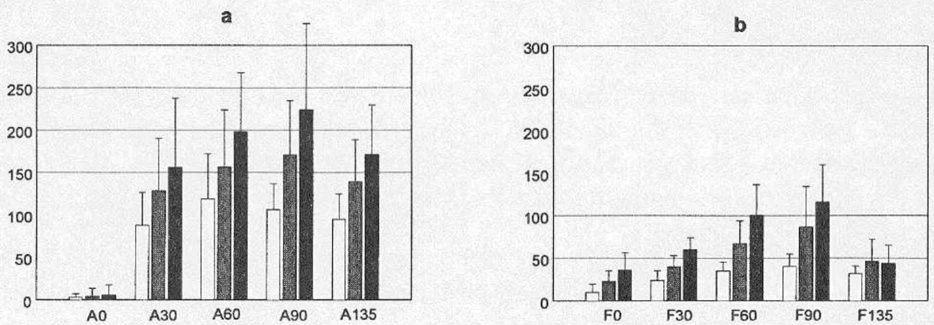

Figure 3. IMP in the *supraspinatus* muscle at a) shoulder abduction and b) shoulder flexion. Open columns, gray columns and filled columns indicate a hand load of 0 kg, 1 kg and 2 kg, respectively. The error bars indicate 1 SD.

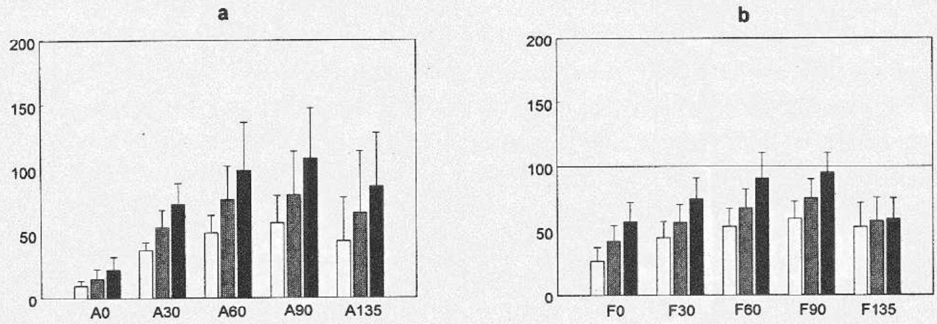

Figure 4. IMP in the *infraspinatus* muscle at a) shoulder abduction and b) shoulder flexion. Open columns, gray columns and filled columns indicate a hand load of 0 kg, 1 kg and 2 kg, respectively. The error bars indicate 1 SD.

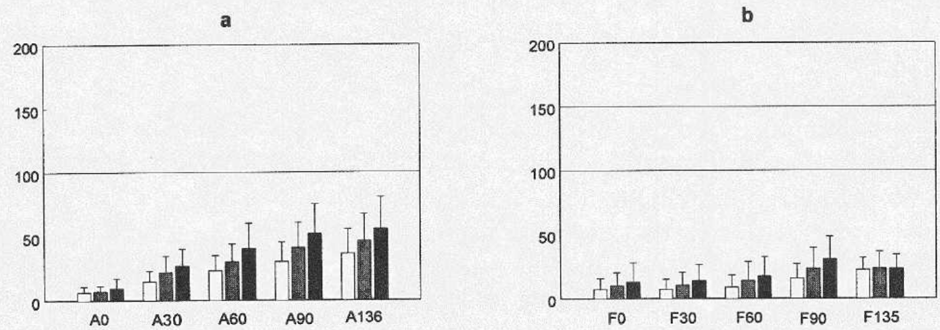

Figure 5. IMP in the *deltoid anterior* muscle at a) shoulder abduction and b) shoulder flexion. Open columns, gray columns and filled columns indicate a hand load of 0 kg, 1 kg and 2 kg, respectively. The error bars indicate 1 SD.

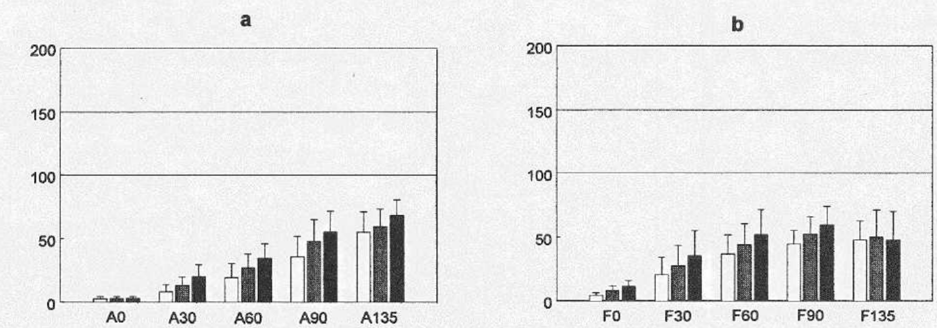

Figure 6. IMP in the *descending part of the trapezius* muscle at a) shoulder abduction and b) shoulder flexion. Open columns, gray columns and filled columns indicate a hand load of 0 kg, 1 kg and 2 kg, respectively. The error bars indicate 1 SD.

In the second experiment, the IMP in the four shoulder muscles was generally higher in abduction compared to flexion (see Figures 3-6). The supraspinatus and, to a certain extent, the infraspinatus showed high IMP (> 50 mmHg) in all positions except at 0° of shoulder abduction and for different hand loads. In the descending part of the trapezius and the anterior part of the deltoid muscles, the IMP seldom exceeded 50 mmHg in any of the arm positions, irrespective of the hand load. There was a strong correlation between normalized EMG and recorded IMP in all muscles and all arm positions, except at 135° of arm elevation. A linear regression analysis of the mean IMP versus the mean EMG in all test situations (135° of arm elevation excluded) and for all four muscles showed a correlation coefficient of 0,96.

The correlation between the recorded EMG from the descending part of the trapezius and the supraspinatus IMP was investigated in shoulder abduction and flexion. There was a fairly good correlation between trapezius EMG and supraspinatus IMP in shoulder abduction, but this was not found in shoulder flexion.

The correlation between the IMP or EMG in the four shoulder muscles and the calculated shoulder torque in abduction and flexion was also investigated for 30°, 60° and 90° of arm elevation. For each separate arm position, there was an almost linear relationship between the mean calculated shoulder torque and IMP or EMG. For the supra- and infraspinatus, there was a strong correlation for all abducted arm positions, whereas the correlation for the flexed arm positions were dependent on the position. The correlation between the mean calculated shoulder torque and IMP or EMG for the deltoid muscle was dependent on arm positions for both elevation planes.

Conclusions

In this study it has been shown that IMP and EMG gave a similar description of the muscle load during isometric contractions, as well as between different arm positions. Substantial differences between the four shoulder muscles were found regarding the IMP generation. High IMP was found in the supra- and infraspinatus, especially in shoulder abduction, as compared to the trapezius and deltoid muscles. This finding supports the suggested theory that bulky muscles within a low compliance compartment will generate higher IMPs during contraction compared to a superficial and flat muscle. This indicates that prolonged work with elevated arms will impede muscle blood flow in the stabilizing muscles of the rotator cuff, and might explain the findings of localized muscle fatigue in EMG from these muscles.

Paper II

This paper is described more in detail to increase comprehension to a wider group of readers.

Aims

- to investigate the sources and amplitudes of error for an optoelectric motion analysis system, when used for assessment of upper extremity orientation in biomechanical and ergonomic applications

Special focus was placed on:

- the limited geometrical precision with which the markers were positioned in the calibration fixture and on the measured objects
- the limited capability to align the calibration fixture and the measured objects to the intended coordinate system
- the limited precision of the protractors, with which angular measurements were made

Material and Methods

The motion analysis system used in this study, called MacReflex, is manufactured by Qualisys AB, Partille, Sweden. It provides non-contact three-dimensional motion measurements. The basic principle of the system is to record the positions of a number of well-defined points in space. The points to be measured are marked with reflective markers. The system comprises of cameras in combination with infrared flash lights, video processors and a Macintosh computer (see Figure 7); the monitors are practical but not necessary. In this study, two cameras were used, but the system is expandable up to seven cameras. The video signal from each camera is fed to a matching video-processor and finally sampled into a Macintosh computer.

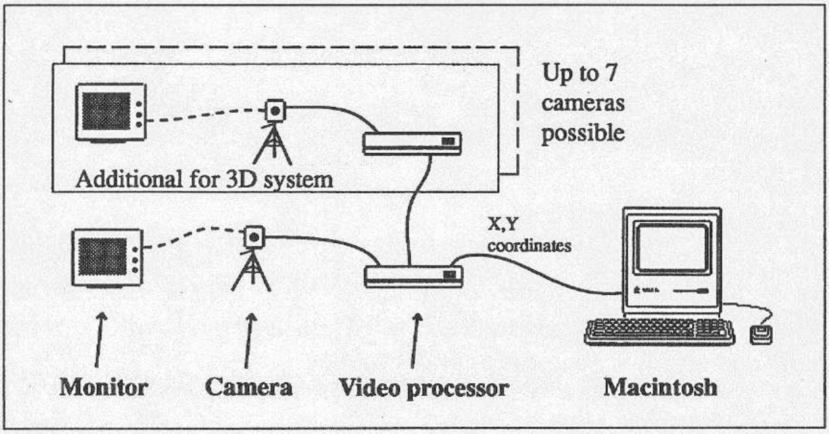

Figure 7. An overview of the physical configuration of the system.

The system has to be calibrated prior to a session of 3D measurements, i.e. the origin and orientation of an orthogonal coordinate system (the laboratory coordinate system) must be established. This calibration is valid as long as the positions and orientations of the cameras are not altered. To facilitate this procedure a calibration fixture with six markers is placed in the measurement volume, that is the volume seen by all cameras. The localization of the six markers is factory pre-set and the corresponding coordinate information relative to the lower left corner of the calibration fixture (defined as origin) is pre-stored in a calibration program. After sampling for approximately one second, the calibration program calculates the current camera positions relative to the chosen laboratory coordinate system.

The relative errors and inconsistencies in using the system were investigated in three experiments. In the first experiment, inherent errors caused by manufacturing defects of the calibration fixture (see Figure 8) and errors introduced due to difficulties in positioning the calibration fixture in alignment with a chosen laboratory coordinate system were studied. In the second experiment, the consistency in repeated positioning of a rigid link system, modelling the adequate body segments of the upper extremity, were studied (see Figure 9). Finally, an experiment evaluating the errors and inconsistencies associated with measurements on a subject furnished with markers attached to an exoskeleton, were performed (see Figure 10).

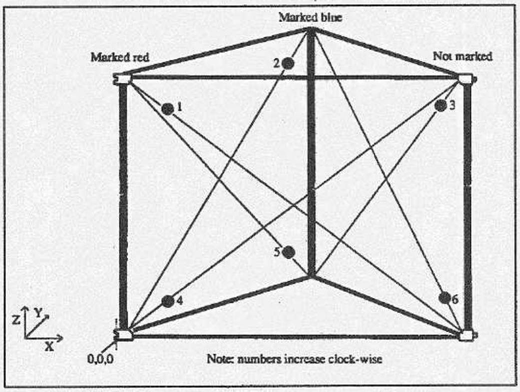

Figure 8. The calibration fixture CAB-800 for 3D measurements is composed of six spherical markers mounted in a framework. The spherical markers, with a diameter of 40 mm, are arranged in two triangles.

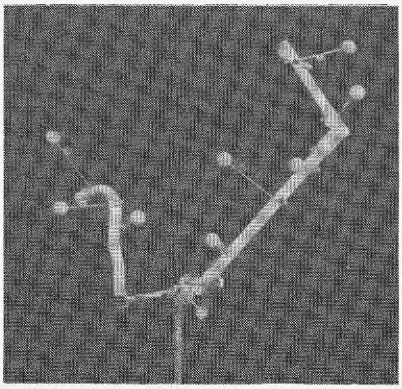

Figure 9. The mechanical link system modelling the thorax, the upper arm and the forearm. Each link is equipped with three spherical markers, with a diameter of 20 mm.

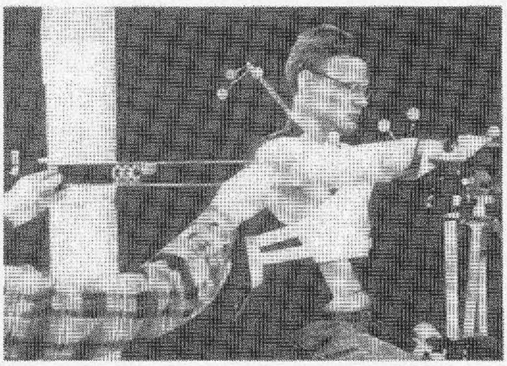

Figure 10. A subject furnished with the exoskeleton system. Each part of the exoskeleton system is equipped with three spherical markers, with a diameter of 20 mm.

Measures for evaluating inconsistency and relative error

Seven different measures were used to characterize the measurement errors when using the MacReflex system. Ideally all these measures should be equal to zero.

When a segment (or a calibration frame) is moved, the motion may be described by a translation followed by a rotation and a deformation. The *orthogonality test* (ε), as well as the *determinant test* (Δ) measure different aspects of the deformation part. For any particular reduction point defining the translatory part of the transformation, both tests estimate the closeness of the best remaining three point transformation (T) to a rotation. In the vernacular of small deformation theory, ε estimates the largest measured shear deformation of the segment, while Δ describes the relative change of volume.

The orthogonality test (ε):

$$\varepsilon_{ij} = \left(T^{tr}T - 1\right)_{ij}$$

where *tr* stands for the transpose and 1 is the unit matrix and ε is defined:

$$\varepsilon = \max_{ij}\left|\varepsilon_{ij}\right|$$

The determinant test (Δ):

$$\Delta = \det T - 1 = (1+\Delta_1)(1+\Delta_2)(1+\Delta_3) - 1 \cong \Delta_1 + \Delta_2 + \Delta_3$$

Without knowing the true position of the markers, a good estimate of the accuracy can be obtained by determining the *maximal coordinate difference (Δx)* or the *maximal distance difference (Δd)* between repeated measurements. The *root mean square of the coordinate differences (RMS)* represents also a measure of the accuracy similar to the *distance difference.* The *relative distance difference (δ)* can be calculated from the distance difference. The *angle error (τ)* defines the angle error in positioning the calibration fixture.

Maximal coordinate difference (Δx):

The coordinate difference between the recorded coordinate and the originally measured coordinate or reference coordinate given by the manufacturer *o* for coordinate *j* of marker *i*:

$$\Delta x_{ij} = x_{ij} - x_{ioj}$$

and the maximal coordinate difference for each recording:

$$\Delta x = \max_{ij} \left| \Delta_{ij} \right| = \max_{ij} \left| x_{ij} - x_{ioj} \right|$$

Maximal distance difference (Δd):

The distance difference between the recorded marker position and the originally measured marker position or the reference marker position given by the manufacturer *o* for marker *i*:

$$\Delta d_i = \sqrt{\sum_{j=1}^{3} \Delta x_{ij}^2} = \sqrt{\sum_{j=1}^{3} \left(x_{ij} - x_{ioj}\right)^2}$$

and the maximal coordinate difference for each recording:

$$\Delta d = \max_i \Delta d_i = \max_i \sqrt{\sum_{j=1}^{3} \Delta x_{ij}^2} = \max_i \sqrt{\sum_{j=1}^{3} \left(x_{ij} - x_{ioj}\right)^2}$$

Root Mean Square of the coordinate differences (RMS):

The RMS of the coordinate difference between the recorded coordinate and the originally measured coordinate or reference coordinate given by the manufacturer o for marker i:

$$RMS_i = \sqrt{\sum_{j=1}^{3}\left(\Delta x_{ij}\right)^2} = \sqrt{\sum_{j=1}^{3}\left(x_{ij} - x_{ioj}\right)^2}$$

and the Root Mean Square for all markers:

$$RMS = \sqrt{\frac{1}{N}\sum_{i=1}^{N}\sum_{j=1}^{3}\left(x_{ij} - x_{ioj}\right)^2}$$

where N indicates the number of markers.

Relative distance difference (δ):

The relative distance difference between marker i and marker j and the originally measured relative distance difference or the "true relative distance difference" given by the manufacturer between marker i and marker j:

$$\delta_{ij} = \frac{d_{ij} - d_{oij}}{d_{oij}}$$

and the maximal relative distance difference for each recording:

$$\delta = \max_{ij}\left|\delta_{ij}\right| = \max_{ij}\frac{d_{ij} - d_{oij}}{d_{oij}}$$

Angle error (τ):

The angle error between the recorded coordinate and the originally measured coordinate or reference coordinate given by the manufacturer o for coordinate j of marker i:

$$\tau_{ij} = \frac{\sqrt{\sum_{j=1}^{3}\left(x_{ij} - x_{ioj}\right)^2}}{d_{oik}} * \frac{180}{\pi}$$

and the maximal angle error for each recording:

$$\tau = \max_{i,k}\tau_{ij} = \max_{i,k}\frac{\sqrt{\sum_{j=1}^{3}\left(x_{ij} - x_{ioj}\right)^2}}{d_{oik}} * \frac{180}{\pi}$$

Results

Measurements on the calibration fixture:

Calibration – Registration

Calibration and registration was carried out five times with repositioning of the calibration fixture after each registration. The following measures were obtained:

$\Delta x \leq 2.4$ mm	$\delta \leq 4.2 \times 10^{-3}$	$\alpha = 0.00° \pm 0.01$
$\Delta d \leq 2.6$ mm	$\varepsilon \leq 1.6 \times 10^{-3}$	$\beta = 0.00° \pm 0.02$
RMS ≤ 1.6 mm	$\Delta \leq 1.0 \times 10^{-3}$	$\lambda = 0.00° \pm 0.02$
	$\tau \leq 0.26°$	

Adjusting the calibration fixture to a chosen laboratory coordinate system

One initial calibration followed by five registrations with repositioning of the calibration fixture before each registration. The following measures were obtained.

$\Delta x \leq 2.8$ mm	$\delta \leq 4.1 \times 10^{-3}$	$\alpha = 0.00° \pm 0.01$
$\Delta d \leq 2.6$ mm	$\varepsilon \leq 1.8 \times 10^{-3}$	$\beta = 0.00° \pm 0.02$
RMS ≤ 1.7 mm	$\Delta \leq 1.1 \times 10^{-3}$	$\lambda = 0.00° \pm 0.04$
	$\tau \leq 0.26°$	

Rotation around one axis

Using a protractor, the calibration fixture was rotated 40° around the 3-axis. Five registrations were made with repositioning of the calibration fixture before each registration. The following measures were obtained:

$\delta \leq 15.0 \times 10^{-3}$	$\alpha = 40.0° \pm 1.4$
$\varepsilon \leq 28.0 \times 10^{-3}$	$\beta = 0.0° \pm 0.2$
$\Delta \leq 15.0 \times 10^{-3}$	$\lambda = 0.0° \pm 0.1$

Successive rotation around two axis

Two successive rotations were made: 40° around the 3-axis and then 20° around the 1-axis. Five registrations were made with repositioning of the calibration fixture before each registration. The following measures were obtained:

$\delta \leq 12.0 \times 10^{-3}$ $\qquad$ $\alpha = 40.0° \pm 1.5$

$\varepsilon \leq 25.0 \times 10^{-3}$ $\qquad$ $\beta = 0.0° \pm 0.1$

$\Delta \leq 13.0 \times 10^{-3}$ $\qquad$ $\lambda = 20.0° \pm 0.5$

Measurement on the mechanical model

Measurement of a reference position (position 1)

The Euler angles for the three body segments relative to the laboratory coordinate system for the reference position:

Thorax	$\alpha t = 0°$	$\beta t = 0°$	$\gamma t = 0°$
Upper arm	$\alpha h = 0°$	$\beta h = 0°$	$\gamma h = 0°$
Forearm	$\alpha a = 90.0°$	$\beta a = 0°$	$\gamma a = 0°$
Elbow angle	$\Phi = 90.0°$		
Forearm rotation	$\gamma = 0°$		

Five registrations were made with repositioning of the mechanical system before each registration. The first picture in the first trial was used as a reference position. The consistency in repeated positioning of the model is presented in Table 4.

Table 4. Repeated measurement on the mechanical shoulder model in the reference position relative to the laboratory system (position 1).

Trial	Body Segment							Euler angles relative the laboratory system			Euler angles relative the thorax system			Elbow angle	Forearm rotation
No.		Δx mm	Δd mm	RMS mm	$\delta \cdot 10^{-3}$	$\varepsilon \cdot 10^{-3}$	$\Delta \cdot 10^{-3}$	α deg.	β deg.	γ deg.	α deg.	β deg.	γ deg.	Φ deg.	γ_u deg.
1								0.0	0.0	0.0					
1								0.0	0.0	0.0	0.0	0.0	0.0		
1								90.0	0.0	0.0	90.0	0.0	0.0	90.0	0.0
2	Thorax	4.0	4.5	4.0	-2.7	13.0	14.0	-0.3	0.2	0.0					
2	Humerus	5.8	6.9	6.0	1.2	5.3	3.5	-0.1	0.8	0.6	0.2	0.6	0.6		
2	Forearm	8.9	11.2	9.6	-2.4	-6.8	1.9	89.8	0.0	-1.0	90.1	0.0	-0.8	90.1	-0.2
3	Thorax	4.7	5.2	4.9	-1.6	9.7	10.0	0.0	0.2	0.1					
3	Humerus	4.2	5.0	4.5	4.8	9.8	2.6	0.1	0.9	0.4	0.1	0.7	0.4		
3	Forearm	6.8	7.5	7.1	3.6	-8.0	7.0	90.0	-0.6	-1.3	90.0	-0.7	-1.0	90.1	-0.3
4	Thorax	4.8	5.5	5.0	-2.9	16.0	17.0	-0.2	0.3	0.1					
4	Humerus	4.6	5.0	4.0	-3.3	7.5	4.9	0.0	0.4	0.9	0.2	0.1	0.9		
4	Forearm	5.7	7.2	6.1	-4.8	8.5	1.5	89.9	0.4	-0.6	90.2	0.3	-0.3	90.0	-0.2
5	Thorax	4.3	5.8	4.8	-3.7	13.0	14.0	-0.5	0.2	0.1					
5	Humerus	5.8	6.7	5.9	-1.7	4.6	4.3	-0.4	0.8	0.1	0.1	0.5	0.0		
5	Forearm	10.7	11.5	9.3	5.4	-13.0	6.6	89.5	-1.1	-0.9	90.0	-1.2	-0.7	90.1	-0.2

Rigid rotation from position 1 to position 2

The mechanical model was rigidly rotated –30° around the 3-axis of the laboratory coordinate system. The following Euler angles for the three body segments relative to the laboratory coordinate system were obtained (position 2):

Thorax	$\alpha t = -30.0°$	$\beta t = 0°$	$\gamma t = 0°$
upper arm	$\alpha h = -30.0°$	$\beta h = 0°$	$\gamma h = 0°$
Forearm	$\alpha a = 60.0°$	$\beta a = 0°$	$\gamma a = 0°$
elbow angle	$\Phi = 90.0°$		
Forearm rotation	$\gamma = 0°$		

Five registrations were made with repositioning of the mechanical system before each registration. The first picture in the first trial in position 1 was used as a reference position. The consistency in repeated positioning of the model is presented in Table 5.

Table 5. Repeated measurement on the mechanical shoulder model after an overall rigid rotation around the 3-axis of the laboratory system (position 2).

Trial No.	Body Segment	$\delta \cdot 10^{-3}$	$\varepsilon \cdot 10^{-3}$	$\Delta \cdot 10^{-3}$	Euler angles relative the laboratory system α deg.	β deg.	γ deg.	Euler angles relative the thorax system α deg.	β deg.	γ deg.	Elbow angle Φ deg.	Forearm rotation γ_u deg.
1	Thorax	-6.9	13.0	15.0	-29.1	-0.8	-0.6					
1	Humerus	-2.3	3.1	4.0	-29.9	-0.2	-0.6	-0.8	0.6	0.0		
1	Forearm	5.4	-13.0	15.0	60.0	-0.4	0.2	89.1	0.2	-0.6	90.1	0.1
2	Thorax	-6.5	13.0	9.4	-29.7	-0.7	-0.9					
2	Humerus	-3.5	6.2	8.5	-30.3	0.7	-0.2	-0.7	1.4	0.7		
2	Forearm	6.2	-19.0	-20.0	59.3	-0.3	-0.7	89.0	0.7	-1.4	90.3	0.0
3	Thorax	-11.0	21.0	17.0	-28.8	-0.8	-1.0					
3	Humerus	39.0	5.6	-3.0	-30.0	1.1	-0.4	-1.3	1.9	0.6		
3	Forearm	59.0	-14.0	15.0	59.9	-0.8	-1.1	88.6	0.2	-1.8	90.1	0.1
4	Thorax	-10.0	20.0	11.0	-29.1	-0.8	-1.1					
4	Humerus	3.9	11.0	4.2	-30.4	0.6	0.1	-1.3	1.3	1.2		
4	Forearm	6.5	-13.0	15.0	59.5	-0.2	-0.4	88.6	0.9	-1.2	90.1	0.2
5	Thorax	-13.0	26.0	13.0	-28.5	-0.7	-1.1					
5	Humerus	-2.6	7.6	6.9	-29.7	0.6	-0.7	-1.2	1.2	0.4		
5	Forearm	7.2	-20.0	22.0	60.3	-1.1	-0.5	88.7	0.1	-1.2	90.0	0.1

Ridged rotation from position 2 to position 3

The humerus and the forearm were rigidly rotated first – 45° around the humerus 2-axis and then 45° around the humerus 1-axis. The following Euler angles for the three body segments relative to the laboratory coordinate system were obtained (position 3):

Thorax	$\alpha_t = -30.0°$	$\beta_t = 0.0°$	$\gamma_t = 0.0°$
upper arm	$\alpha_h = -30.0°$	$\beta_h = 45.0°$	$\gamma_h = 45.0°$
Forearm	$\alpha_a = 95.0°$	$\beta_a = 30.0°$	$\gamma_a = 55.0°$
elbow angle	$\Phi = 90.0°$		
Forearm rotation	$\gamma = 0.0°$		

Five registrations were made with repositioning of the mechanical system before each registration. The first picture in the first trial in position 1 was used as a reference position. The consistency in repeated positioning of the model is presented in Table 6.

Table 6. Repeated measurement on the mechanical shoulder model after a rotation around the 2- and 1-axis of the humerus system (position 3).

Trial No.	Body Segment	$\delta \cdot 10^{-3}$	$\varepsilon \cdot 10^{-3}$	$\Delta \cdot 10^{-3}$	Euler angles relative the laboratory system α deg.	β deg.	γ deg.	Euler angles relative the thorax system α deg.	β deg.	γ deg.	Elbow angle Φ deg.	Forearm rotation γ_u deg.
1	Thorax	-6.8	13.0	10.0	-29.2	-0.7	-0.7					
1	Humerus	-13.0	29.0	31.0	-30.9	45.1	46.3	-2.5	45.7	47.2		
1	Forearm	-2.3	8.7	9.5	94.6	30.4	-54.9	124.3	30.5	-56.0	90.2	0.0
2	Thorax	-6.5	12.0	4.9	-29.7	-0.7	-0.9					
2	Humerus	-5.8	21.0	23.0	-31.8	45.8	45.3	-3.2	46.4	46.7		
2	Forearm	-6.4	11.0	9.1	93.4	29.3	-55.0	123.6	29.7	-56.3	90.0	0.3
3	Thorax	-9.2	18.0	12.0	-28.9	-0.7	-0.9					
3	Humerus	-4.6	19.0	16.0	-31.3	45.1	46.5	-3.3	45.8	47.8		
3	Forearm	-3.2	11.0	12.0	94.8	30.2	-55.2	124.4	30.5	-56.5	89.8	0.1
4	Thorax	-11.0	21.0	9.0	-29.2	-0.6	-1.2					
4	Humerus	-8.3	27.0	31.0	-31.5	45.3	46.8	-3.5	45.8	48.5		
4	Forearm	-3.1	9.7	11.0	94.7	30.3	-55.3	124.6	30.9	-56.7	90.0	0.1
5	Thorax	-13.0	25.0	11.0	-28.5	-0.6	-1.3					
5	Humerus	-3.0	12.0	10.0	-31.1	44.7	46.5	-3.9	45.2	48.3		
5	Forearm	-3.9	11.0	13.0	94.9	30.3	-54.8	124.1	31.0	-56.2	89.6	0.1

Measurements on a subject with an exoskeleton

The subject was oriented relative the laboratory coordinate system using protractors and reference lines. The position chosen was the thorax facing to the right, 90° forward flexion of the upper arm, and 90° of elbow flexion with a horizontal forearm. The following Euler angles for the three body segments relative to the laboratory coordinate system outline this position.

Thorax	$\alpha_t = -90.0°$	$\beta_t = 0.0°$	$\gamma_t = 0.0°$
upper arm	$\alpha_h = -90.0°$	$\beta_h = 0.0°$	$\gamma_h = 0.0°$
Forearm	$\alpha_a = 0.0°$	$\beta_a = 0.0°$	$\gamma_a = 0.0°$

The first picture in the first experiment was used as a reference for all the others. Eight registrations were made with repositioning of the subject before each registration, except for the first trial. The consistency in repeated positioning of the model is presented in Table 7.

Table 7. Repeated measurement on a subject with an exoskeleton.

Trial No.	Body Segment	$\delta \cdot 10^{-3}$	$\varepsilon \cdot 10^{-2}$	$\Delta \cdot 10^{-2}$	Euler angles relative the laboratory system α deg.	β deg.	γ deg.	Euler angles rel.ative the thorax system α deg.	β deg.	γ deg.	Elbow angle Φ deg.	Forearm rotation γ_u deg.
1	Thorax	3.60	-1.80	-1.50	-89.9	0.4	0.1					
1	Humerus	-0.51	-0.09	0.04	-90.0	0.0	0.1	0.0	-0.3	0.0		
1	Forearm	-0.96	0.20	0.20	0.0	0.0	-0.1	89.9	-0.1	0.3	90.0	-0.1
2	Thorax	17.00	-3.70	-3.90	-90.3	0.5	-4.0					
2	Humerus	13.00	-2.30	-2.30	-91.4	2.1	3.7	-1.1	1.5	7.7		
2	Forearm	9.50	-3.40	-3.50	-5.3	-3.1	-5.8	85.1	0.9	-4.9	93.8	-3.9
3	Thorax	1.30	0.50	0.31	-88.9	3.0	-2.8					
3	Humerus	-0.15	2.50	0.57	-86.6	2.9	-0.5	2.3	0.0	2.5		
3	Forearm	-9.10	1.90	-0.82	-2.1	-3.2	-5.0	87.1	-0.5	-1.9	95.3	-2.1
4	Thorax	16.00	-2.60	2.30	-90.3	0.9	-3.4					
4	Humerus	-33.00	4.70	3.40	-87.9	0.6	-0.9	2.4	-0.2	2.5		
4	Forearm	-17.00	2.90	-0.07	-3.9	-4.7	0.2	86.5	-1.3	1.4	95.9	0.9
5	Thorax	-26.00	7.10	7.40	-92.8	5.2	-2.9					
5	Humerus	-55.00	11.00	12.00	-81.8	-0.6	-0.8	11.3	-5.2	3.1		
5	Forearm	-17.00	2.30	2.20	1.5	1.6	-11.7	94.1	4.9	-6.7	96.7	-12.2
6	Thorax	-2.20	1.10	1.10	-88.1	1.5	-7.0					
6	Humerus	-18.00	2.70	2.50	-85.7	1.9	-0.9	2.3	0.7	6.1		
6	Forearm	4.50	-1.90	-1.80	-0.5	-3.2	-7.4	87.7	3.7	-5.7	94.6	-5.4
7	Thorax	-2.00	0.26	0.28	-89.9	1.1	5.3					
7	Humerus	-16.00	2.40	2.60	-85.3	0.7	0.1	4.5	-0.8	-5.0		
7	Forearm	6.10	-2.00	-1.90	-2.6	-3.3	-7.9	87.3	-8.6	-7.1	97.3	-7.2
8	Thorax	-4.60	1.10	1.20	-90.9	2.9	-0.4					
8	Humerus	-53.00	10.00	11.00	-83.7	-2.3	-0.7	7.3	-5.2	0.1		
8	Forearm	7.20	-2.10	-1.80	0.8	-2.0	-6.4	91.8	-1.5	-3.6	95.5	-8.7

Conclusions

In the first experiment on the calibration fixture, the intrinsic errors of the system were investigated. The results depended upon the manufacturing and the rigidity in the calibration fixture. The values of the relative distance difference (δ) and the root mean square of the coordinate differences (RMS) were found to be consistent with what could be expected, considering the construction, and can be regarded as essentially negligible. Also, the alignment of the calibration fixture to the chosen laboratory coordinate system does not seem to introduce significant error. Additionally, in the experiment with the rotation of the calibration fixture, there was reasonable correspondence between the protractor measurements and MacReflex registrations.

Therefore, the position of a rigid body can easily and accurately be determined with the MacReflex system.

The measurement on the mechanical model, indicating the ability to accurately position the model, showed a maximal difference of 1.8° from the expected value; in general, the difference was smaller. However 1.8° is an acceptable value for ergonomic situations.

Although a relatively crude method for positioning was used in the exoskeleton measurements, the angles obtained are fairly constant (except for trial 5 and 8, probably due to mixed markers).

In conclusion, the MacReflex motion analysis system makes a valuable contribution to the existing small group of instruments for non-contact posture measurements in the fields of biomechanics and ergonomics. It determines position information with adequate accuracy under normal handling.

Paper III

Aims

- to evaluate voluntary reduction of muscle activity in the shoulder by EMG feedback assistance, while maintaining a fixed arm position

- to evaluate the reliability of trapezius EMG as a descriptor of total shoulder muscle load

Material and Methods

Six young female subjects participated in this study. Their mean age was 30 years (range 24-39), the mean body weight 64 kg (range 58-72), and the mean height 169 cm (range 167-173). All subjects were right-handed, and the EMG registration was made from the right shoulder. Five shoulder muscles were studied: the supraspinatus, the infraspinatus, the anterior and middle parts of the deltoid, and the upper part of the trapezius. The EMG activity was picked up by means of single intramuscular wire electrodes and the indifferent electrode was a Ag/AgCl surface electrode, taped at the processus prominens (C7).

Signal processing

The EMG signals were amplified 1000 times in an amplifier with a frequency range of 2 Hz - 3 kHz. The amplified EMG signals were recorded on a FM tape-recorder with a bandwidth covering DC to 1,25 kHz (TEAC R-71). The quality of the recorded muscle signals was visually monitored on an oscilloscope and on a chart recorder. One of these signals was also presented to the subject on a VDU. The final processing of the EMG signals was performed on a digital computer (PDP 11/44) with a special analysis software package (Arvidsson 1982, Kadefors et al. 1983). The EMG signals were sampled from the tape recorder via an anti-aliasing filter with a cut-off frequency of 800 Hz, to an A/D converter with a sampling rate of 2048 Hz. RMS and MPF were calculated for consecutive segments with half a second duration. Each segment was submitted to an automatic quality control procedure to eliminate possible signal disturbances (Arvidsson 1982).

Experimental Procedure

The subject sat on a chair in a vertical fixed position, with the VDU in a comfortable observing location, as illustrated in Figure 11. The subject was instructed to position her right arm in different angles of abduction in a vertical plane 45° from the sagittal plane (here called the scapula plane). The rotation of the humerus was controlled by the orientation of the thumb, which was kept upwards. The subject was asked to try to minimize the displayed EMG signal, without changing the position of her arm. The shoulder muscles were tested one by one, and only the signal from the current muscle was displayed to the subject.

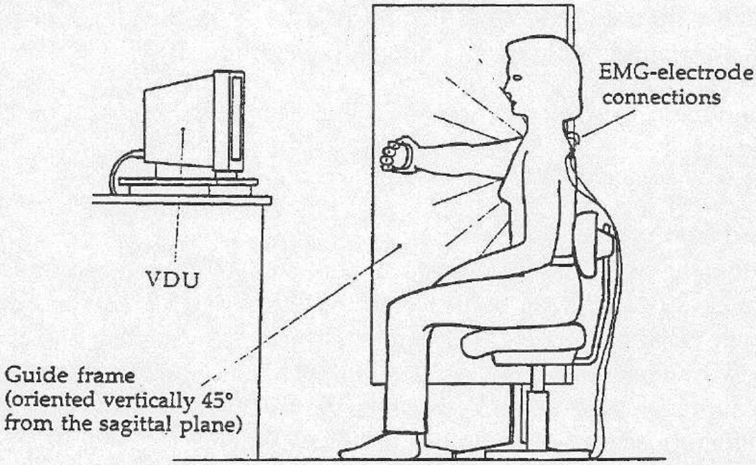

Figure 11. The experimental set-up

The minimizing effort continued for about one minute. The mean value of the segments during the period between the 5th and the 15th second was regarded as the initial muscle activity (IMA), and the mean value of the period between the 30th and the 40th second as the final muscle activity (FMA). The positioning of the arm was controlled by a guiding frame and was carefully checked throughout the experiment. The statistical comparisons were carried out using the t-test.

The ability to reduce and control the EMG activity was studied in different shoulder muscles and positions. In the first study three muscles in three arm positions were investigated, and in the second study five muscles in one arm position were investigated (see Table 8). The EMG activity in each muscle was, one at a time, fed back to the subject via the VDU. The EMG activity of the shoulder muscles was recorded. Between each trial, the subject was allowed to rest for two minutes. The order of the muscle and arm position combinations was randomized for each subject.

Table 8. Shoulder muscles and arm positions investigated in study 1 and study 2. All arm positions were performed in the scapula plane.

	Study 1	Study 2
Muscles	Supraspinatus Infraspinatus Trapezius descendens	Supraspinatus Infraspinatus Trapezius descendens Deltoid anterior Deltoid medialis
Shoulder abduction	30°, 60° and 90°	30°
Elbow flexion	0°	90°

Results

The change in muscle activity was described as the ratio between the final muscle activity (FMA) and the initial muscle activity (IMA). The value was first calculated for each individual, and then the mean value for the whole group was determined.

In the first study, the trapezius muscle activity was significantly ($p < 0.01$) reduced to an average of 53% of the initial muscle activity at 90° of abduction (Figure 12). At 30° and 60° of abduction, the final muscle activities decreased to 78% (n.s.) and 64% ($p < 0.05$), respectively, from the initial muscle activity. In the supraspinatus muscle, there was also a tendency toward a reduction (89%, n.s.) at 30° of abduction, but at 60° and 90° the EMG activity was unaffected. The middle part of the deltoid muscle showed a tendency toward an increase (n.s.) in activity for all three position.

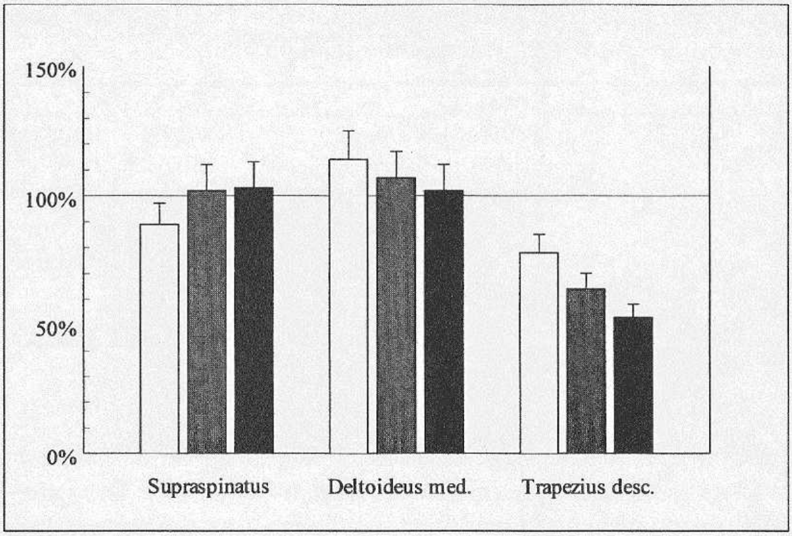

Figure 12. The ratio between the FMA and the IMA in three shoulder muscles and three arm positions. The standard error of the mean (SEM) is indicated for the mean values from six subjects.

In the second study, the trapezius muscle again appeared to be subject to a voluntary control. The muscle activity was reduced to 56% ($p < 0.01$) of the initial muscle activity (Figure 13). The supraspinatus muscle activity was also slightly reduced, however not significantly. The infraspinatus muscle and the anterior and medial parts of the deltoid muscles showed tendencies toward increased activity to 116% (n.s.), 123% (n.s.) and 130% (n.s.), respectively.

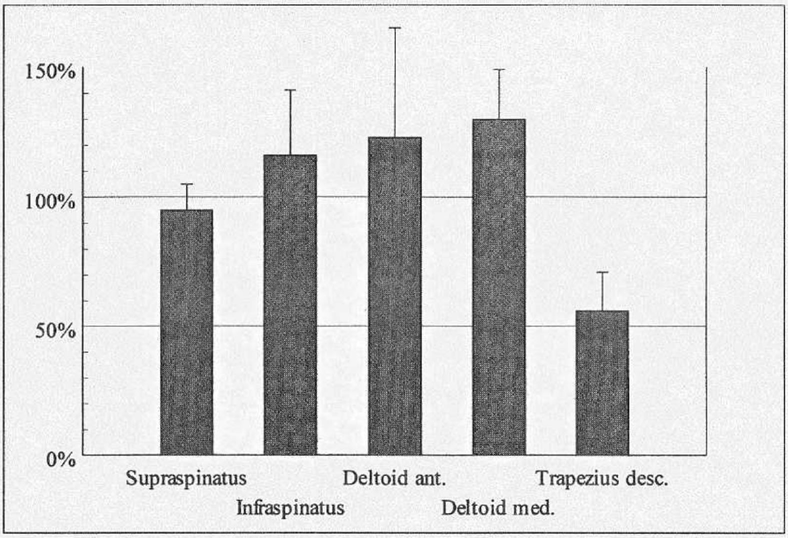

Figure 13. The ratio between the FMA and the IMA in five shoulder muscles and one arm position. The standard error of the mean (SEM) is indicated for the mean values from six subjects.

When the trapezius muscle activity was voluntarily reduced by the subjects, the remaining four shoulder muscles were also affected, as displayed in Figure 14. It shows that there was a tendency toward an increase in the muscle activity in the infraspinatus muscle to 108% ($p < 0.02$). The activity in the supraspinatus muscle and the anterior part of the deltoid muscle was unaffected, and the activity in the middle part of the deltoid muscle was reduced to 87% (n.s). An analysis of MPF of the EMG spectrum as an indicator of fatigue was performed. It was found that significant EMG changes indicative of localized muscle fatigue were found only in the infraspinatus muscle.

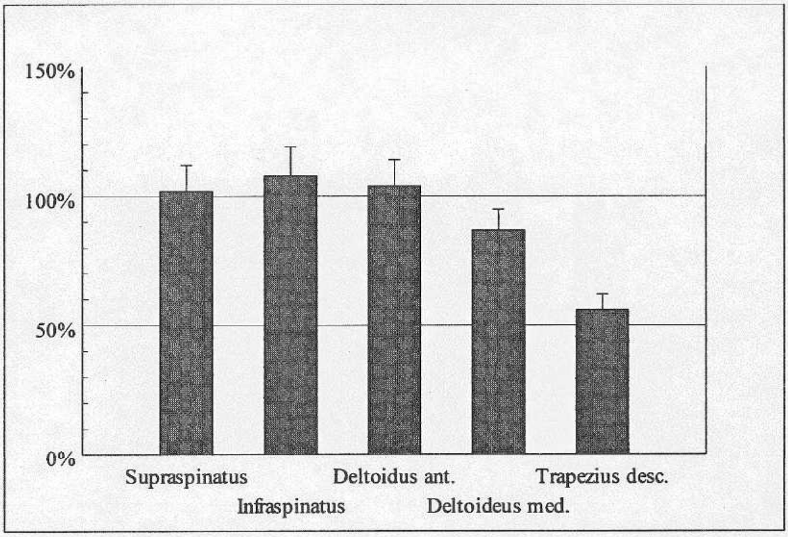

Figure 14. The relative changes in muscle activity in five shoulder muscles induced by voluntary reduction of trapezius activity assisted by EMG feedback from the trapezius muscle.

Conclusions

There is an ability to reduce EMG activity voluntarily in the trapezius muscle, without changing the hand load or the arm posture. This ability could not be detected in any other shoulder muscle investigated in this study.

It is suggested that it is not possible to rely uncritically on trapezius EMG measurements when estimating total shoulder load, since there is a significant voluntary effect in this muscle, despite a fixed total shoulder load.

Paper IV

Aims

- to investigate whether voluntary reduction of trapezius muscle activity was possible due to an initial overstabilization in the descending part of the trapezius muscle and its antagonists, or due to redistribution of load among its synergists.
- to identify the compensating synergists of the descending part of the trapezius muscle, if there is a redistribution of muscle activity.

Material and Methods

Subjects

Eleven out of twelve healthy female subjects, with no history of shoulder pain, completed the whole test. The mean age of the subjects was 33 years (range 26-45), the mean body weight 64 kg (range 52-80) and the mean height 167 cm (range 165-171). All subjects were right-handed, and the EMG registrations were all made from the right side.

Simulated redistribution of shoulder muscle activity

To select relevant muscle for recording EMG, the shoulder model was utilized. First, the "normal" force distribution was calculated for all shoulder muscles in six different arm positions: 30°, 60° and 90° of humeral abduction with a straight arm, and 30°, 60° and 90° of humeral abduction with 90° of flexion in the elbow. Then the force distribution was calculated for the corresponding positions, but with an imposed condition, i.e. the descending part of the trapezius muscle was totally relaxed. The difference between the "normal" and the "manipulated" force distribution was calculated for each muscle and arm position, and the muscles were arranged according to the magnitude of force differences for all arm positions. The six most affected shoulder muscles (minor muscles or muscles included in the study described in Paper III excluded) were included in the study, besides the descending part of the trapezius (see table 9).

Electrodes

Bipolar surface electrodes as well as single intramuscular wire electrodes were used. The indifferent electrode was a Ag/AgCl surface electrode, taped at the processus prominens (C7).

Table 9. Electrode type and electrode placement for the seven studied muscles.

Muscle	Electrode type	Electrode placement
Deltoid anterior	Bipolar, surface	within the elongated oval area below the lateral end of the clavicle
Levator scapulae	Monopolar, intramuscular	approximately 2 cm medial and ½ cm cranial of the superior angle of the scapula
Rhomboid major	Monopolar, intramuscular	horizontally: approximately 1 cm medial of the medial border of the scapula and vertically: midway between the base of the spine of the scapula and the inferior angle of the scapula
Rhomboid minor	Monopolar, intramuscular	horizontally: approximately 1 cm medial of the medial border of the scapula and vertically: close to the base of the spine of the scapula
Medial part of the Serratus ant.	Bipolar, surface	on the 4th rib from the top on the mid-axillary line
Descending part of the Trapezius	Monopolar, intramuscular	half-way on a line between the acromial angle and the 7th cervical vertebrae
Transverse part of the Trapezius	Bipolar, surface	half-way on a horizontal line between the midpoint of the spine of the scapula and the vertebral column

Arm positions

The ability to reduce and control the EMG activity was studied in six different arm positions. The positions were 30°, 60° and 90° of humeral abduction with a straight elbow and 30°, 60° and 90° of humeral abduction with 90° of flexion in the elbow. All arm positions were performed with the upper arm in a vertical plane 45° from the sagittal plane. A guiding frame was arranged close by the subject's upper arm to assist the subject to assume and maintain the correct elevation angle and elevation plane, and to assist the investigator in supervising the arm position.

In all arm positions, the rotation of the upper arm and the forearm was kept neutral. In all arm positions with a flexed elbow, the rotation of the forearm was controlled by the orientation of the thumb, which was kept in the same plane as the upper arm and the forearm. In all positions with a straight arm, the thumb was kept pointing perpendicular to the elevation plane.

Signal processing and calculations

The electrodes were connected to an eight-channel EMG amplifier. The gain was adjusted to the pick-up conditions of each electrode, implying a gain of 1000-5000. The frequency range of the complete amplifier was 8- 800 Hz. The amplified EMG signals were sampled to a personal computer with a sampling rate of 2048 Hz. The EMG signal from the descending part of the trapezius was displayed on a VDU using an analogue video mixer.

The mean value of the EMG signal during the first five seconds was regarded as the initial muscle activity (IMA), and the mean value of the last five seconds as the final muscle activity (FMA). If momentary artifacts occurred during a segment, that segment was excluded from the analysis, or if the artifacts occurred during the whole interval, the interval was shifted forward or backwards in time sufficiently enough to avoid any artifacts.

The change of muscle activity is described as the ratio between the final muscle activity (FMA) and the initial muscle activity (IMA). The value was first calculated for each individual, and then the mean value for the whole group was formed.

The statistical comparisons were carried out using a two-tailed t-test.

Experimental procedure

The subject was sitting in a vertical fixed position, with the VDU in a comfortable observing location (Figure 11). The subject was first requested to relax the shoulder supporting the arm in the lap, to control for psychogenic muscle activity, especially in the descending part of the trapezius. The subject was then instructed to position the right arm in different angles of abduction in the scapula plane. The order of arm positions was randomized for each subject. Each position was measured twice.

The subject was then asked to minimize the displayed EMG signal from the descending part of the trapezius, without changing the position of the arm. The minimizing effort continued for about one minute, during which time the EMG activities of the seven shoulder muscles were recorded. Between each trial, the subject was allowed to relax for two minutes in a sitting position, with the forearms resting in the lap. The subject was allowed no training prior to the experiment.

Results

The ability to reduce the muscle activity of the descending part of the trapezius, as found in Paper III, was confirmed. Nine subjects out of eleven were able to reduce the muscle activity. The mean value (11 subject) of the muscle activity for the *descending part of the trapezius* was 67%, ranging from 60% to 76% depending on the arm position (table 10).

The reduction in activity in the descending part of the trapezius muscle influenced all the other muscles studied. The muscles most affected were the *major and minor rhomboid* and the *transverse part of the trapezius*, showing significant ($p<0.05$) increased activities of 232%, 175% and 201%, respectively, compared to the initial activity. Also, the *anterior part of the deltoid* and the *serratus anterior* activity increased significantly to 119% and 131%, respectively, compared to the initial activity. However, the muscle activity of the *levator scapulae* decreased in all positions, except for 60° of humeral abduction with a flexed elbow. The mean value for all positions was 93% compared to the initial activity, which was however not significant.

Table 10. The ratio between the final muscle activity voluntarily reduced by EMG feedback from the descending part of the trapezius, and the initial muscle activity in seven shoulder muscles for six different arm positions. A= abduction in the scapula plane, f= elbow flexion, s = significant on 5% level, ns = not significant on 5% level).

Shoulder Muscles	A30°f0° (%)		A60°f0° (%)		A90°f0° (%)		A30°f90° (%)		A60°f90° (%)		A90°f90° (%)		Mean (%)
Levator Scapulae	89	ns	89	ns	90	ns	94	ns	109	ns	93	ns	94
Rhomboid Major	327	s	204	s	193	s	235	s	187	s	185	s	222
Rhomboid Minor	252	s	158	s	143	s	183	ns	172	ns	137	s	174
Middle part of the Serratus Anterior	132	s	119	s	130	s	127	s	133	s	135	s	129
Transverse part of the Trapezius	250	s	187	s	173	s	195	s	173	s	182	s	193
Anterior part of the Deltoid	127	s	120	s	109	ns	123	s	121	s	110	ns	118
Descending part of the Trapezius	65	s	64	s	68	s	75	s	81	ns	72	s	71

To investigate the possible influence of training effects on the motor control of the involved shoulder muscles during the experimental procedure, the first registration was compared with the second registration for each position and each muscle. This investigation indicated that there was no significant trend for any position or muscle.

Conclusions

Although individual differences were manifested, these results clearly indicate the following:

In relaxation of the descending part of the trapezius muscle when maintaining a specific arm posture, the load is distributed to other synergistic shoulder muscles.

A main part of the load is transferred to the rhomboid major and minor and the transverse part of the trapezius.

The levator scapulae did not adopt the load from the descending part of the trapezius. This was in contrast to results from simulation using a biomechanical model of the shoulder.

Special attention should be paid to the rhomboid muscles and the transverse part of the trapezius muscle when introducing feedback techniques to relax the descending part of the trapezius muscle.

Paper V

Aims

- to conduct a survey of IMP in the supra- and infraspinatus muscles related to different arm positions and external loads
- to identify a zone for the upper extremity, based on the affect of IMP on muscle recovery, where accumulated muscle fatigue might occur.
- to identify a zone for the upper extremity, based on the affect of IMP on local circulation, where impaired muscle blood flow might occur.

Material and Methods

The subjects consisted of a homogeneous group of young male athletes with a mean age of 26 years (22 - 33 years), a mean body weight of 80 kg (66 – 92 kg) and a mean height of 182 cm (174 - 195 cm). All of the subjects were right handed, except for one.

The preparation for the IMP registrations began after the experimental proceeding had been explained to the subject and he had given his consent. IMP was recorded from two shoulder muscles, the infraspinatus and supraspinatus, on the right hand side. IMP was studied in a total of 112 combinations of arm positions and hand loads. The investigated arm positions, were comprised of eight vertical *arm elevation planes* (AEPs) ranging from -15° to 90° with reference to the sagittal plane, and seven *arm elevation angles* (AEAs) ranging from 0° to 90° with reference to the vertical line. An increment of 15° was used for AEA as well as for AEP. All positions were performed with a straight arm, with or without a hand load of 1 kg. The order of AEPs and hand load was randomized.

Microcapillary infusion technique (MCI) was used for IMP measurements. Arm position measurements were made using a 3D motion analysis system (MacReflex). In the registration and calculation of the arm position, the torso, the upper arm and the forearm were considered to be rigid body segments. A forward flexion with a straight arm was chosen as the reference position. The amplified IMP signals and the posture data were sampled with a rate of 10 Hz. All data were synchronized using the BioPac® system and saved on a Macintosh computer.

The IMP variable was assumed to depend on subject, AEA, AEP and other unidentified factors. A general linear model was applied to estimate the distribution of the total variance and the significance of the independent factors.

The model was applied to each combination of hand load and shoulder muscle (4 combination).

Experimental Procedure

The subject was seated in a experimental chair with the trunk in a vertical position. On a given signal he started to elevate his right arm slowly from a vertical position along the side of the body, intending to reach an AEA of at least 90° in 10 seconds (angular velocity approximately 10°/s). The subject was allowed to rest for about two minutes between each arm elevation. The total experimental session lasted for approximately one hour, excluding the initial preparation of the subject. The load on the shoulder in the most strenuous positions corresponded to approximately 20% of the maximal voluntary contraction, and considering the activity/pause-ratio, the risk of fatigue accumulation was negligible.

Results

The relation between the mean IMP and the arm position showed a characteristic trend which was specific to each muscle. The mean IMP of the infraspinatus and the supraspinatus muscles increased monotonously from a resting pressure of 0 - 10 mmHg at 0° of AEA to a maximal pressure at 90° of AEA, for all eight vertical AEPs. The influence of the AEP on the supraspinatus IMP was obvious, but less prominent on the infraspinatus IMP. The development of the infraspinatus IMP was most conspicuous in forward flexion and adjacent AEPs. The development of the supraspinatus IMP was, on the contrary, more in abduction. The level of IMP in the supraspinatus muscle was considerably higher, compared to that of the infraspinatus muscle. A hand load of 1 kg resulted in a general increase in IMP in the investigated muscles for all arm positions. However, the general characteristics of the IMP and arm position relation did not change. The infraspinatus muscle appeared to be more load dependent than the supraspinatus, with the average increase being 94% for the infraspinatus and 58% for the supraspinatus.

Due to the disparities in IMP development, the levels of non-recovery (NR) (20 mmHg/2.6 kPa) and blood flow impairment (BFI) (40 mmHg/5.3 kPa) were reached at different arm positions for the infra- and supraspinatus muscles. The level of NR was reached at an AEA of approximately 30° in the supraspinatus as well as in the infraspinatus muscle. With 1 kg of hand load, the level of NR was exceeded already at an AEA of approximately 15° and 20° for the supraspinatus and infraspinatus muscles, respectively. The corresponding AEA for the BFI level was 50° for the supraspinatus muscle with no hand load. With 1 kg of hand load, the level of BFI was reached at an AEA of approximately 35° and 50° for the supraspinatus and infraspinatus muscles, respectively. New isobar diagrams were extracted from the infra- and supraspinatus data, by selecting the maximal IMP value of the two muscles in each arm position (see Figure 15). The zones, limited by the isobars, identify arm positions where none of the muscles exceeded a set level of IMP. As indicated by the medium gray color, NR level was not exceeded in any of the muscles, if the AEA was less than 30° and with 1 kg of hand load, none of the muscles exceeded the NR level, if the AEA was less than 15° (see Figure 15). As indicated by the light gray color, the BFI level was not exceeded in any of the muscles, if the AEA was kept lower than 50° with the unloaded hand, and an AEA lower than 35° when loaded with 1 kg(see Figure 15). If the upper arm was positioned in the dark gray zone, the IMP exceeded 40 mmHg in at least one of the investigated muscles.

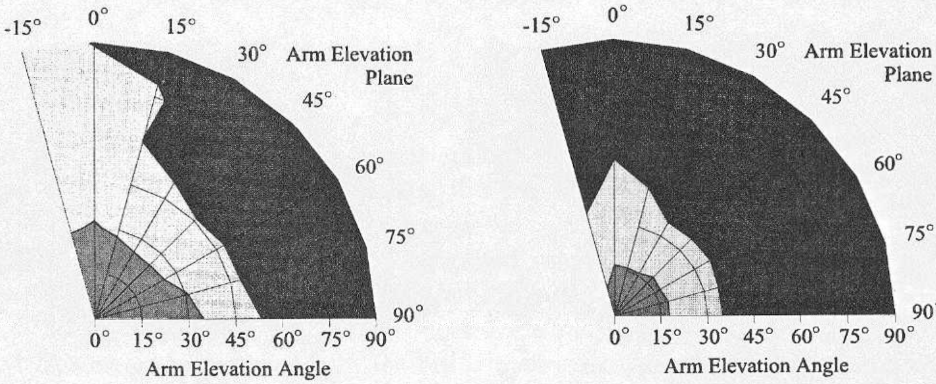

Figure 15. Zones identified from the intramuscular pressure in the infraspinatus and supraspinatus muscles for eight AEPs and seven AEAs. Arm elevations was performed with a straight arm and no hand load (left diagram) or 1 kg of hand load (right diagram). Medium gray corresponds to an IMP of 0-20 mmHg, light gray to 20-40 mmHg and dark gray to an IMP exceeding 40 mmHg.

A significant systematic difference was established between the IMP means for the AEA ($p<0.01$) as well as for the AEP ($p<0.01$), in all combinations of muscles and hand loads. The major part of the variance was explained by the AEA, but the variance explained by the subjects was larger than the variance due to the AEP, concerning the IMP in the supraspinatus muscle. Concerning the variance of the IMP in the infraspinatus muscle, the dominating factor of explanation was the subject, followed by the AEA and the AEP, in that order.

Conclusions

Accumulated muscle fatigue might occur in the infra- and/or supraspinatus muscle, if the upper arm exceeds 30° of arm elevation in prolonged static or repetitive working postures, comprising arm elevation with a straight elbow and no hand load.

Accumulated muscle fatigue might occur in the infra- and/or supraspinatus muscle, if the upper arm exceeds 15° of arm elevation in prolonged static or repetitive working postures, comprising arm elevation with a straight elbow and hand load of 1 kg.

Muscle blood flow may be impaired in the infra- and/or supraspinatus muscle if the upper arm exceeds 50° of arm elevation in prolonged static or repetitive working postures, comprising arm elevation with a straight elbow and no hand load.

Muscle blood flow may be impaired in the infra- and/or supraspinatus muscle if the upper arm exceeds 35° of arm elevation in prolonged static or repetitive working postures, comprising arm elevation with a straight elbow and hand load of 1 kg.

In working situations with prolonged static or repetitive working postures, arm flexion should be preferred to arm abduction.

The IMP in the supraspinatus was influenced more by the individual factor than the factor of the arm elevation plane.

The IMP in the infraspinatus was influenced more by the individual factor than the factors of arm elevation angle and arm elevation plane.

Paper VI

Aims

- to contribute to the validation of the shoulder model using EMG recordings from ten relevant shoulder muscles.

Material and Methods

EMG data from two experimental studies were used for the validation for the shoulder model.

Subjects

Six subjects from the first study and 11 from the second were included in the validation of the shoulder model. Mean ages, mean height and mean weight, see the corresponding paragraphs in the summary of Paper III and Paper IV Anthropometrical data was measured individually for all the subjects.

Muscles

Five shoulder muscles were investigated in the first study: the supraspinatus muscle, the upper part of the infraspinatus muscle, the anterior and middle parts of the deltoid muscle and the descending part of the trapezius muscle. In the second study, seven shoulder muscles were investigated: the anterior part of the deltoid muscle, the levator scapulae, the major and minor parts of the rhomboid muscle, the middle part of the serratus anterior muscle and the descending and transverse parts of the trapezius muscle.

Electrodes

Acquisition of the EMG activity from the superficial muscles was accomplished by bipolar surface electrodes and from the deep lying muscles by means of in-dwelling single wire electrodes. The reference electrode was an Ag/AgCl surface electrode, taped at the processus prominens (C7).

Arm positions

In the first study, the EMG activity was recorded with the arm positioned in 30° of humeral abduction and with 90° of elbow flexion. Arm positions used in the second study, see the corresponding paragraphs in the summary of Paper IV.

Experimental procedure

See the corresponding paragraphs in the summary of Paper III and Paper IV.

Signal processing

See the corresponding paragraphs in the summary of Paper III and Paper IV.

Calculations
See the corresponding paragraphs in the summary of Paper IV.

The comparison of the predicted and measured muscle activity
The muscle force of each muscle investigated was calculated for each combination of shoulder muscle, arm position and level of voluntarily reduced activity of the descending part of the trapezius muscle. The number of levels of voluntary reduced activity of the descending part of trapezius muscle was restricted to five: 10%, 20%, 40%, 60% and 80% of normal activity for each of the six arm positions studied. The relative change in force (F^R_{mpt}) was calculated for each combination of muscle, arm position and level of voluntary reduced force of the descending part of trapezius muscle, corresponding to the measured values.

$$F^R_{mpr} = \frac{F_{mpr}}{F_{mp}}$$

where **m** indicates the muscle, **p** the arm position and **r** the level of voluntary reduced activity in the descending part of trapezius muscle. F_{mp} stands for the normal activity (=100%) in the descending part of trapezius muscle.

The relative change in EMG activity (EMG^R_{mps}) measured for each muscle, arm position and subject was compared to the corresponding relative change in force (F^R_{mpt}) calculated.

$$K_{mpr} = \frac{EMG^R_{mps} - F^R_{mpr}}{F^R_{mpr}}$$

where K_{mpr} stands for the relative difference between the relative change in muscle activity estimated from model calculations and the relative change in muscle activity estimated from EMG measurements.

Results
The mean of the relative change in EMG values and the mean of the relative change in calculated values for each muscle and arm position showed discrepancies as well as conformities. EMG measurements indicate that the *supraspinatus muscle* activity was not significantly affected by the reduction in the descending part of the trapezius muscle. The muscle activity calculated by the shoulder model however predicted a reduction of approximately 60% of the initial activity. In the *upper part of the infraspinatus muscle* as well as in the *medial part of the deltoid muscle*, EMG and model calculations agreed upon a minor increase. The *anterior part of the deltoid muscle* in the first study, also indicates an increased in muscle activity from the measurements as well as from

the calculations, although calculations predicted a larger increase. There was a fairly good conformity between model calculations and EMG measurements in the second study, except for the *levator scapulae.* In all arm positions the model predicted an increase of approximately 200–300% due to reduced activity in the descending part of the trapezius. EMG measurement indicated only minor changes. The *transverse part of the trapezius* increased to 150-200% approximately in all arm positions according to EMG measurements as well as model calculations. *Serratus anterior* showed an insignificant rise indicated from both methods except in 30° of arm elevation, where the model suggested a minor reduction and the EMG a minor increase. In the *rhomboid minor* muscle there was an acceptable agreement between EMG and model values, especially for the larger arm elevation angles. The *rhomboid major* muscle had the largest EMG changes of all shoulder muscles measured, ranging between 200% and 350 %. Model calculations also indicated large changes (300-500%), except in the middle position, where no significant changes at all were are found. There was a conspicuous spread of calculated individual values for the rhomboid major muscle, especially for 90° of arm elevation.

The relative difference (K) between the calculated changes in muscle force and the measured changes in EMG activity for all arm positions and muscles are given in Tables 11 and 12. The most conspicuous result concerns the *levator scapulae* and the *supraspinatus* muscles. The mean muscle activity estimated from EMG measurements indicates only half the activity compared to model calculations for the levator scapulae (good precision but poor accuracy). On the contrary, the EMG measurements of the supraspinatus indicate double the activity compared to the calculated muscle activity. The best concordance between model simulation and EMG measurement was obtained for the *anterior part of the deltoid* muscle, resulting in a relative difference of -28% in the first study and -16% in the second. Also the *transversal part of the trapezius* muscle (28%), the *anterior part of the serratus* muscle (26%), the *middle part of the deltoid* medialis (-13%) and the *upper part of the infraspinatus* (25%) showed an acceptable correspondence between the EMG activities and the model predictions, although the relative difference varied substantially between arm positions for the anterior part of the serratus. *The rhomboid minor* and *major* muscles showed the largest variance between arm positions although the relative difference was moderate (30% and 43 %, respectively).

Table 11. The relative difference (%) between the estimated muscle activity calculated by the shoulder model and the measure EMG activity.

	supraspinatus	*infraspinatus upper part*	*deltoid anterior*	*deltoid medialis*	*mean*	*std dev*
A30F90	207	25	-28	-13	42	95

Table 12. Relative difference (%) between the estimated muscle activity calculated by the shoulder model and the measure EMG activity.

	deltoid anterior	*levator scapulae*	*rhomboid major*	*rhomboid minor*	*serratus medialis*	*trapezius transversus*	*mean*	*std dev*
A30F0	-38	-58	19	14	55	45	6	45
A60F0	-28	-59	185	104	-4	43	40	91
A90F0	12	-40	9	48	0	30	10	30
A30F90	-28	-46	-23	-18	98	10	-1	52
A60F90	-28	-68	86	18	-11	10	1	52
A90F90	12	-49	-19	15	15	31	1	29
Mean	-16	-53	43	30	26	28	10	
std dev	22	10	80	42	43	16		

Conclusions

- Shoulder muscle forces estimated by the biomechanical shoulder model acceptably agree with registered EMG activity, except for the levator scapulae and the supraspinatus.
- The closest correlation was found for the middle and anterior part of the deltoid, the upper part of the infraspinatus and the middle part of the serratus anterior.
- The biomechanical shoulder model gives valuable information on the force contribution in the human shoulder.
- The biomechanical shoulder model can probably be improved by introducing a force-resisting component for non-active muscles.

GENERAL DISCUSSION

There is an urgent need for hints and directions on how to evaluate risks for developing chronic work-related shoulder disorders, as well as for methods of assessing hazardous working conditions. To date, efforts to prevent and treat work-related shoulder disorders have not been particularly successful. It is therefore extremely important to arrive at scientifically based recommendations on work place design and work organization, in order to prevent the development of chronic work-related shoulder disorders. The current knowledge of underlying causes for developing work-related shoulder disorders, is in several aspects insufficient. This clearly emerges from several doctoral theses published during the last decade (Järvholm 1990, Dimberg 1991, Hägg 1991, Jensen 1991, Takala 1991, Öberg 1992, Mathiassen 1993, Sporrong 1997). The basis for medical and ergonomic actions exposes the fact that there remain several unexplored regions. It is the duty of science to fill in those gaps of knowledge, and in the present work several methods were used to obtain this goal.

Electromyography

In this work, fine wire electrodes were used extensively. The techniques applied are well established. However, in Paper I EMG recordings were performed simultaneously with the pressure recordings and at the same location in the muscle. This method was developed by Järvholm and co-workers. (1989). Here a bipolar wire electrode was introduced into the muscle with the same Vasculon® cannula as the pressure catheter. EMG recordings with bipolar electrodes suffer from the disadvantage that the actual interelectrode distance can not be controlled and will remain unknown unless an X-ray or some equivalent technique is used. By electrical stimulation through the wire electrode valuable information can be achieved on the electrode tip location (Kadaba et al., 1992). The two electrodes were bent into hooks of different lengths, to prevent short-circuiting of the deinsulated tips. During the experiment, migration of the electrodes could occur due to muscle contractions and arm movements (Jonsson, 1970; Krivickas et al., 1996), and the change in interelectrode distance would influence the signal pickup properties like band-width and spatial resolution in an unpredictable way. This can however be avoided by using single wire electrodes in combination with a surface reference electrode (Kadefors et al., 1976). A single wire technique also makes the uptake volume larger in a predictable way (Kadefors and Herberts, 1976). The reason, however, for using bipolar electrodes was to have a spatial resolution of the EMG, similar to the pressure generating muscle fibers in the vicinity of the catheter.

In surface electromyography, the placement of the electrodes is crucial for the result. EMG recordings from the trapezius muscle are a good illustration of this. Jensen and co-workers (1993) investigated different pick-up areas for the descending part of the trapezius muscle. They found, by using an array of electrodes placed across the shoulder, that the EMG amplitude varied along the acromion-vertebra line. A depression of the EMG amplitude was localized to the pick up area recommended by Basmajian (1983). He assumed that this area coincides with the innervation zone of the muscle. This aligns with the observations by Lindström (1974), who showed that for reasons of symmetry a dip in signal amplitude is found at the innervation zone. Jensen and co-workers recommended an electrode placement 2 cm lateral to the former recommendation. The EMG studies in Papers I, III and IV were, however, performed according to the original recommendations by Basmajian, since these later findings were not known at the time of measurement.

The techniques applied for EMG signal acquisition in the present study have worked well and made available signals suitable for quantitative analysis. Monitoring of the levator scapulae muscle implies however a methodological challenge. The levator scapulae is situated in the posterior part of the neck. It originates from the transverse processes of the upper three or four cervical vertebra and attaches to the medial margin of the scapulae between the superior angle and the base of the spine. Its lower part is completely covered by the descending part of the trapezius muscle. Both muscles are thin (approximately 1 cm) and there is a demand for spatial comprehension and surgical talent to place the fine wire electrode in the levator scapulae. Besides the difficulties of placing the fine wire electrode in the levator scapulae muscle, cross-talk from the trapezius is another aggravating circumstance.

The amplitude of the EMG signal depends on several factors e.g. type of electrode, electrode location, thickness of subcutaneous fat, etc. To make it possible to quantitatively compare different muscles and different subjects a calibration of the EMG signal against a known reference is necessary (normalization). A frequently used reference value is the EMG signal at maximal voluntary contraction (MVC) in primary functional activity for the muscle under study (or at least maximal voluntary torque in the joint across which the muscle is acting) (Schüldt K et al. 1986, Hagberg & Hagberg, 1989, Wærsted et al., 1991). This reference value is however subjected to substantial unreliability, due to motivational and physiological reasons. Many investigators have used submaximal contraction as the normalizing reference values (Perey and Bekey, 1981; Yang and Winter, 1983; Lundberg et al. 1994). The reference value is obtained by performing a standardized arm position. To increase the reliability of the reference, a series of submaximal contractions are recommended. In the study of IMP and EMG in four shoulder muscles (Paper I), mean value normalization was performed according to a method used by other investigators (Sigholm et al., 1984; Järvholm et al., 1989). This method implies that each RMS value was divided by average of all RMS values obtained from the specific muscle and subject. The reference value obtained from this method is less sensitive to single artifacts and the influence of different muscle length has been compensated for. In a review on normalization and surface EMG, Mathiassen and co-workers (Mathiassen et al., 1995) point out that there are a variety of methods for normalizing EMG amplitude from the upper trapezius. They purpose a standard terminology relating to normalization of EMG and suggest a procedure for normalization of EMG from the descending part of the trapezius muscle.

Intramuscular pressure

IMP acquisition was performed in static arm position or during slow arm elevation. The latter case involved registration during a slow and constant speed arm elevation of approximately 10°/s. This method was adopted to facilitate the registration of IMP in a large number of positions, to sufficiently describe the IMP in the whole sector of functional arm positions. Using the former technique, this would have been very time consuming. The outcome of these registrations compared to other studies (Järvholm et al., 1989; Järvholm et al., 1991; Jensen et al., 1995a), indicates a 50-60% lower level of IMP for all arm positions except in the start position for infraspinatus as well as for supraspinatus. Individual variations, the composition of subjects regarding gender, catheter location etc. are plausible contributing factors behind the discrepancy. Insufficient dynamic properties of the measuring device might also end in underestimation of the pressure. However Styf and Körner (25) investigated the dynamic properties of the microcapillary infusion (MCI) technique and estimated the rise time to 35 - 70 ms. Since arm elevation was performed very slowly ($\approx$10°/s), the change of IMP versus. time (< 20mmHg/s) would be slow enough to give a genuine registration, reflection basic properties of skeletal muscle. The semidynamic measuring technique could also play a role. According to Komi and Buskirk (1972) eccentric and concentric muscle activity influences the muscle force and accordingly also the IMP. In this study only concentric muscle contractions were performed, which results in lower IMP levels compared with static or eccentric work. This is supported by recent laboratory studies (Sporrong and Styf, unpublished results).

Posture recordings

When recording postures and movements with optical non-contact 3D motion analysis systems, the human body is regarded as a system of rigid segments linked together by joints. Each segment studied is equipped with marker(s). The MacReflex system, used in two studies (Papers II and V), comprises passive reflecting markers. This implies advantages like lightweightness, easily applicable and no cables, compared to, e.g. the Selspot 3D motion analysis system. A draw- back however, is that the identification of the markers is left to the investigator, since the cameras cannot distinguish the markers. In systems with active markers (e.g. the Selspot system), an automatic identification is feasible with, for instance, time-multiplexing. A methodological problem, common to all optical movement analysis systems, is the occurrence of the temporarily hidden or coinciding markers. This could be dealt with by increasing the number of cameras, enough to cover all spaces with at least two cameras at the same time. Another way to deal with this problem, used here, is to spread the markers away from each other and from the body segments. This is accomplished by attaching the markers to thin bars sticking out from an exoskeleton. In position measurements during normal locomotion, this measure would normally impair accuracy, due to increased movements of the markers relative to the corresponding segment. In the present case however, the position measurements were performed during static or semidynamic conditions, with only minor influence on the accuracy.

The location of the markers can be utilized in different ways. If only coarse information on joint angles is required, the markers may be located as close to the center of the joints as possible. This technique produces position data, sufficient for calculations of position of body segments, joint angles, angle velocity etc. This data will, however, be insufficient for an accurate and complete determination of the segment location and orientation. In the studies concerning position measurements on the upper extremities (Papers II and V), the ambition was not only to measure the location of the torso, the upper arm and the forearm, but also their rotation. This implies that three markers are recorded for each body segment. A drawback with this procedure is, however, the complexity of the calculations.

Shoulder model

A validation of the biomechanical shoulder model was performed by comparing muscle forces predicted by the model, with muscle activity estimated from EMG recordings in ten relevant shoulder muscles (Paper VI). The biomechanical shoulder model was found to predict shoulder muscle forces with an acceptable accuracy except for the levator scapulae and the supraspinatus. The supraspinatus muscle was underestimated compared to the EMG measurements and the levator scapulae was overestimated. There are several plausible reasons for the discrepancies. Insufficiencies in the modelling of the shoulder can depend on the algorithm or the available data on antropometry, anatomy, physiology etc. One imperfection discovered in the study, was the necessity to introduce a force-resisting component for tonus in non-active muscles. A modification of the geometric and kinematic modeling of the acromioclavicular joint has also been suggested. However, it must be remembered, that the EMG recordings only reflect the exerted muscle forces during restricted conditions. The biomechanical shoulder model constitutes a valuable tool for accessing the intrinsic distribution of forces from an external torque on the shoulder, although modifications are necessary.

Muscular synergy
From the study on voluntary reduction of muscle activity (Paper III), it was found that there is an ability reduce the muscle activity in the descending part of the trapezius. This implies that feedback training, in order to change the muscle recruitment pattern in the shoulder, could be successful and favorable for the descending part of the trapezius muscle, in cases of overuse syndrome in the trapezius muscle. However, the load from the trapezius muscle is transferred to other muscles in the shoulder, particularly the rhomboid major and minor muscles and the transverse part of the trapezius muscle, as demonstrated in the study on consequences of trapezius relaxation on the distribution of shoulder muscle forces (Paper IV). The relative tolerance to excessive use among the shoulder muscles is not fully clear. This means that a reduction of the descending part of the trapezius muscle assisted with feedback technique, must be done only with great caution. Pain from the area corresponding to the location of the rhomboid muscles has been reported from physiotherapists as a result of trapezius feedback therapy.

Ergonomic recommendations
In the study on IMP in the infra- and supraspinatus muscles (Paper V) it was found that the limit for non-recovery (NR) was exceeded when the AEA was larger than 30° and with a hand load of 1 kg, the limit for NR was exceeded when the AEA was larger than 15°. It was also found that limit for blood flow impairment (BFI) was exceeded when the AEA was larger than 50° and with a hand load of 1 kg, the limit for BFI was exceeded when the AEA was larger than 35°. The results are only applicable for arm elevation in prolonged static or repetitive working postures with a straight arm. These findings imply a more restricted tolerance to working situations with frequent or long lasting situation with elevated arms.

In the formulations of environmental laws and regulations, recommendations and guidelines, there are aspirations to present explicit limits. For instance in the field of ergonomics concerning musculoskeletal load, there is an indulgence in expectations on "safe zones". Biomechanical calculations from externally observable parameters, e.g. weight of load, lifting distance, applied force, etc., have been applied particularly for the lower back. Chaffin and colleagues (1984) developed a computer-aided model, 3D SSPP (3D Static Strength Prediction Program™) for the analysis of joint torque and back compression. From data on working posture and exerted forces, the program predicts back compression and joint torque. The SSPP also produces an interpretation of the results in terms of estimated fraction of men and women with the capability to perform the task. SSPP also estimates whether the maximal permissible load on the spinal column is exceeded. This limit value was based on biomechanical studies on intravertebral discs.

For more specific risk assessment on an individual basis, knowledge of the actual exposure and the relation between exposure and the etiology of work-related diseases is necessary. The etiology of WMSDs is only vaguely known and several biomechanical, physiological, immuniological and other hypotheses have been proposed. Theories have be suggested on the mechanical, metabolic and biochemical responses. For performing risk assessments from exposure to external factors, it is also necessary to know the influence of the structures effected. Different observation methods like OWAS, REBA, VIRA, RULA ect. are available. External factors such as working posture, speed of work, load of tools and material handled, force applied, length of working cycle, etc., are collected and harmful influences are assessed. To be successful it is however crucial that the chosen limits are based on dose-effect relations.

Comparison with the NIOSH Guidelines
In a comparison between the recommendation in Paper V and the NIOSH guidelines, discrepancies as well as similarities are found. Calculations from the NIOSH lifting equation show that the recommended weight limit (RWL) is high for small arm elevation angles (AEAs) and decreases for increasing arm elevation. This applies for all arm elevation planes (AEPs), the general magnitude decreases for the RWL, when going from flexion to abduction (asymmetrical lifting). This roughly agrees with the general characteristic of the IMP levels presented in Paper V, although our results advocate a more conservative and restricting attitude concerning load exposure to the upper extremities. For example, an arm position of 30° of arm elevation in the sagittal plane results in a RWL of 17 kg (two hands), when calculated from the NIOSH equation. In this arm position the IMP in the infra- and supraspinatus muscles exceed the level of non-recovery already at 1 kg of hand load. With an arm elevation of 45° in an elevation plane of 30° from the sagittal plane, a calculation results in a RWL of 12.5 kg (two hands). The IMP in supraspinatus muscle however, exceeds the level of blood flow impairment in this position already with a hand load of 1 kg. This means that the NIOSH Guidelines, although highly credible in protecting from low back pain in manual lifting, are clearly insufficient in preventing shoulder muscle overload in work with elevated arms.

Table 13. A summary of restrictions and applications between the method described in Paper V and the NIOSH guidelines for manual lifting.

	Method of Paper V	NIOSH guidelines for manual lifting
Body posture	Sitting	Standing
Risk criterion	Musculoskeletal disorders of the shoulder	Biomechanical, Physiological and Psychophysical
Number of upper extremities	1	2
Character of effort	Static or semistatic	Dynamic
Limiting factor	IMP in infraspinatus and supraspinatus	Spinal disc compression, Aerobic lifting capacity and Perceived acceptable limit

Comparison with the RULA method

McAtamney and Corlett (1993) developed the RULA (Rapid Upper Limb Assessment) method. The method is intended for assessing working postures and muscle forces for the back, neck, shoulder and upper extremities and it supports demands on occupational health and safety stated by the EU directives (90/270/EEC).

In the RULA instruction manual, a selection of scores are listed for each body segment. The magnitude of the scores indicating the level of postural or force strain to that particular body section and the total sum of scores indicates the risk of injury. For the upper extremities the scores depend on arm elevation angle, elevation plane and characteristics of movements (see Table 14).

The assessment of working postures and hand loads with the RULA method, concerning the upper extremities, coincide in several aspects with the results in paper V. The RULA method reflects the increasing risk of working postures with elevated arms. It also considers the elevated risk of shoulder disorders from shoulder abduction. Prolonged static postures and repetitive movements are also observed. However, the discrepancies and conformities between the RULA method and the results in paper V, concerning the level of risk assessment for working postures and muscle loads on the whole, are difficult to estimate.

Table 14. The scoring system for the upper extremities according to RULA method for work task assessment.

Arm elevation angle	Flexion <2kg	Add if abduction	Add if static	Add if repetitive
0°-20°	1	+1	+1	+1
20°-45°	2	+1	+1	+1
45°-90°	3	+1	+1	+1
>90°	4	+1	+1	+1

Comparison with the OWAS method

OWAS (Ovako Working Posture Analyzing System) is a method for evaluation of postural load during work, based on a simple and systematic classification of work postures combined with observations of work tasks. Information extracted from an analysis with the OWAS method can be the level of strain on each body part and the degree of urgency pro body segment to change these conditions.

The OWAS method considers the elevated risk of shoulder disorders from working situation with elevated arms and hand loads. However, the assessment of shoulder muscle load follows a coarse classification, with only three levels of postural load and three levels of external load (see Tables 15 and 16). With the OWAS method, it is not possible to discriminate between acceptable and unacceptable working conditions concerning the shoulder load, considering the results from Paper V. This conclusion is however only valid in prolonged static or repetitive working postures.

Table 15. The scoring system concerning hand load or requisite force according to the OWAS method for work task assessment.

Hand load or requisite force	**Score**
≤ 10 kg (100N)	1
>10 kg (100N) but ≤ 20 kg (200)	2
> 20 kg (200N)	3

Table 16. The scoring system concerning arm posture according to the OWAS method for work task assessment.

Arm positions	Elevation angle	Score
Upper arms along the side of the body	0°/0°	1
One elevated upper arm at or above shoulder height	≥90°/0°	2
Elevated upper arms below shoulder height	<90°/<90°	3
Elevated upper arms at or above shoulder height	≥90°/≥90°	4

Future Research

Despite of the intensified high quality research in the field of biomechanics and ergonomics during the last decades, there is still insufficient knowledge on e.g. the muscle recruitment pattern of the shoulder and its temporal and individual variation, the intramuscular variation of activity, the importance of muscular rest during occupational and leisure activities, the exposure-effect relationship in occupations with high prevalence of work-related musculoskeletal disorders, etc. To understand the natural history of WMSDs of the shoulder and to improve the basis for prevention, future basic research is suggested in to the following items.

In the contemporary research on muscle activity, the muscle is often regarded as a homogenous functional entity. The existence of autonomous functional units within a muscle should be investigated. This knowledge is important in the modelling of the shoulder, as well as in estimating the validity of EMG measurements on large muscles, e.g. the trapezius muscle.

The recruitment pattern of motor units and its dependence on muscle contraction level and duration, muscle fatigue and mental stress is incompletely known. Also the variations of recruitment pattern due to muscle type, gender, age and physical training need to be elucidated.

The variation of IMP, EMG activity and blood perfusion (BF) within different parts of the muscle is incompletely known. Studies on the IMP, EMG and BF in different shoulder muscles are needed, to elucidate temporal and spatial diversities.

Working tasks, characterized by long periods of arm elevation, are known to generate shoulder disorders. There is a lack of quantitative information on exposure to arm elevation in occupations with high prevalence of shoulder disorders. Studies of working postures with objective methods in some of these occupations are suggested.

The biomechanical shoulder model constitutes a valuable tool for accessing the intrinsic distribution of forces from an external torque on the shoulder. To improve model estimations, more data on force-resisting components and tonus in non-active muscles would increase the accuracy of the model.

CONCLUSIONS

The studies relating to the specific aims, questions and hypotheses addressed in this thesis may be concluded as follows:

1. IMP measurements were shown to produce reliable estimates of isometric muscle load in the shoulder. IMP produces a similar description of the load as EMG measurements do. Substantial differences were found between the supraspinatus, the infraspinatus, the deltoid and the trapezius muscles regarding their pressure generating characteristics. The character of the IMP variation due to arm position and hand load corresponds reasonably well to the total shoulder torque for all four muscles. The supra- and infraspinatus muscles were, however, found to generate high IMP especially in shoulder abduction, as compared to the trapezius and deltoid muscles. These findings support the hypothesis that bulky muscles within a low compliance compartment will generate higher IMPs during contraction compared to superficial and flat muscles. This might explain the differences in endurance and vulnerability between muscles during prolonged muscle contraction of equal fraction of MVC. These findings also indicate that prolonged work with elevated arms will impede muscle blood flow in the stabilizing muscles of the rotator cuff, which might explain the findings of EMG-detected localized muscle fatigue in these muscles.

2. The MacReflex motion analysis system represents a valuable contribution to the existing small group of methods for non-contact posture measurements in the fields of biomechanics and ergonomics. The intrinsic errors originating from the manufacturing and the rigidity in the calibration fixture were found to be consistent and essentially negligible. Also, the alignment of the calibration fixture to the chosen laboratory coordinate system does not seem to introduce significant error. A maximal deviation of 1.8° from the expected value was obtained, when testing the ability to accurately position and reposition a mechanical structure. In most ergonomic applications an error in this range is acceptable. Although simple and coarse methods were used for the positioning when measuring on a subject, the angles obtained were fairly constant. In conclusion, the MacReflex system determines position with adequate accuracy in normal handling.

3. From the investigation on the ability to reduce muscle activity in the shoulder assisted by a feedback technique, it was concluded that the muscle activity of the descending part of the trapezius could be voluntarily reduced to almost half the initial activity, without changing the hand load or the arm position. This ability could not be demonstrated in the supraspinatus, the infraspinatus, the anterior deltoid or the middle part of the deltoid muscles. As a

consequence of this ability concerning the descending part of the trapezius muscle, it was also concluded that the descending part of the trapezius muscle is not a reliable estimator of the total shoulder load.

4. From the extended study on the consequences of trapezius relaxation on the distribution of the shoulder muscle forces, it was concluded that the load was distributed to other synergistic shoulder muscles, when the descending part of the trapezius muscle was relaxed. The main part of the load was transferred to the rhomboid major and minor and the transverse part of the trapezius, but the levator scapulae did not adopt load as predicted by the biomechanical shoulder model. This implies that special attention should be paid to the rhomboid muscles and the transverse part of the trapezius muscle, when introducing feedback techniques to relax the descending part of the trapezius muscle.

5. From the study on the identification of risk zones in work engaging the upper extremities, it was concluded that depending on the magnitude of the intramuscular pressure in the infra- and supraspinatus muscles, accumulated muscle fatigue might occur in the infra- and/or supraspinatus if the upper arm exceeds 30° of arm elevation without hand load. With a hand load of 1 kg, accumulated muscle fatigue might occur already when the upper arm exceeds 15° of arm elevation. It was also concluded that impaired muscle blood flow might occur in the infra- and/or supraspinatus if the upper arm exceeds 50° of arm elevation, without hand load. With a hand load of 1 kg, ichemia might occur already when the upper arm exceeds 35° of arm elevation. The risk for accumulated muscle fatigue and impaired muscle blood flow concerns prolonged static or repetitive working postures with a straight elbow. Shoulder flexion should be preferred to shoulder abduction, because of a more favorable pressure development in shoulder flexion. From the comprehensive survey of intramuscular pressure in the infra- and supraspinatus muscles, it was concluded that the IMP of the supraspinatus muscle was more influenced by the individual factor than the factor of the arm elevation plane, and that the intramuscular pressure of the infraspinatus was more influenced by the individual factor than the factors of arm elevation angle and arm elevation plane.

6. From the study on the estimation of the load-sharing pattern in the shoulder and the comparison between electromyographic measurements and model calculations, it was concluded that the shoulder muscle forces estimated by the biomechanical shoulder model acceptably agree with the shoulder muscle forces estimated by the measured EMG activity. The model calculations overestimated the levator scapulae and underestimated the supraspinatus

muscles compared to the EMG measurements. The closest correlation was found for the middle and anterior part of the deltoid, the upper part of the infraspinatus and the middle part of the serratus anterior. Concerning the biomechanical shoulder model, it can be concluded that the model produces valuable information on the force contribution in the human shoulder, but improvements should be made, for instance by introducing a force-resisting component for nonactive muscles.

ACKNOWLEDGEMENTS

I wish to express my sincere appreciation and gratitude to everyone who made this work possible. I especially wish to thank:

my tutor professor **Peter Herberts,** for his neverending support and guidance throughout the years, and for his gentle attitude towards me even when I did not always deserve it.

my tutor professor **Roland Kadefors,** for his patient and inspiring supervision during all stages of my scientific education, for introducing me to the interdisciplinary field of ergonomics, for his constructive criticism, and his never failing intuition of what's the right thing to do.

Professor **Björn Rydevik,** head of the Department of Orthopaedics, Sahlgrenska University Hospital for accepting me without hesitation as a PhD student at the Department of Orthopaedics.

Roland Nilsson, manager of Lindholmen Utveckling, for providing stimulating working facilities, and for sharing his own experiences on being an engineer of electronics in the field of medical science.

Håkan Sporrong, orthopaedic surgeon, research college, co-author, and appreciated friend, for many memorable occasions on conferences as well as in the laboratory and for his everlasting optimism.

Ulf Järvholm, co-author and good friend who tragically died.

Christian Almström, my long time friend and former head of the research department at Lindholmen Utveckling for introducing me into the field of EMG and computers.

Jorma Styf for his everlasting support and generosity.

Christian Högfors, Dan Karlsson, Mohsen Maksous and Bo Peterson, co-authors and affiliation at the former Centre of Biomechanics, Chalmers University of Technology, for their valuable contribution in the discipline of biomechanics and their contribution to the interdisciplinary discussions.

Michael Forsman, my friend and co-author, for his expert assistance in data computing and graphic visualization.

Leif Sandsjö, a long time friend and colleague, for his straight answers and for being my patient travelling companion.

Allison Kaigle, for proofreading and for enlightening discussions on the linguistic curiosities of the English language.

Klas Hjelmgren and **Per Hörfelt**, Center for Mathematics and Statistics, Chalmers University of Technology, for artful support on statistical matters.

All the **staff members** from Sahlgrenska University Hospital/Östra, **students** at Chalmers University of Technology and **athletes** of Partille Floor Hockey Club, who very bravely and without hesitation volunteered to participate in my sometimes painful and time-consuming experiments.

All my **colleges** and **friends** at Lindholmen Utveckling, for their friendship and inspiring discussion on most subjects.

My **parents,** for their uncompromising belief in me.

My **family**, Lisbeth, Lina, Lotten and Linnea, for love, support, understanding and for being my root in reality.

The **Swedish Council for Work Life Research**, the former Swedish Work Environment Fund, for many years of economic support.

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