THE SHOULDER MECHANISM
a dynamic approach

Frans C.T. van der Helm

doctoral thesis

Man-Machine Systems Group
Lab. for Measurement and Control
Dept. of Mechanical Engineering and Marine Technology
Delft University of Technology
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Frans C.T. van der Helm,
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Chapter 1

Introduction

I Historical perspective

In the last two decades rehabilitation research has been a major topic in the research program of the Man-Machine Systems Group of the Laboratory for Measurement and Control at the Delft University of Technology (Stassen, 1989). In cooperation with, amongst others, the rehabilitation center 'De Hoogstraat' in Utrecht, research has been focussed on the development of communication aids and on the rehabilitation of patients with injuries of the upper extremities. For the latter group of patients prosthetic and orthotic devices were developed (Cool, 1989). Another research item was a system-theoretical approach of the rehabilitation process in which the closed-loop interaction between the patient and the treatment team was modelled in order to predict the outcome of the rehabilitation process. After applying this approach successfully to patients with a spinal cord lesion (Stassen et al., 1980), the rehabilitation of patients with a brachial plexus lesion was modelled (Jaspers, 1990). The brachial plexus is a nerve network in the neck region. Due to a brachial plexus lesion muscles of the upper extremity will be paralysed. If after two years of rehabilitation and, when applicable, attempts of neurosurgical repair, the muscles of the gleno-humeral (shoulder) joint are still paralysed, sometimes the patients undergo an operation to arthrode the joint. Due to this operation the humerus (upper arm bone) is rigidly fused to the scapula (shoulder blade) after which this combination acts as one bone. Hence, if the motions of the shoulder girdle which consists of scapula and clavica (collarbone), can be described, and if the desired motion range of the humerus is defined, the optimal fusion angles between scapula and humerus can be calculated. Reliable predictions of the function restoration which can be achieved, could help the patient to decide whether or not to undergo this drastic and irreversible surgery. About ten years ago, this was the reason to start a research project on the motion behavior of the shoulder girdle.

Pronk (1991) explored problems encountered when studying the shoulder girdle. From an extensive literature search it was concluded that very little data were available to quantify the motions of the shoulder girdle. A motion recording method was developed in which bony landmarks were palpated and subsequently digitized (Pronk, 1987). So, the three-dimensional (3D) position of the bones could be reconstructed. Following motion recording, the description of the 3D motions was undertaken. The development of a kinematic model of the shoulder girdle was initiated, in order to predict hand positions from recorded motions of the scapula (Pronk, 1989). In addition, arm and scapular motions of 18 patients were recorded for analysis of the motions which remained after the
brachial plexus injury and glenohumeral arthrodesis. Meanwhile, the original reason for the research project was partly outdated by the development of an externally fixating device for the glenohumeral arthrodesis (Nieuwenhuis & Pronk, 1989). Using this device, the optimal fusion angles can be assessed by experimenting in the patient’s own environment, before the actual arthrodizing operation takes place.

As a result of the glenohumeral arthrodesis, the moment and force balance around the shoulder mechanism joints will change. The thoracoscapular muscles must exert all necessary forces to move the arm. Using the external fixator, patients are restrained to try the whole achievable motion range. The altered muscle functions and subsequently changed motion range and ability to exert forces, were still a research item. Therefore, the next stage in the shoulder research project which is presented in this thesis, is the analysis of the dynamic behavior of the shoulder mechanism.

II Shoulder mechanism
Research in the shoulder region has mainly focussed on the glenohumeral joint. Therefore, due to the lack of official definitions, the terms 'shoulder' and 'shoulder joint' are commonly used to denote the glenohumeral joint. The shoulder girdle consists of the clavicula and scapula of one side of the body. In contrast to the pelvic girdle, the circle is not complete. Motions of the shoulder girdle are closely related to motions of the humerus. Therefore, it seems logical to bring the humerus into the research of the shoulder girdle. The scapula is not only connected to the thorax (chest) via the clavicula, but also slides at the backside over the thorax. Motions of the scapula are determined by the shape of the thorax. The constellation of bones, i.e. thorax, clavicula, scapula and humerus, is defined as the so-called shoulder mechanism (Fig. 1).

Figure 1: Bony structures of the shoulder mechanism.
The clavícula is connected to the thorax by the sternoclavicular (SC-) joint, whereas the scapula in turn is connected to the clavícula by the acromioclavicular (AC-) joint. In addition, the medial border of the scapula slides over the dorsal side of the thorax at the so-called scapulothoracic gliding plane. Hence, motions of the scapula are severely constrained. On the one hand, the acromion (shoulder top) moves more or less on a sphere around the SC-joint. On the other hand, the medial border moves over the ellipsoidal-shaped thorax, pressed onto it by the combined action of m. (musculus) serratus anterior and m. rhomboideus. The double connection between the scapula and the thorax makes the shoulder girdle a closed-chain mechanism. The humerus is connected to the scapula by the glenohumeral (GH-) joint. The shoulder girdle forms a moveable but stable base for motions of the humerus. The large motion range of the humerus is due to simultaneous motions at the SC-, AC- and GH-joint.

Furthermore, three extracapsular ligaments constrain the joint motions (Fig. 2). The costoclavicular ligament runs from the first rib to the clavícula, crossing the SC-joint. The conoid and trapezoid ligaments originate at the clavícula and insert at the processus coracoideus, a ventrally pointing bony protuberance of the scapula, just underneath the clavícula.

All motions of the joints of the shoulder mechanism are generated and controlled by a large number of muscles. Morphologically, muscles of the shoulder mechanism can be divided into three groups (Table 1): Thoracoscapular muscles running from the thorax to the shoulder girdle, scapulohumeral muscles from scapula to humerus and thoracohumeral muscles running from the thorax directly to the humerus. Figs 3A and 3B show a ventral and dorsal view of the superficial layer of muscles. Muscles of the shoulder mechanism are often fan-shaped and have large

Figure 2: Extracapsular ligaments of the shoulder mechanism.
attachment sites. Many muscles like the thoracospinal muscles are bi-articular. Thoracohumeral muscles are even tri-articular. With the scapula as intermediary, thoracospinal and scapulohumeral muscles are connected in series. For instance, it is often assumed that scapular motions serve to provide m. deltoideus optimal positioning with respect to its force-length relationship. Therefore, it is no wonder that motions of the shoulder girdle and humerus are closely related.

The large motion range of the humerus is combined with the ability to exert forces in almost any direction. Forces and moments are transmitted to the trunk via the bones and muscles of the shoulder mechanism. Since bones usually transmit compression forces and muscles only can exert traction forces, an interesting equilibrium of traction and compression forces results. In addition, at the scapulothoracic gliding plane the thorax exerts forces at the scapula.

### III Literature

Because of the complex motion constraints of thorax and clavicular, and the large number of muscles of which the lines of action are certainly not in one particular plane, analysis of the shoulder mechanism should be essentially three-dimensional. At the turn of the century, mainly German anatomists studied the shoulder by dissecting cadavers and carefully watching the course.
of the muscles (e.g. Fick & Weber, 1877; Braune & Fischer, 1888; Fick, 1911). Hypotheses about muscle functions were based on a kinematic approach. If a muscle would shorten during a bony motion starting from the anatomical position, the muscle was categorized as agonist for that motion. Mollier (1899), Shiino (1913) and Hvorslev (1927) extended this approach by building physical models of the shoulder in which muscles were replaced with cords (Fig. 4). The cords were connected via pulleys to a keyboard. Length change of the cords due to motion of the bones was quantified by the excursion of the keys. Vice versa, by pressing a key motions of the bones could be established. In fact, many contemporary theories about muscle function as stated in anatomical textbooks date back to the research done hundred years ago.

New recording methods as electromyography (EMG) and roentgenography enabled quantification of muscle activity and bony motions, respectively (e.g. Inman et al., 1944; Saha, 1961). EMG was mainly recorded on easy accessible, superficial muscles like m. trapezius, m. pectoralis major and m. deltoideus. Using roentgenograms the spatial motions of the shoulder girdle are reduced to planar projections, thereby introducing non-reconstructible distortion errors. Because of the neglect of these errors, theories based on roentgenography still hamper the development of correct theories about the motion behavior of the shoulder mechanism, e.g. by assessing imaginary joint motions (Inman et al., 1944) or by assessing the center of rotation from a projection (Poppen & Walker, 1978; Dvir & Berme, 1978). Of late, 2D biomechanical models of the glenohumeral joint were introduced (DeLuca & Forrest, 1973; Poppen & Walker, 1978; Scholten et al., 1982; Dul, 1987).

Figure 3: Superficial layer of muscles of the shoulder mechanism.
A: Ventral view.
B: Dorsal view.
Since these models only represented a few morphological structures, the contribution to complete comprehension of the mechanics of the shoulder mechanism was very limited. Very recently, some outlines of 3D biomechanical models of the shoulder mechanism were published (Wood et al., 1989ab; Karlsson et al., 1989; Karlsson, 1990). However, these studies did not present much detail on the modelling approach and hardly produced any results about the mechanical behavior.

Scope of the thesis

Very little quantitative as well as qualitative knowledge apparently exist about the mechanics of the shoulder mechanism. Aim of the research is to get insight into the mechanics of the shoulder mechanism, and especially into the function of muscles and ligaments. Therefore, in this thesis the existing kinematic model will be extended to a dynamic model. In this dynamic model, muscles and ligaments will be incorporated, in addition to external forces. Using this model, positions of the shoulder mechanism bones can be simulated. With the help of the location of the muscles and their moment arms, it can be calculated which muscles are suited to contribute to the exertion of external forces. Hence, the possible muscle function can be derived. Due to the extension of the kinematic shoulder girdle model to a dynamic model of the shoulder mechanism, possible

![Figure 4: Physical model of the shoulder (Mollier, 1899). Muscles are replaced by cords which are connected to a keyboard.](image)
applications of the model have grown beyond the glenohumeral arthrodesis. Simulation results can be applied in medical practice, e.g. to improve diagnostics through better understanding of the normal function of morphological structures and to improve surgical interventions like implementation of a glenohumeral endoprosthesis or prevention of glenohumeral dislocation. Shoulder complaints can be prevented by analysing the load on morphological structures using a model of the shoulder mechanism. In the field of ergonomics this knowledge can be used in the assessment of admissible work load and in the design of products, adapted to the human ability to exert forces.

This thesis is a compilation of eight articles which describe the development of a dynamic model of the shoulder mechanism, motion recordings of the shoulder mechanism bones and some applications of the model. Since each article had to be readable independently, some overlap was inevitable.

The model is based on the finite element method in which each gross morphological structure is represented by mechanical equivalents (elements) like beams, trusses and hinges. Since the kinematic and dynamic behavior of each element is well-known, the behavior of the whole mechanism can be calculated. Though the model is mathematical, elements of the model and input and output variables are comparable to the physical model as built by Mollier (1899) nearly hunderd years ago. Bones are represented by rigid bodies and muscles by force-generating, length deformable cords. Input variables of the model are positions and motions of the bones as well as external forces; output variables are muscle length and muscle forces (Fig. 5). The model structure is based on the representation of all mechanical relevant gross morphological structures. Model parameters describe the morphology of the system: Inertia, position coordinates of morphological

![Diagram](image)

**Figure 5:** Block diagram of the musculoskeletal model of the shoulder mechanism.
structures and muscle physiological cross-sectional areas. Because of the complexity of the morphology of the shoulder mechanism, and the relation assumed between the morphology of separate parts, an accurate parameter set obtained at one and the same cadaver is necessary. Chapter 2 describes the methodology of an extensive cadaver study comprehending right and left shoulders of seven cadavers. Results of inertia parameters and muscle contraction parameters are presented. Chapter 3 describes the fitting of regular geometric forms to the recorded position coordinates of morphological structures. These geometric forms can be implemented in the model. In the shoulder mechanism a number of muscles has large attachment sites. In order to model these muscles, a new theory for representing the mechanical effect by several muscle lines of action was developed. This theory is presented in Chapter 4. In Chapter 5 the finite element method is outlined. Special attention is paid to the inverse dynamic part and two special purpose elements to describe the scapulothoracic gliding plane and curved muscle lines of action, respectively. Results of model simulations are compared with EMG recordings. In Chapter 6 the model has been used for a kinematic and dynamic analysis of the behavior of the shoulder mechanism. The function of all muscles during humeral abduction and anteflexion is discussed. Chapter 7 describes the palpation method used for recording 3D motions of the shoulder mechanism. Accuracy and results of the method will be discussed. Focus has been laid on an interpretable way for describing motions of the various bones. Chapter 8 returns to the first onset of the shoulder project: The glenohumeral arthrodesis. The effect of fusion angles on hand position and maximal force exertion is simulated. In Chapter 9 the model has been applied in the field of ergonomics: The wheelchair user. Bony motions, external forces and EMG have been recorded simultaneously. Using the model, muscle forces and the load on a number of morphological structures could be calculated. Finally, in Chapter 10 the research presented in this thesis will be viewed in retrospective. The place of the current research approach will be discussed. The chapter will finish with a preview on further research items.
Chapter 2
Inertia and muscle contraction parameters for musculoskeletal modelling of the shoulder mechanism.

H.E.J. Veeger¹, F.C.T. Van der Helm², L.H.V. Van der Woude¹, G.M. Pronk², R.H. Rozendaal¹

¹: Faculty of Human Movement Sciences, Free University, Amsterdam, The Netherlands
²: Man-Machine Systems Group, Lab. for Measurement and Control, Dept. of Mechanical Engineering and Marine Technology, Delft University of Technology, Delft, The Netherlands


Abstract
To develop a musculoskeletal model of the shoulder mechanism, both shoulders of seven cadavers were measured to obtain a complete set of parameters. Using anthropometric measurements, the mass and rotational inertia of segments were estimated, followed by three-dimensional measurements of all morphological structures relevant for modelling, i.e. muscle origins and insertions, muscle bundle directions, ligament attachments and articular surfaces; all in relation to selected bony landmarks. Subsequently, muscle contraction parameters as muscle mass and physiological cross-sectional area were measured. The method of data collection and the results for inertia and muscle contraction parameters as prerequisites for modelling are described.
I Introduction

1.1 Musculoskeletal models
The development of musculoskeletal models is one of the major topics in biomechanics. Musculoskeletal models are used to establish a relation between the motions of the bones and the muscle forces causing these motions (Fig. 1). The aim of these models is on the one hand to gain insight in the factors which affect the relation, e.g. position of the rotation center, muscle attachments, muscle characteristics etc., and on the other hand to determine which muscles are involved and what the joint reaction forces are.

To describe the mechanical behavior of a musculoskeletal system, many characteristics of the system have to be considered. Due to the complexity of the system and depending on the purpose of the model involved, assumptions have to be made for most of these characteristics in order to develop an adequate and manageable model.

Firstly, bones are usually assumed to be rigid bodies. In the Newton-Euler approach the free body diagram and motion equations of this rigid body are derived. These equations include inertia properties, motion constraints and forces. To describe the dynamics of a musculoskeletal system, inertia properties of the system should be included. This means that segment mass, the segment center of mass and the segment moments of inertia have to be implemented in the model.

Secondly, within a model the direction of muscle force depends on the assumed geometry of muscle attachments, while the magnitude depends on assumptions concerning contraction characteristics and stimulation. Usually, the direction of the force of a muscle will be modelled along one muscle line of action, in spite of occasionally large attachment sites and complex

![Diagram](image)

**Figure 1:** Block diagram of a musculoskeletal model. \( \mathbf{x} \): Vector of position, velocity and acceleration of the bones; \( \mathbf{F} \): Vector of muscle forces.
architecture. One can distinguish the centroid line approach in which the moment arm of a muscle depends on the curved line that is formed by the centroid of that muscle, and the straight line approach which connects origin and insertion of a muscle (Jensen & Davy, 1975). Thirdly, contraction characteristics for force magnitude development involve muscle architecture, bundle length, tendon length, muscle length, contraction velocity and physiological cross-sectional area (PCSA). In most cases only the PCSA is included in a musculoskeletal model to get an approximation of the maximal force which a muscle can exert. The magnitude of the force exerted by a ligament depends on non-linear elastic characteristics, whereas the force direction depends on the position of ligament attachments. The effect of soft tissue surrounding bones, joints and muscles, including the effect of the joint capsule, is usually neglected.

To develop a sound musculoskeletal model, parameters describing all characteristics mentioned above should be included. For consistency, these parameters should preferably be measured within one subject. Despite the availability of new techniques like Magnetic Resonance Imaging (MRI), information on parameters is still highly dependent on cadaver studies. To date, no studies describing all information needed to derive the inertia, geometry and muscle contraction parameters have been published.

In a musculoskeletal model a relation between muscle force and calculated or prescribed motion is established. Thus in order to validate the model, both motion and muscle force have to be known. In addition, to use such a model with a registration of gross motor actions like walking or riding a wheelchair, the position of the optical markers has to be known with respect to the geometry of the musculoskeletal system in order to combine recorded motion with the accurate position of e.g. joint rotation centers and muscle attachments. To date, the parameters of the musculoskeletal system and the motion pattern can not yet be measured within one subject. Moreover, muscle force can not be measured directly and is usually estimated with the help of EMG. Therefore, musculoskeletal models can only be validated in a more qualitative sense.

1.2 The shoulder mechanism
The shoulder mechanism is an example of a very complex musculoskeletal system. The shoulder girdle consists of scapula and clavicle and functions as a movable but stable base for the motions of the humerus. The sternoclavicular (SC-) joint connects the clavicle and sternum, and the scapula in its turn is connected to the clavicle by the acromioclavicular (AC-) joint. Another connection between scapula and thorax is the scapulothoracic gliding plane, which constrains possible movements with two degrees-of-freedom and makes the system a closed chain. The
humerus articulates with the scapula at the glenohumeral (GH-) joint, a ball-and-socket joint. Three extracapsular ligaments can be identified in the shoulder girdle: the costoclavicular ligament limiting the range of motion of the SC-joint and the conoid and trapezoid ligaments acting at the AC-joint. Seventeen muscles are crossing the joints of the shoulder mechanism. Most muscles are polyarticular, fan-shaped and have large attachment sites.

For the movements of the humerus the large range of motion of the scapula over the thorax is essential. Since the motion of the scapula over the thorax is non-planar, it is very hazardous to measure it or describe it as a planar motion without making mistakes due to distortion of the image or omitting the constraints of clavicula and thorax. It is however difficult to reconstruct the actual motion of the scapula since it moves underneath the skin and is inaccessible for camera registration, except for X-ray of which the complications are obvious.

Scapular motions are related to those of the humerus. For this reason the position of the scapula is standardized with respect to the humerus (Inman et al., 1944; Pronk, 1987). Positioning the scapula is an active mechanism and consequently it is not self-evident that passive movements of the humerus result in the same scapulohumeral rhythm as active movements. Presently, knowledge about the shoulder mechanism as a whole is generally qualitative. Until now no 3D musculoskeletal model of the shoulder mechanism has been published. To develop such a model, many parameters describing the inertia, geometry and muscle contraction characteristics of the system have to be known. Despite a number of studies on the shoulder girdle, each study has included only part of the parameters needed and even if all studies are combined a large number of data will be missing.

1.3 Literature review

Formulae for the prediction of segment masses based on dissection and anthropometric data have been published by Barter (1957), Clauser et al. (1969) and Clarijs & Marfell-Jones (1986), whereas Clauser et al. also give predictions for the volumes and positions of the centers of gravity of the segments. Hinrichs (1985) determined predictions for moments of inertia based on the study of Chandler et al. (1975). During preparation of this manuscript new, non-linear prediction equations based on Chandler et al. (1975) were published by Yeadon & Morlock (1989).

The morphology and functional characteristics of the sternoclavicular joint were studied by Bearn (1967) and Depalma (1973). The orientation of joint surfaces of the acromioclavicular joint was described by Urist (1946) and Moseley (1972). The glenohumeral joint is the most extensively studied joint, probably due to the near resemblance with a ball-and-socket joint. The orientation of the glenoid (Friedel, 1926; Freedman & Munro, 1966; Oxnard, 1967; Saha, 1971), and the
Inertia and muscle contraction parameters

radius of the glenoid and/or the humeral head (Mollier, 1923; Dempster, 1955; Saha, 1961, 1971, 1973; Poppen & Walker, 1976, 1978; Hogfors et al., 1987) are the most elaborated items.

Braune & Fischer (1888) and Pfuhl (1934) have studied the function of the shoulder articulations, whereas Mollier (1899), Shiino (1913) and Hvorslev (1927) tried to simulate shoulder joint movements.

The length and thickness of both conoid and trapezoid ligament have been investigated by Salter et al. (1987). Fukuda et al. (1986) measured the distance between coracoid process and clavicle and deduced the length of the coracoclavicular ligaments.

Hogfors et al. (1987) have started to measure the position of the muscle attachments of the shoulder girdle three-dimensionally. Due to the limited description of their measurement method and data processing, it is not yet clear how their recorded data must be valued. Recently, Wood et al. (1980\textsuperscript{a}, 1980\textsuperscript{b}) published results of a cadaver study, based on a centroid line approach. The length of some muscles have been measured (Mollier, 1899; Fick & Weber, 1877; Shiino, 1913; Hvorslev, 1927; Bassett, 1983), as well as bundle length (Shiino, 1913; Bassett, 1983; Howell et al., 1986), centroids (Bassett, 1983) and moment arms (Bassett, 1983; Howell et al., 1986).

The physiological cross-sectional area (PCSA) of shoulder muscles is measured directly by Fick (1910, 1911), Shiino (1913) and Poppen and Walker (1978). The PCSA has been calculated from the product of muscle mass and inverse density divided by bundle length (Weber, 1851; Fick, 1910, 1911), the quotient of volume and muscle length (Bassett, 1983; Wood et al., 1980\textsuperscript{a}), the muscle volume and bundle length (Bassett, 1983; Howell et al., 1986) and maximum cross-sectional area perpendicular to the centroid (Wood et al., 1980\textsuperscript{a}).

Motion studies of the shoulder joint have often been limited to the glenohumeral joint (Poppen & Walker, 1978; Engin & Peindl, 1987; Peindl & Engin, 1987). Motion recording of the shoulder girdle has been performed using 2D X-ray photographical studies (Inman et al., 1944; Saha, 1961; Meijers, 1961; Freedman & Munro, 1966; Poppen & Walker, 1976; Dvir & Berme, 1978), cinematographical studies (Engen & Spencer, 1968; Hvorslev, 1927), goniometers (Conway, 1961; Doody et al., 1970\textsuperscript{a}, 1970\textsuperscript{b}), or pins inserted into the clavicle and scapula (Inman et al., 1944; Inman & Saunders, 1946; Kennedy & Cameron, 1954). All of these studies recorded only part of the essential 3D movement of clavicle, scapula and humerus. Only few authors (Wallace, 1982; Wallace & Johnson, 1982; Peterson et al., 1985) reported the results of 3D X-ray recordings of the motions of the shoulder girdle. They did however not present a 3D motion description. Recently in our research group attempts have been made for a 3D recording with use of a new instrument, the so-called palpator (Pronk, 1987).
I.4 Towards a musculoskeletal model of the shoulder mechanism

Our goal is to develop a 3D musculoskeletal model of the shoulder mechanism. Such a model will improve the understanding of complex behavior of this mechanism, so that it will be useful in clinical practice. Classic Newton-Euler and Lagrangian approaches involve a large number of equations and derivations. In addition, the complex motion constraint of the scapulothoracic gliding plane will limit the application of these approaches. Therefore a kinematic and dynamic finite element method is used to develop a 3D musculoskeletal model (Van der Helm, 1988; Van der Helm & Pronk, 1989; Pronk, 1989).

The predictive values of this musculoskeletal model will heavily depend on the parameters describing the characteristics of the system. It would be preferable to derive a whole set of parameters from one subject, since the interaction between certain parameters is not understood yet. Therefore, an extensive morphological study has been performed in order to acquire a complete set of parameters for the shoulder mechanism. To date the only way to derive these parameters is in a dissection experiment. In this paper the results are reported of a study of seven cadavers. For each cadaver the inertia parameters as segment mass and center of gravity were estimated from anthropometric data using regression equations of Clauser et al. (1969) and the 3D coordinates of morphological structures were measured. Obviously, muscle contraction parameters as force-length and force-velocity relations were impossible to derive from these dissection experiments. The only parameter which can reasonably be derived is the maximal force output of a muscle. Therefore, for each muscle in the shoulder mechanism the PCSA and muscle mass were measured.

II Materials and methods

Seven preserved human bodies (five males and two females) were used. No pre-experimental selection of cadavers on age at death or physical appearance took place.

During the experiment the following sequence of steps was made:

1. Measurement of relevant body dimensions for the derivation of inertia parameters segment mass, volume, segment mass position and moments of inertia.
3. To enable reconstruction after dissection of the original positions of humerus, scapula, clavicle and thorax in a global, whole body, coordinate system, small screws were placed in the four different 'segments' so as to define the separate local coordinate systems.
4. Measurement of the positions of the screws on the four segments with respect to a global,
Cartesian coordinate system.

5. Dissection and exarticulation of the cadavers at the scapulothoracic, acromioclavicular, sternoclavicular and glenohumeral joints. During the process muscles were cleaned and cut and fibre bundles were marked with colored beads for future identification.

6. Measurement of the locations of muscle and ligament insertions and relevant articular surface shapes on the exarticulated segments relative to the screws.

7. Muscle parameters mass and cross-sectional area were determined.

II.1 Determination of inertia parameters

On all cadavers the body dimensions necessary for the calculation of inertia parameters from regression equations as published by Clauser et al. (1969) and Hinrichs (1985) were measured. Measurements were made on the left side of the body, following the protocol and definitions of Clauser et al. as much as possible. Chandler et al. (1975) presented inertia tensors about anatomical axes of segments which had been determined for each segment using the pendulum technique. From this the direction of the principal axes around which the matrix becomes diagonal and the related principal moments of inertia could be calculated. Hinrichs (1985) and Yeadon & Morlock (1989) reduced data for limb segments further by averaging the anterior-posterior and medial-lateral moments of inertia and subsequently determined regression equations for one mean transversal moment of inertia and a longitudinal moment of inertia. The total inertia tensor could thus be reduced to two moments of inertia about a transverse axis and the longitudinal principal axis of a segment through its center of mass. An illustration of procedures for obtaining segment data has been given by Kaleps et al. (1984).

Centers of mass have generally been estimated relative to proximal landmarks (Clauser et al., 1969). Moments of inertia have been given as predictions following Hinrichs (1985) about the principal axes through the centers of mass based on the study by Chandler et al. (1975). Original definitions of landmarks are given in McConville et al. (1980).

II.2 Determination of geometry parameters; identification of coordinate systems

For each cadaver a global coordinate system (Fig. 2) was defined as related to the anatomical conventions with its origin in the Incisura Jugularis, the X-axis is along the frontal plane towards the right shoulder, the Y-axis is directed cranially and the Z-axis is in the sagittal plane towards the back.
Configuration measurements needed to determine the position of the segments with respect to the global system took place prior to dissection. For measurements a metal frame was constructed in which the cadaver could be fixed (Fig. 3). After the legs were removed, the upper half of a cadaver was positioned into the frame via a rod through the head in such a way that the frontal plane was as much as possible parallel to the frontal face of the frame. Movement in frontal and lateral direction was prevented by fixation of the pelvis. The elbows of the cadaver were extended.

Figure 2:
Ventral view of a right shoulder mechanism with its global coordinate system. Shown are thorax, scapula, clavicle and humerus. Origin: Incisura Jugularis; X-axis: left to right; Y-axis: caudal to cranial; Z-axis: ventral to dorsal.

Figure 3: Fixation frame for cadaver measurements. The cadaver is seen from the back with the palpator fixed to the frame on its left-hand side.
maximally and the arms tied to the torso. Since the influence of gravity during the measurements might affect the accuracy of results, sagging was checked for one cadaver. Over a period of 2 hours, (twice the measurement period) no change in the position of reference markers could be measured.

The faces of the frame were used as the reference planes of an orthogonal coordinate system. Three-dimensional (3D) positions of morphological structures were measured with use of a "palpator" (Fig. 4). The palpator is an open chain of four links of 0.2 m each, connected by four perpendicular hinges. The rotation of each hinge is recorded with a potentiometer and is on-line A-D converted (Olivetti M21, DT2801 AD converter). The 3D position of the end point of the fourth link can be calculated with respect to an internal coordinate system of the palpator, defined previously in calibration measurements. Using the palpator, a large number of datapoints can be collected in a fast and easy way. The measurement error of the palpator was estimated to have a standard deviation of 0.96 mm per coordinate or 1.43 mm in absolute distance (Pronk & van der Helm, 1991).

Due to the impossibility to reach all structures within the same coordinate system without damaging the interrelations and the restricted reach of the palpator, measurements of the four segments within their own coordinate systems were needed. To reconstruct data of the segment measurements in the global coordinate system and thus to allow descriptions of segments relative to each other

Figure 4: The palpator: a measurement device to record three-dimensional coordinates.
within a whole body setting, at least four non-collinear joint reference markers which are measured in both the global and the separate segment systems were required. For reference markers six Pozidriv screws (2.5x12 mm) were screwed into the bone of each segment: humerus, clavicula, scapula and thorax. Using conjoined markers and regarding the segments as rigid bodies, the rotation matrix and translation vector for transformation of data from the individual segment system measurements to the global (whole body) configuration could be calculated according to Veldpaus et al. (1988). For each side of a cadaver, reference markers were measured five times, thus forming the database needed for transformation towards the global system. For future calculations the data of the left side of the body were transformed to the right.

Additional to the measurements described, a selection of anatomical landmarks was measured for future connection of these landmarks with subcutaneous dimensions. These landmarks are listed in Table 1.

Summing up, two subsequent transformations would be required after data collection:
- transformation of data measured on insertions and shapes of articular surfaces from individual segment coordinates to the global coordinate system using the configuration of reference markers to estimate rotation matrix R and translation vector v.
- transformation of data measured on the left side of the body to the right, thus creating a set of descriptions of 14 shoulder mechanisms, all within coordinate systems comparable by their definitions with respect to the global coordinate system.

II.3 Determination of geometry parameters: exarticulation and data collection

Following the global measurements, cadavers were partly dissected by students as part of their dissection course under supervision of the authors. Dissection consisted of identification and cleaning of shoulder muscles and their attachments. Before they were removed by standardized cuts, some representative muscle fiber bundles were marked with colored beads which were fastened close to the attachments of those bundles. Thus after exarticulation the direction of muscle fiber bundles could be reconstructed by connecting the coordinates of beads of the same color. Exarticulation was performed by the authors at the glenohumeral, sternoclavicular and acromioclavicular joints and the scapulothoracic wall with as little damage to ligaments, attachments and articular surfaces as possible.

Local system measurements took place using the same frame as for the configuration measurements. Segments were positioned and fixed in order that all data points could be measured
Table 1: Structures measured on scapula, humerus, thorax and clavicle of all cadavers. The additions '-O' and '-I' stand for origin and insertion (* measured on the forearm).

<table>
<thead>
<tr>
<th>THORAX</th>
<th>CLAVICULA</th>
<th>SCAPULA</th>
<th>HUMERUS</th>
</tr>
</thead>
<tbody>
<tr>
<td>REFERENCE MARKERS</td>
<td>REFERENCE MARKER 1</td>
<td>Reference marker 1</td>
<td>Reference marker 1</td>
</tr>
<tr>
<td>Reference marker 2</td>
<td>Reference marker 2</td>
<td>Reference marker 2</td>
<td>Reference marker 2</td>
</tr>
<tr>
<td>Reference marker 3</td>
<td>Reference marker 3</td>
<td>Reference marker 3</td>
<td>Reference marker 3</td>
</tr>
<tr>
<td>Reference marker 4</td>
<td>Reference marker 4</td>
<td>Reference marker 4</td>
<td>Reference marker 4</td>
</tr>
<tr>
<td>Reference marker 5</td>
<td>Reference marker 5</td>
<td>Reference marker 5</td>
<td>Reference marker 5</td>
</tr>
<tr>
<td>Reference marker 6</td>
<td>Reference marker 6</td>
<td>Reference marker 6</td>
<td>Reference marker 6</td>
</tr>
</tbody>
</table>

MUSCLES

latissimus dorsi-O
trapezius-O
rhomboideus-O
levator scapulae-O
pectoralis major-O
pectoralis minor-O
serratus anterior-O
latissimus dorsi-O
trapezius-I
rhomboideus-I
levator scapulae-I
pectoralis major-O
pectoralis minor-I
serratus anterior-I
deltoideus-O
deltoideus-I
subscapularis-O
subscapularis-I
supraspinatus-O
supraspinatus-I
infraespinaus-O
infraespinaus-I
teres minor-O
teres minor-I
teres major-O
teres major-I
biceps c. breve-O
biceps-I*
biceps c. longum-O
biceps-I*
coracobrachialis-O
coracobrachialis-I
triceps c. longum-O
triceps-I*

LIGAMENTS

lig. costoclavicular
lig. costoclavicular
lig. conoideum
lig. conoideum
lig. trapezoid
lig. trapezoid

ARTICULAR SURFACES

art. sternoclavicular
art. sternoclavicular
art. acromioclavicular
art. acromioclavicular
art. coracoclavicular
art. coracoclavicular
art. glenohumeral
art. glenohumeral

scapulothoracic wall

ANATOMICAL LANDMARKS

incisura jugularis
trigonum spine
angulus inferior
angulus acromialis
art. acromioclavicular
tuberculum minus
tuberculum majus
collum humeri
sulcus humeri

C7
epicond. medialis
epicond. lateralis
olecranon
art. acromioclavicular

BONY CONTOURS

muscle path serratus ant.
without changing the position of either the segment measured or the base of the palpator. The humerus was fastened with the elbow extended.

For each muscular or ligamentous attachment or articular surface a large number of datapoints was collected. Table 1 is a listing of structures measured. The position of these structures will subsequently be described by fitting data to a geometric form. Dependent on the fitting procedure for each structure sufficient data had to be collected, ranging from 3-5 points for, for instance, the origin of m. coracobrachialis, up to 30-50 points for a structure needing a more complex fitting procedure (e.g. surface of the glenohumeral joint). Fitting procedures used in this project will be estimations of the parameters of a point, line, plane, ellipsoid, cylinder or a sphere. The algorithms for these fittings are described extensively elsewhere (Van der Helm et al., 1991).

II.4 Determination of muscle contraction parameters

Information on muscle bundle direction was collected by measuring coordinates of the affixations of muscle bundles that were marked with beads earlier.

Following the measurement of attachment locations and articular surfaces, all muscles listed in Table 1 were removed from their bony affixtures for weighing. To ensure comparable measurements for all cadavers at equal levels of saturation, muscles were kept in a solution of formalin and weighed three times at intervals of approximately one week. Based on the recordings of muscles that were observed not to be completely saturated, results for each muscle recording less than 96% of the mean weight of that muscle with a minimum difference of 2.5 grams were discarded on the assumption of incomplete saturation.

Physiological cross-sectional areas (PCSA) were determined using the following procedure: Out of all muscles thinouples are cut at the level of the apparently largest cross-sectional area perpendicular to the bundle direction of the muscle. The couples coming from flat muscles (e.g. m. serratus anterior) are rolled up and together with couples of easier measurable muscles (e.g. m. biceps brachii) photographed for area determination using a digitizer (Summagrahics Supergrid, accuracy 1/40 mm). All PCSA couples were digitized within an accuracy of 10 mm². Finally, all humeri, claviculae and scapulae were cleaned and stored for future reference.

III Results

The population of this study was selected on availability. Moreover, as a consequence of the duration of the dissection project embalmed cadavers were used. Table 2 gives the population characteristics, combined with available data on other relevant cadaver studies. In relation to those
Inertia and muscle contraction parameters

Table 2: Summary of the population characteristics of this and other relevant studies.
   #: number of cadavers; em.: number of embalmed cadavers.

<table>
<thead>
<tr>
<th></th>
<th>Age (yrs)</th>
<th>s.d.</th>
<th>length (cm)</th>
<th>s.d.</th>
<th>weight (kg)</th>
<th>s.d.</th>
<th>em.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bischoff (1863)</td>
<td>1</td>
<td>33</td>
<td>-</td>
<td>168.0</td>
<td>-</td>
<td>69.7</td>
<td>-</td>
</tr>
<tr>
<td>Theile (1884)</td>
<td>1</td>
<td>26</td>
<td>-</td>
<td>166.7</td>
<td>-</td>
<td>64.0</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>64.5</td>
<td>19.1</td>
<td>-</td>
<td>-</td>
<td>33.9</td>
<td>4.5</td>
</tr>
<tr>
<td>Dempster (1955)</td>
<td>8</td>
<td>68.5</td>
<td>11.0</td>
<td>169.4</td>
<td>-</td>
<td>59.8</td>
<td>8.3</td>
</tr>
<tr>
<td>Clauser et al. (1969)</td>
<td>13</td>
<td>49.3</td>
<td>13.7</td>
<td>172.7</td>
<td>11.2</td>
<td>66.5</td>
<td>8.7</td>
</tr>
<tr>
<td>Basset (1983)</td>
<td>4</td>
<td>66.3</td>
<td>9.9</td>
<td>-</td>
<td>13.7</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Clarijs &amp; M-J (1986)</td>
<td>6</td>
<td>66.8</td>
<td>25.7</td>
<td>-</td>
<td>-</td>
<td>57.6</td>
<td>13.4</td>
</tr>
<tr>
<td>Hogfors et al. (1987)</td>
<td>3</td>
<td>55-71</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Wood et al. (1989)</td>
<td>1</td>
<td>-</td>
<td>-</td>
<td>183.0</td>
<td>-</td>
<td>90.7</td>
<td>-</td>
</tr>
<tr>
<td>This study</td>
<td>7</td>
<td>80.0</td>
<td>7.0</td>
<td>171.1</td>
<td>-</td>
<td>76.1</td>
<td>16.4</td>
</tr>
</tbody>
</table>

studies this population was considerably older (\bar{x} = 80.0) and heavier (\bar{x} = 76.1) than other cadaver populations. Mean stature however equals the population used by Clauser et al. (1969). Cadaver studies specifically focusing on the shoulder have usually been performed on half or quarter specimen (Shiino, 1913; Poppen & Walker, 1978; Basset, 1983; Howell et al., 1986) and general anthropometric data describing the individual bodies are missing. A comparison with those studies in that respect was thus not possible.

III.1 Determination of inertia parameters

The results for the anthropometric measurements are summarised in Table 3. The cadaver population was very heterogeneous: Cadaver 7 was half as heavy as cadaver 1 (51.7 vs 104 kg), whereas for instance waist circumference varied from 79 to 115.5 cm. The anthropometric data were used for the estimation of segment properties on basis of regression equations published by Clauser et al. (1969). For definition of segmentations the reader should refer to Clauser et al. (1969). The longitudinal and transversal principal moments of inertia were predicted relative to the center of mass of segments (Hinrichs, 1985). Values are given in Table 4. For one specimen (K7), estimated longitudinal moments of inertia of upper arm and forearm were negative.
Table 3: Results for a selection of anthropometric measurements required for the calculation of inertia parameters. Measuring procedures and descriptions were according to Clauser et al. (1969). All values except for weight (kg) are given in cm.

<table>
<thead>
<tr>
<th>Gender</th>
<th>K1 Male</th>
<th>K2 Male</th>
<th>K3 Male</th>
<th>K4 Female</th>
<th>K5 Female</th>
<th>K6 Male</th>
<th>K7 Male</th>
<th>Mean</th>
<th>s.d.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>76</td>
<td>90</td>
<td>70</td>
<td>86</td>
<td>82</td>
<td>82</td>
<td>74</td>
<td>80</td>
<td>7.0</td>
</tr>
<tr>
<td>Weight</td>
<td>104</td>
<td>72.9</td>
<td>69.2</td>
<td>71.3</td>
<td>88.9</td>
<td>74.9</td>
<td>51.7</td>
<td>76.1</td>
<td>16.4</td>
</tr>
<tr>
<td>Stature</td>
<td>181.6</td>
<td>172.7</td>
<td>179.4</td>
<td>164.2</td>
<td>166.1</td>
<td>167.6</td>
<td>166.5</td>
<td>171.2</td>
<td>6.9</td>
</tr>
<tr>
<td>Top of head to chin/nech int.</td>
<td>24.7</td>
<td>23.6</td>
<td>22.2</td>
<td>22.8</td>
<td>21.9</td>
<td>24.3</td>
<td>21.5</td>
<td>23.0</td>
<td>1.2</td>
</tr>
<tr>
<td>Top of head to trochanterion</td>
<td>85.8</td>
<td>85.5</td>
<td>83.9</td>
<td>79.8</td>
<td>86.7</td>
<td>81.8</td>
<td>83.5</td>
<td>83.9</td>
<td>2.4</td>
</tr>
<tr>
<td>Top of head to tibiale length</td>
<td>48.4</td>
<td>45.2</td>
<td>51.2</td>
<td>47.7</td>
<td>45.3</td>
<td>44.8</td>
<td>43.7</td>
<td>46.6</td>
<td>2.6</td>
</tr>
<tr>
<td>Chest depth</td>
<td>27.1</td>
<td>19.7</td>
<td>24.3</td>
<td>25.3</td>
<td>23.6</td>
<td>22.1</td>
<td>20.1</td>
<td>23.2</td>
<td>2.7</td>
</tr>
<tr>
<td>Bicristal breadth</td>
<td>34.7</td>
<td>32.2</td>
<td>27.3</td>
<td>34</td>
<td>36.8</td>
<td>34.3</td>
<td>29.5</td>
<td>32.7</td>
<td>3.3</td>
</tr>
<tr>
<td>Bispinous breadth</td>
<td>24.4</td>
<td>26.5</td>
<td>23.7</td>
<td>25.1</td>
<td>29.6</td>
<td>27.4</td>
<td>24.0</td>
<td>25.8</td>
<td>2.1</td>
</tr>
<tr>
<td>Elbow breadth</td>
<td>8.3</td>
<td>6.9</td>
<td>7.5</td>
<td>5.9</td>
<td>6.6</td>
<td>7.0</td>
<td>7.0</td>
<td>7.0</td>
<td>0.7</td>
</tr>
<tr>
<td>Hand breadth</td>
<td>8.8</td>
<td>8.4</td>
<td>8.3</td>
<td>7.6</td>
<td>7.2</td>
<td>8.5</td>
<td>8.0</td>
<td>8.1</td>
<td>0.5</td>
</tr>
<tr>
<td>Wrist breadth</td>
<td>6.4</td>
<td>5.7</td>
<td>6.0</td>
<td>5.5</td>
<td>5.4</td>
<td>5.7</td>
<td>5.9</td>
<td>5.8</td>
<td>0.3</td>
</tr>
<tr>
<td>Head circumference</td>
<td>62.6</td>
<td>59.0</td>
<td>58.8</td>
<td>58.4</td>
<td>60.2</td>
<td>60.0</td>
<td>60.5</td>
<td>59.9</td>
<td>1.4</td>
</tr>
<tr>
<td>Chest circumference</td>
<td>120.3</td>
<td>101.7</td>
<td>99.8</td>
<td>104.0</td>
<td>110.7</td>
<td>100.0</td>
<td>89.5</td>
<td>103.7</td>
<td>9.7</td>
</tr>
<tr>
<td>Waist circumference</td>
<td>115.5</td>
<td>91.3</td>
<td>83.7</td>
<td>103.0</td>
<td>113.2</td>
<td>102.0</td>
<td>79.0</td>
<td>98.2</td>
<td>14.1</td>
</tr>
<tr>
<td>Axillary arm circumference</td>
<td>36.1</td>
<td>28.0</td>
<td>29.3</td>
<td>32.6</td>
<td>32.8</td>
<td>27.7</td>
<td>23.0</td>
<td>29.9</td>
<td>4.3</td>
</tr>
<tr>
<td>Biceps circumference</td>
<td>31.1</td>
<td>24.7</td>
<td>26.6</td>
<td>31.3</td>
<td>30.9</td>
<td>25.6</td>
<td>21.3</td>
<td>27.4</td>
<td>3.9</td>
</tr>
<tr>
<td>Forearm circumference</td>
<td>29.0</td>
<td>23.9</td>
<td>25.1</td>
<td>25.3</td>
<td>26.0</td>
<td>24.9</td>
<td>22.3</td>
<td>25.2</td>
<td>2.1</td>
</tr>
<tr>
<td>Wrist circumference</td>
<td>19.5</td>
<td>17.9</td>
<td>18.1</td>
<td>17.2</td>
<td>17.7</td>
<td>18.4</td>
<td>17.5</td>
<td>18.1</td>
<td>0.8</td>
</tr>
<tr>
<td>Acromion-radiale length</td>
<td>35.1</td>
<td>34.7</td>
<td>35.5</td>
<td>31.0</td>
<td>29.7</td>
<td>31.9</td>
<td>29.8</td>
<td>32.6</td>
<td>2.5</td>
</tr>
<tr>
<td>Ball of humerus-radiale length</td>
<td>33.8</td>
<td>34.5</td>
<td>35.2</td>
<td>30.6</td>
<td>31.1</td>
<td>31.1</td>
<td>28.8</td>
<td>32.2</td>
<td>2.4</td>
</tr>
<tr>
<td>Radiale-stylin length</td>
<td>28.1</td>
<td>27.0</td>
<td>27.7</td>
<td>24.6</td>
<td>22.3</td>
<td>24.3</td>
<td>26.2</td>
<td>25.7</td>
<td>2.1</td>
</tr>
<tr>
<td>Fat iliac crest</td>
<td>18.0</td>
<td>11.7</td>
<td>7.9</td>
<td>12.5</td>
<td>23.0</td>
<td>18.0</td>
<td>7.3</td>
<td>14.1</td>
<td>5.8</td>
</tr>
</tbody>
</table>

III.2 Determination of geometry parameters

To calculate the position of data points measured in the global coordinate system, a transformation function was estimated using a least squares criterion (Veldpaus et al., 1988). This transformation function describes the rotation and translation of the configuration of screws in a local coordinate system to the configuration in a global coordinate system. Normally six screws were used to estimate the transformation function. However, in some cases screws were irretrievable or malpositioned and could thus not be digitized. The mean and standard deviation of the mean residual error associated with estimations of the transformation function are shown in Table 5.

III.3 Determination of muscle contraction parameters

Results of PCSA measurements and muscle masses, combined with relevant data from other
Inertia and muscle contraction parameters

Table 4: Mass, mass position and volume and moments of inertia for the segments, calculated on the basis of regression equations by Clauser et al. (1969) and transversal and longitudinal moments of inertia, calculated on the basis of equations by Hinrichs (1985). Given in brackets are the positions relative to which segment mass positions are given. Original definitions of landmarks are given in McConvil e et al. (1980). LI = Incisura Jugularis, AC = Acromion, CR = Caput Radii, MCIII = Distal head of third metacarpal.

<table>
<thead>
<tr>
<th>Segment masses (kg)</th>
<th>K1</th>
<th>K2</th>
<th>K3</th>
<th>K4</th>
<th>K5</th>
<th>K6</th>
<th>K7</th>
<th>mean</th>
<th>s.d.</th>
</tr>
</thead>
<tbody>
<tr>
<td>head</td>
<td>5.88</td>
<td>5.04</td>
<td>4.96</td>
<td>4.95</td>
<td>5.41</td>
<td>5.17</td>
<td>4.88</td>
<td>5.19</td>
<td>0.35</td>
</tr>
<tr>
<td>trunk</td>
<td>54.23</td>
<td>39.46</td>
<td>37.66</td>
<td>37.35</td>
<td>48.33</td>
<td>37.90</td>
<td>29.30</td>
<td>40.60</td>
<td>8.17</td>
</tr>
<tr>
<td>head&amp;trunk</td>
<td>60.76</td>
<td>43.16</td>
<td>42.94</td>
<td>41.98</td>
<td>53.92</td>
<td>42.81</td>
<td>32.95</td>
<td>45.50</td>
<td>9.07</td>
</tr>
<tr>
<td>total arm</td>
<td>4.55</td>
<td>3.29</td>
<td>3.43</td>
<td>3.68</td>
<td>4.00</td>
<td>3.49</td>
<td>2.64</td>
<td>3.58</td>
<td>0.60</td>
</tr>
<tr>
<td>upper arm</td>
<td>2.71</td>
<td>1.72</td>
<td>1.86</td>
<td>1.95</td>
<td>2.02</td>
<td>1.56</td>
<td>0.87</td>
<td>1.81</td>
<td>0.55</td>
</tr>
<tr>
<td>forearm</td>
<td>1.44</td>
<td>1.04</td>
<td>1.12</td>
<td>1.06</td>
<td>1.14</td>
<td>1.14</td>
<td>0.93</td>
<td>1.12</td>
<td>0.16</td>
</tr>
<tr>
<td>forearm&amp;hand</td>
<td>2.01</td>
<td>1.56</td>
<td>1.67</td>
<td>1.45</td>
<td>1.44</td>
<td>1.55</td>
<td>1.41</td>
<td>1.58</td>
<td>0.21</td>
</tr>
<tr>
<td>hand</td>
<td>0.57</td>
<td>0.46</td>
<td>0.49</td>
<td>0.40</td>
<td>0.40</td>
<td>0.48</td>
<td>0.45</td>
<td>0.46</td>
<td>0.06</td>
</tr>
</tbody>
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<td>246.4</td>
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<td>1.3</td>
<td>1.4</td>
<td>0.7</td>
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* Estimation inaccuracy (see Discussion)
Table 5: Mean and standard deviation (in mm) of the residual errors $e_t$, $e_c$, $e_s$ and $e_h$, associated with the estimation of the transformation from segment measurements to configuration measurements of thorax, clavicle, scapula and humerus. In most cases the mean residual error is averaged over the residues of six screws. Different numbers are indicated in brackets.

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<tr>
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<th>$e_c$</th>
<th>$e_s$</th>
<th>$e_h$</th>
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</tr>
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<td>1.73</td>
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</tr>
<tr>
<td>K6 right left</td>
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<td>1.12 (4)</td>
<td>1.94 (4)</td>
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<td>2.68 (5)</td>
</tr>
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authors are given in Tables 6 and 7. In this study of m.triceps only the long head was measured. For those authors who stated data for the separate parts of m.deltoides (Shiino, 1913; Poppen & Walker, 1978; Howell et al. 1986, Wood et al., 1989), the combined figure is given. Data from Wood et al. (1989) on PCSA listed in Table 6 are values calculated on the basis of Coons surface grid data. Also included are their cross-sectional data calculated from volume and muscle length. Statistical comparison on the existence of left-right differences for PCSA or muscle mass within each body (T-test, 16 or 17 muscles, p<0.01) did not lead to significant results. Between PCSA and muscle mass a strong correlation was found; all muscles except m. levator scapulae, m. trapezius, m. infraspinatus and m. supraspinatus showed a significant correlation (Pearson correlation; N=14, p<0.01).
Table 6: Physiological cross-sections of shoulder muscles (in cm²) of this study compared with other results.

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<td>(N=6)</td>
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<td>4.75</td>
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1. Specimen combined with Fick's data.
2. Long head only.
3. Data from Weber.
4. Calculated with the use of coxen's surface grids.
5. Calculated as volume/muscle length.
Table 7: Muscle weights (in g) as measured in this study compared with other results.

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<th>Theile (1884)</th>
<th>Shiino (1913)</th>
<th>Wood et al. (1989)</th>
<th>This study (N=7)</th>
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<td>T.XLI</td>
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<td>left</td>
<td>left</td>
<td>(left?)</td>
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<td>259.4</td>
<td>223.6</td>
<td>99.1</td>
<td>73.9</td>
<td>3</td>
</tr>
<tr>
<td>weight left side</td>
<td>1982.4</td>
<td>2806.3</td>
<td>3217.0</td>
<td>1092.2</td>
<td>797.4</td>
<td>1697.7</td>
</tr>
</tbody>
</table>

|                  | right        | right          | right         | right         | mean       | s.d.       |                 |
|                  |              | right          | right         | right         |            |           |                 |
| biceps           | 210.7        | 181.3          | 97.0          | 108.2\(^1\)  | 112.2      | 20.2      |                 |
| triceps          | 177.8        | 44.1           | 95.0\(^2\)    | 127.8\(^2\)   | 99.7\(^2\) | 10.1      |                 |
| coracobrachialis | 59.1         | 53.4           | 30.0          | 21.3          | 30.6       | 8.6       |                 |
| levator scapulae | 47.4         | 46.0           | 3              | 68.8          | 71.4       | 20.2      |                 |
| rhomboideus      | 83.5         | 99.4           | 3              | 198.0         | 339.2      | 226.1     | 71.8            |
| lat. dorsi       | 361.8        | 3              | 104.0         | 85.2          | 109.8      | 16.2      |                 |
| trapezius        | 347.7        | 245.4          | 3              | 370.4         | 185.8      | 25.3      |                 |
| teres major      | 140.5        | 525.0\(^4\)    | 101.0         | 70.5          | 88.3       | 25.9      |                 |
| teres minor      | 30.5         | 26.1           | 23.0          | 24.6          | 28.3       | 6.1       |                 |
| infraspinatus    | 178.0        | 180.2          | 104.0         | 85.2          | 109.8      | 16.2      |                 |
| supraspinatus    | 76.8         | 50.9           | 49.0          | 39.3          | 36.2       | 7.2       |                 |
| subscapularis    | 200.8        | 225.2          | 135.0         | 121.3         | 138.6      | 21.1      |                 |
| deltoideus       | 473.8        | 472.8          | 254.0         | 386.7         | 314.4      | 72.9      |                 |
| pect. major      | 408.5        | 431.1          | 235.0         | 286.8         | 202.6      | 38.7      |                 |
| pect. minor      | 66.5         | 58.1           | 3              | 52.4          | 41.5       | 11.7      |                 |
| serr. ant.       | 276.0        | 223.6          | 193.4         | 204.8         | 41.9       |           |                 |
| weight right side| 3139.4       | 3259.6         | 2295.8        | 3824.9        | 3824.9     |           |                 |

1 given as volumes with a density of 1 g/cm\(^3\)
2 long head only
3 not available
4 as given
5 infraspinatus + teres minor
IV Discussion

Descriptions of data on inertia parameters, geometry parameters and muscle contraction parameters (PCSA and mass) of the shoulder mechanism muscles have not often been reported on. This is not surprising since the importance of such data, besides being merely descriptive, has largely been dependent on the application of such data in a musculoskeletal model. Since the physical models of the 19th and early 20th century (Mollier, 1899; Shiino, 1913; Hvorslev, 1927) and the, until now essentially two-dimensional models of Poppen & Walker (1978) and Howell et al. (1986), progressing computerisation has facilitated the development of a 3D model of the shoulder mechanism and has thus increased the need for a complete set of data.

The cadavers used in this study were generally older, taller and heavier than in other cadaver studies (Table 2), although the difference in average stature with Clauser et al. (1969) is small. The average difference in weight with the Clauser population seems mainly to be the result of a higher percentage bodyfat: the difference in waist circumference between both populations is 22% whereas the thickness of the panniculus adiposus on the iliac crest differed 33% (14.1 vs. 10.6 mm).

The population in this study was heterogeneous. A pre-selection of cadavers on, for instance, weight or stature might have lowered the variance in anthropometrical results. This would however not necessarily have led to a lower variance in (muscle parameter) data concerning the shoulder. Unfortunately, no information was available on shoulder complaints or shoulder malfunctioning during life, and thus the possibility exists that modelling will be performed using data derived from one or more malfunctioning shoulders.

IV.1 Determination of inertia parameters

Within the aims and methods of this study clearly direct measurement of inertia parameters was not feasible. Fortunately, regression equations on the relation of anthropometric variables with inertia parameters have been published (Barter, 1957; Clauser et al., 1969; Hinrichs, 1985; Clarijs & Marfell-Jones, 1986). Indirect determination of these parameters from anthropometric dimensions was thus tenable (Tables 3 and 4). For the interpretation of the accuracy of these derived segment property data, two factors should however be kept in mind. Firstly, the cadaver population of Chandler et al. (1975) on the basis of which Hinrichs (1985) calculated the inertia regression equations was small (N=6). Secondly, the difference between anthropometric dimensions of individual cadavers in this study and the Clauser and Chandler populations should be kept in mind since predictions on the basis of Multiple Regressions will become inaccurate when calculated over data diverging too much from the original data set. The negative longitudinal
moments of inertia calculated for K7 are thus probably the result of prediction inaccuracy due to the application of the formulae on diverging data. K7 was a light specimen with a body weight of only 51.7 kg. Hinrichs (1985) describes a comparable estimation error. It is possible that non-linear regression equations given by Yeadon et al. (1989) might have led to better results. However, since measurements had taken place before publication of their results, not all anthropometric parameters needed were available (Table 4) and their equations have not been used.

Since of all but one cadaver the skin on the legs was partially loosened, but not removed when the anthropometry was performed, for those cadavers no leg anthropometry data could be collected. Hence no comparison could be made between the sum of the segment masses and total body weight as in Miller & Morrison's (1975) study on the applicability of the regression equations on living subjects. For K2 this indirect control method of the comparison of total mass of all segments with body weight led to an overestimation of only 1.9%. It should however be kept in mind that the anthropometric dimensions of this cadaver fell well inside the range of anthropometric dimensions of the Clauser population. The moments of inertia estimated here were principal moments of inertia or the inertia tensor belongs to a set of orthogonal axes in which only the diagonal values are non-zero. These principal axes have their origin in the center of mass and the longitudinal principal axis is approximately aligned with the longitudinal axis of the segment as defined by the bony landmarks (Chandler et al., 1975; McConville et al., 1980; Kaleps et al., 1984). In most limb segments excellent alignment between the principal axes and the anatomical axes can be achieved by a minor rotation over a single axis (McConville et al., 1980).

IV.2 Determination of muscle contraction parameters

The PCSA of a muscle is used as an indication of its maximal strength. The relation between force and area for human subjects has been calculated by several authors. On the basis of an analysis of data from earlier studies and on their own results, Weis & Hillen (1985) determined this relation for human jaw muscles as $P = 0.37\times10^6 \text{ Nm}^{-2}$.

This value includes however the uncertainty of the effect of muscles possibly being measured at different lengths as the result of differences in body posture at embalming, thus representing different positions in the force-length diagram.

Apart from data published by Weber (1851), and recently Wood et al. (1989) no complete set of PCSA of shoulder mechanism muscles has come to the knowledge of these authors. Between data from this study and earlier publications, some striking differences in PCSA can be noted (Table 6). It should however be kept in mind that Weber measured on a young, muscular cadaver and
Shiino (1913) selected three muscular specimen out of a total of thirty for the calculation of his average results. Moreover, Weber's (1851) measurements return in Fick's (1911) and Poppen & Walker's (1978) data. Also, different determination methods have been used: calculation of the PCSA from muscle mass and length (Weber, 1851); calculation on basis of volume and muscle length (Bassett, 1983, Wood et al., 1989\textsuperscript{a}), volume and bundle length (Bassett, 1983; Howell et al., 1986) or on the basis of Coons surfaces (Wood et al., 1989\textsuperscript{a}). Wood et al.'s calculation methods led to considerable differences in which the PCSA values calculated from volume and muscle length were considerably lower. Next to data reported from Weber, data given by Fick also contain directly measured values, as well as Poppen & Walker's data extension.

Despite differences in measurement method, data of Bassett agreed well with our data ($r=0.938$, coeff. = 1.002) probably indicating the usefulness of indirect calculations. On the other hand data from Weber and Shiino differed significantly from our results (T-test correlated samples; $p<0.05$ and $p<0.01$ respectively), possibly indicating postural differences.

Data on muscle masses have been published by Weber (1851), Bischoff (1863), Theile (1884) and Shiino (1913), although Shiino expressed his doubts on the accuracy of measurements on cadavers which were not completely fresh. Theile published a vast amount of data on the masses of body components but generally omitted the determination of bodyweight. For this reason data of only three of his specimen are reported here (Table 6). Results of Theile (specimen III and XLI) and Bischoff differed significantly with this study (T-pair, $p<0.01$). These differences disappeared when normalised for total muscle mass. Data reported by Wood et al. (1989\textsuperscript{a}) are volume data. The authors state however that muscle masses could be calculated with the use of a muscle density value of 1 gram*cm\textsuperscript{-3}.

As already mentioned in the Introduction, the purpose of this article was the description of the method of data collection and results from descriptive anatomy in the process of development of a three-dimensional model of the shoulder mechanism. Van der Helm et al. (1991) describe the method of and results on the calculation of positions of muscle and ligament attachments, muscle paths and positions of rotation centers of articulations involved.

Acknowledgements - The help of Drs. H. Schutte during the anthropometric measurements and dissection course was most appreciated. We would like to thank Prof. J.P. Clarrijs for the use of partly unpublished data of his extensive cadaver measurements and for the access to his literature.
Chapter 3

Geometry parameters for musculoskeletal modelling of the shoulder mechanism.

F.C.T. Van der Helm\textsuperscript{1}, H.E.J. Veeger\textsuperscript{2}, G.M. Pronk\textsuperscript{1}, L.H.V. Van der Woude\textsuperscript{2}, R.H. Rozendaal\textsuperscript{2}.

\textsuperscript{1}: Man-Machine Systems Group, Lab. for Measurement and Control, Dept. of Mechanical Engineering and Marine Technology, Delft University of Technology, Delft, The Netherlands

\textsuperscript{2}: Faculty of Human Movement Sciences, Free University, Amsterdam, The Netherlands.

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Abstract

A dynamic finite element model of the shoulder mechanism consisting of thorax, clavicle, scapula and humerus, is outlined. The parameters needed for the model are obtained in a cadaver experiment consisting of both shoulders of 7 cadavers. In this part, particularly the derivation of geometry parameters from the measurement data is described. The results for one cadaver are presented as a typical example.

Morphological structures are modelled as geometric forms. Parameters describing these forms are estimated from 3D position coordinates of a large number of data points on the morphological structure, using a least squares criterion. Muscle and ligaments attachments are represented as a plane or as a (curved) line. Muscle paths are determined by a geometric form of the bony contour around which the muscle is wrapped. Muscle architecture is determined by the distribution of muscle bundles over the attachment area, mapping the distribution of the origin to the insertion. Joint rotation centers are derived from articular surfaces. Hence, muscle moment arms can be calculated. The result of this study is a set of parameters for each cadaver, describing very precisely the geometry of the shoulder mechanism. This set allows positioning of muscle force vectors a posteriori, and recalculation of position coordinates and moment arms for any position of the shoulder.
I Introduction

I.1 Dynamic finite element model

The shoulder mechanism is a constellation of bones, i.e. thorax, clavicula, scapula and humerus, which forms the connection between the trunk and upper extremity. Since motions of these bones are closely related, the shoulder mechanism should be studied as a whole (Inman et al., 1944). In the late 19th and early 20th century Mollier (1899), Shiino (1913) and Hvorslev (1927) have built physical models of the shoulder mechanism, based on shoulder specimens. They replaced muscles on a cadaver by cords running from origin to insertion. Changes in muscle length could be established by changes in cord length and have been described extensively. Maximal muscle moments were calculated using the physiological cross-sectional area (PSCA) of muscles. Since then biomechanical models were used to calculate the muscle forces and moments. The models derived were large simplifications of the shoulder mechanism (De Luca & Forrest, 1973; Poppen & Walker, 1978; Bassett, 1983). The arrival of powerful computers has enabled the development of more complex musculoskeletal models, e.g. of the lower limb (Brand et al., 1982; Seireg & Arvikar, 1975; Pierrynowski, 1982) or the spine (Gracovetsky & Farfan, 1986).

Thus far, the shoulder mechanism has been studied less extensively, probably due to the 3D motions, the complex motion constraints caused by the scapulothoracic gliding plane and the number of links and muscles involved. Usually only part of the mechanism, e.g. the glenohumeral joint, have been taken into account, or the role of separate morphological structures like muscles and ligaments have been described without referring to the function of the total shoulder mechanism. However, to gain insight into the role of each morphological structure in the functioning of the total mechanism, all structures should be enclosed in one model. Therefore, a dynamic model of the shoulder mechanism has been developed using the finite element method (Werff, 1977; Werff & Jonker, 1983; Jonker, 1988; Van der Helm, 1988; Pronk, 1989; Van der Helm & Pronk, 1989). In this finite element model morphological structures are represented by elements of which the dynamic behavior is well-known. Then, despite its complexity, motion equations for the whole mechanism can be derived by simply connecting the elements.

In the finite element model of the shoulder mechanism clavicula and humerus are modelled as single rigid BEAM elements, and the scapula as a combination of two rigid BEAM elements. The sternoclavicular, acromioclavicular and glenohumeral joints are each represented by three orthogonal HINGE elements. Muscles are modelled as one or more active TRUSS or CURVED-TRUSS elements which deliver the force necessary for position changes and dynamic equilibrium. The costoclavicular, conoid and trapezoid ligaments are modelled as flexible passive TRUSS
Geometry parameters

elements, whereas the scapulothoracic gliding plane is modelled by two SURFACE elements which constrain the motions of two points at the medial border of the scapula. Finally, mass and rotational inertia of segments are represented by lumped inertia at the nodes while external forces (e.g. gravity) also act at these nodes. With above assumptions the shoulder mechanism model has seven degrees of freedom: four at the scapula and three at the humerus. For simulation the input variables of the model will be the prescribed (measured) position, velocity and acceleration of bony landmarks. The output will be the calculated stress in muscles (active TRUSS elements) using an optimization criterion.

1.2 Determination of parameters

A cadaver study has been performed in order to acquire data for inertia, geometry and muscle contraction parameters of the dynamic finite element model of the shoulder mechanism described (Veeger et al., 1991a). In general, the number and nature of parameters included in a model depend on the sort of model, the characteristics of the system to be modelled and the availability of measurement techniques to derive an accurate estimation of the parameters. In the finite element method the most important characteristics of the shoulder mechanism can be included. To describe the geometry of the shoulder mechanism, the parameters of the following morphological structures are defined:

- The location of origin and insertion of all muscles.
- The shape and position of bony contours determining the muscle path.
- The location of the attachments of the costoclavicular, conoid and trapezoid ligaments.
- The location of the joint rotation centers of the sternoclavicular, acromioclavicular and glenohumeral joints.
- The shape and position of the scapulothoracic gliding plane.
- The position of well-defined bony landmarks.

In this study, a large number of data points located on morphological structures has been measured using a new measurement instrument: the palpator (Pronk & Van der Helm, 1991; Veeger et al., 1991a). The palpator is a spatial digitizer which consists of four links connected by four hinge joints. Rotations of these joints are recorded using high-precision potentiometers and hence the position of the endpoint of the final link can be calculated. Spatial coordinates are measured by simply touching the palpator's endpoint to a point and pressing a trigger. Morphological structures can subsequently be described mathematically by geometric forms which are fitted to the data
points using a least squares criterion.

Within our dynamic finite element model the muscles are represented by active TRUSS elements, ligaments by passive TRUSS elements, whereas the joints are represented by HINGE elements and the scapulothoracic gliding plane by SURFACE elements. For each element position and orientation parameters are needed, as well as parameters describing the dynamic properties of elements. These parameters are derived from the recorded data points. For each morphological structure it will be discussed which characteristics have been implemented in the model.

1. Muscles

In the finite element model a muscle is represented by one or more active TRUSS elements. For an adequate representation of muscle action, the following items must be considered:

1) The number of force vectors (elements) as well as their position and direction.
2) Whether or not a muscle should be represented by curved muscle lines of action.
3) The contraction characteristics, i.e. characteristics affecting the magnitude of each force vector.

1.1 Number, position and direction of force vectors

Most authors (e.g. Mollier, 1899; Shiino, 1913; Hvorslev, 1927; Gracovetsky et al., 1981; De Luca & Forrest, 1973; Pierrynowski, 1982; Hogfors et al., 1987, Wood et al., 1989ab) divided large muscles in convenient portions which represent more or less functionally distinguishable parts. Each part is modelled with one muscle line of action. Large attachment sites are sometimes represented by "optically estimated centroids" (Hogfors et al., 1987). Only Scholten et al. (1982) defined an arbitrary large number of elements (18) to represent the function of the m. deltoideus. Both procedures have several disadvantages. Firstly, the correctness of the chosen points can never be ascertained afterwards. Secondly, there is no theory which defines a minimum or maximum number of force vectors representing the muscle action. In the third place the effect of the chosen location and direction of the force vector on the calculated muscle force is unknown.

When a muscle has a clearly distinguishable tendon, it can be assumed that the muscle force vector is positioned at one point representing the attachment of the tendon. However, when the muscle has a large attachment site, the position and direction of the muscle force vector are not clear. Obviously in musculoskeletal modelling, the number and position of muscle lines of action affect the force attributed to the muscle. Each muscle exerts a moment around a joint. The muscle force at the attachment is one force component of the torque couple, normally the joint reaction force is the other. However, if the attachment area of a muscle is sufficiently large and parts of the
muscle with different bundle orientations contract with different directions of force one muscle by itself can provide two force components of a torque couple and can exert a rotating moment on a bony segment. Clearly, such a muscle should not be represented by just one force vector but must be represented by two or more force vectors.

An example in the shoulder region is the m. trapezius, of which the laterally rotating moment cannot be ignored (Fig. 1). In order to represent such large muscles adequately, the total attachment area has to be mathematically described. One can not settle just for the center of the area. Then the muscle force vectors (~ active TRUSS elements) can be positioned afterwards. In the current study the attachment area of a muscle has been recorded by a large number of data points. This area is described mathematically by a geometric form (e.g. a curved line or a plane) which was fitted to data points using a least squares criterion.

The force vectors (active TRUSS elements) can then be positioned afterwards to represent muscle action. The direction of force vectors depends on the orientation of muscle bundles. Thus if a bundle orientation, i.e. the position of the origin and insertion of the bundle, is known, the direction of the force vector is clear. For this reason, during dissection a number of muscle bundles has been marked at origin and insertion with colored beads (Veeger et al., 1991a). The distribution of these beads over both attachment sites enabled the mapping of bundles from origin to insertion.

Figure 1: The lateral rotating torque of m. trapezius. When m. trapezius is represented with one force vector, the torque can never be modelled; with two or more force vectors the distribution over the attachment site and direction of the force vectors is critical for muscle force calculations.
By means of this distribution the direction of force vectors in our model can be calculated (Van der Helm & Veenbaas, 1991).

1.2 Curved muscle lines of action
For some muscles the muscle line of action can not be described by a straight line between origin and insertion, but should be defined around a bony contour. If the shape and position of this bony contour is known, i.e. measured and modelled, the curved line representing the muscle line of action can be calculated assuming that the muscle is free to shift over this surface. For instance the m. serratus anterior wraps around the thorax, which is modelled as an ellipsoid. In this approach the thickness of the muscle belly is neglected. A complete list of muscles with curved muscle lines of action and adjoining bony contours is presented in Table 1. Also presented are the geometric forms fitted to the bony contours.

1.3 Contraction characteristics
The magnitude of muscle force depends on a number of characteristics, e.g. the force-length and force-velocity relation, architecture, physiological cross-sectional area (PCSA), etc. However, those characteristics are mainly describing the maximal muscle force whereas factors affecting submaximal muscle force are not precisely known as yet, but are implicitly assumed in optimization criteria as minimal use of energy, minimal joint reaction force or minimal muscle stress. Only a few of the above mentioned characteristics can be measured easily at cadavers, i.e. PCSA and muscle weight. The PCSA is used as an estimation of the maximal muscle force, or for calculation of muscle stress (= actual muscle force divided by PCSA). It is assumed that all muscles have been measured at the same percentage of muscle optimum length (Woittiez, 1984). The effects of change of muscle length can eventually be included afterwards, assuming that the total muscle volume does not change due to muscle contraction.
Another way to describe the same characteristics is to use muscle weight for the calculation of muscle volume. When the muscle length is measured, an approximation of the PCSA can subsequently be calculated (Fick, 1911). Data of PCSA in this cadaver study have been presented in Veeger et al. (1991).
Table 1: The muscle lines of action of the following muscles are determined by a bony contour. In the last column the geometric form used to model the bony contour is presented.

<table>
<thead>
<tr>
<th>muscle</th>
<th>bony contour</th>
<th>geometrical form</th>
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<tbody>
<tr>
<td>m. serratus anterior</td>
<td>thorax</td>
<td>ellipsoid</td>
</tr>
<tr>
<td>m. deltoideus</td>
<td>tuberculum majus &amp; minus</td>
<td>sphere</td>
</tr>
<tr>
<td>m. pectoralis major</td>
<td>collum humeri</td>
<td>cylinder</td>
</tr>
<tr>
<td>m. latissimus dorsi</td>
<td>collum humeri</td>
<td>cylinder</td>
</tr>
<tr>
<td>m. teres major</td>
<td>collum humeri</td>
<td>cylinder</td>
</tr>
<tr>
<td>m. infraspinatus</td>
<td>caput humeri</td>
<td>sphere</td>
</tr>
<tr>
<td>m. supraspinatus</td>
<td>caput humeri</td>
<td>sphere</td>
</tr>
<tr>
<td>m. subscapularis</td>
<td>caput humeri</td>
<td>sphere</td>
</tr>
<tr>
<td>m. biceps, caput longum</td>
<td>sulcus intertubercularis</td>
<td>line</td>
</tr>
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</table>

attachment area, one (passive) TRUSS element is assumed to represent the force vector of each ligament. All ligament attachments are represented by the centroid of the attachment, and the ligament line of action by a line connecting the centroids of origin and insertion.

3. Joints

In the finite element model, each joint is represented as three orthogonal HINGE joints with zero length, thus behaving as a ball-and-socket joint, with three degrees of freedom and no translations. The joint surfaces of sternoclavicular, acromioclavicular and glenohumeral joint have been measured in order to estimate the rotation center. The glenohumeral joint can be represented as a perfect ball-and-socket joint with no translation of the rotation center of the humerus with respect to the scapula (Van der Helm et al., 1989). Its rotation center is located at the center of a sphere fitted to the glenoid, using a radius equal to the radius of the humeral head.

Both the sternoclavicular and acromioclavicular joint do not resemble ball-and-socket joints. However, considering the capsule of both joints translation of the articular surfaces with respect to each other will most likely be restricted to a few millimeters. Hence, rotation will be the main degree-of-freedom of the joints. If translation is neglected, the rotation center can maximally shift from one edge of the articular surface to the other. Consequently, half the size of the intersection of the articular surface will be the maximum error in the estimation of the rotation center. Considering the small rotations of the joints (Pronk, 1987) the effect of these assumptions for the
model will be small.

It is assumed that the medial border of the scapula is always connected to the thorax during elevation of the humerus, due to the action of m. serratus anterior and m. rhomboideus (Mollier, 1899). The so-called scapulothoracic gliding plane has been modelled as an ellipsoid (Pronk & Padt, 1986). This "joint" is represented by two SURFACE elements, each constraining a point of the medial border of the scapula to an ellipsoid.

4. Input and output of the model

Input variables for a simulation of the model are the positions, velocities and accelerations of bony landmarks (Pronk, 1987). Hence in the model, positions of these landmarks are included in order to relate observed motions to the geometry of the shoulder mechanism. Output of the model is the magnitude of muscle forces. Unfortunately, this output can not be compared with direct measurements of muscle forces. Hence, the model can be validated only in a qualitative sense by comparison with the EMG of shoulder muscles for several positions and loads of the shoulder mechanism.

II Data processing

The methodology of the data collection has been described in Veeger et al. (1991). Seven cadavers, selected on availability, were measured, totalling 14 shoulders. Three-dimensional coordinates of data points have been defined with respect to a global coordinate system with its origin at the incisura jugularis and the orientation of coordinate axes parallel to the main anatomical axes (x-axis: left to right; y-axis: caudal to cranial; z-axis: ventral to dorsal).

The position of an anatomical structure has been described by a geometric form, e.g. a line or a surface, fitted to the recorded data points. In this study the following geometric forms have been used: a plane, a sphere, an ellipsoid, a (curved) line and a cylinder. The approach is the same for each form: Parameters were estimated by minimizing the distance between data points and form with a least squares criterion.

In Tables 2 to 5 the morphological structures that have been measured are listed. These consist of 16 muscles, 3 ligaments, 6 joint surfaces, 4 bony structures which define muscle paths, and bony landmarks for comparison with in-vivo movement registration. Also for each structure the descriptive geometric form is shown. Each form is chosen with three contradictory objectives: To minimize the number of parameters (i.e. a low-order approximation), to adequately describe the structure by selecting a similar form and to minimize the residual error.
Table 2: Description of the morphological structures of the thorax by geometric forms, i.e. a point, (curved) line, plane, sphere, cylinder or ellipsoid (in mm).

- $\bar{e}$: mean residual error.
- $t_x$, $t_y$, $t_z$: parameters of the $t$-polynome of $x$-, $y$- and $z$-coordinates. The order of the line is equal to the number of parameters minus one.
- $n_x$, $n_y$, $n_z$: normalized normal vector to the plane.
- $c_x$, $c_y$, $c_z$: coordinates of the centroid.
- $w_x$, $w_y$, $w_z$: distribution (1st order moment) of the attachment over the plane.
- $m_x$, $m_y$, $m_z$: coordinates of the center of the sphere (tables 3 and 4).
- $r$: radius.
- $x_0$, $y_0$, $z_0$: position vector of the central axis of the cylinder ($z_0 = 0$).
- $d_x$, $d_y$, $d_z$: normalized direction vector of the central axis of the cylinder.
- $a_x$, $a_y$, $a_z$: length of the $x$-, $y$- and $z$-axis of the ellipsoid.

### Plane

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<th>$n_y$</th>
<th>$n_z$</th>
<th>$c_x$</th>
<th>$c_y$</th>
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<td>1.7</td>
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Table 3: Description of the morphological structures of the clavícula by geometric forms (see Table 2 for explanation).

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For the parameter estimation of a plane the total linear least squares (TLLS) technique has been used (Van Huffel & Vandewalle, 1985). This technique fits a lower dimensional subspace, in this case a plane, to the data points with a noise distribution over each coordinate, as distinct from an ordinary least squares technique which will attribute all the noise to one coordinate. The TLLS technique is only suitable if the mathematical description is linear. (Appendix A). For each plane-shaped attachment the centroid \( C(c_x, c_y, c_z) \) and the distributions \( w_x \), \( w_y \) and \( w_z \) in \( x \)-, \( y \)- and \( z \)-direction have been calculated. The distribution has been calculated as a first-order moment around the centroid, e.g.:

\[
\begin{align*}
    \ w_x &= \frac{1}{A_{xy}} \int_{x_{\min}}^{x_{\max}} (y_{\max}(x) - y_{\min}(x)) \cdot |x - c_x| \, dx \\
    \ &= \frac{1}{A_{xz}} \int_{x_{\min}}^{x_{\max}} (z_{\max}(x) - z_{\min}(x)) \cdot |x - c_x| \, dx.
\end{align*}
\]
### Table 4: Description of the morphological structures of the scapula by geometric forms (see Table 2 for explanation).

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Table 5: Description of the morphological structures of the humerus by geometric forms (see Table 2 for explanation).

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where $A_{xy}$ is the surface area of the attachment projected to the $xy$-plane, $x_{\text{min}}$ and $x_{\text{max}}$ are the minimum and maximum values of the contour and $y_{\text{min}}(x)$ and $y_{\text{max}}(x)$ are the minimum and maximum $y$-values for a certain $x$. Similar notations are used for $z$. Distributions $w_y$ and $w_z$ are calculated analogously.

For the non-linear descriptions, e.g. in case of a sphere, cylinder or ellipsoid, a somewhat different approach has been used. This approach consisted of a first approximation of parameters using a linear approach, followed by an iterative Gauss-Newton method to update the parameters until the residual error was minimal. In Appendix B the fitting of a sphere is described, in Appendix C the cylinder and in Appendix D the ellipsoid.

Appendix E describes the modelling of muscle attachments by a curved line. The $x$-, $y$- and $z$-coordinates are expressed as functions of the variable $t$, $t \in [0, 1]$. The parameters of these functions have been estimated for each coordinate separately using a least squares criterion. Muscle architecture, i.e. bundle direction, has been described by mapping the origin of muscle bundles to their insertion. Before exarticulation and the coupled cutting of muscle bellies, the origin and insertion of a number of muscle fibers have been marked with beads of the same color. In this way the distribution of muscle bundles over attachment areas was known so that the bundle orientation could be calculated. For each marked bundle $i$ the value of variable $t$ of origin $t_0(i)$ and insertion $t_1(i)$ is calculated. Next $t_0(i)$ has been fitted to $t_1(i)$ with a polynomial function, thus allowing for each attachment point at the origin the connected point at the insertion to be interpolated. So for each bundle the orientation could be calculated.

### III Results

In the cadaver study both shoulders of 7 cadavers have been measured, totalling 14 shoulders. It is hard to tell beforehand whether it will be possible to develop one general model with averaged model parameters or that parameters have to be set for every individual shoulder. In contrast to Hogfors et al. (1987), scaling of geometry parameters with respect to bony dimensions did not result in a sufficient similarity of attachment shape. For a number of cadavers intra-individual (left-right) differences were even as large as inter-individual differences. In our opinion comparison of geometry parameters without considering the functional morphology is useless which implies that in the scope of this article no further comparisons are discussed. Hence, it was difficult to present all results of this cadaver experiment in an interpretable fashion and therefore it has been chosen to present in Tables 2 to 5 the parameters describing the geometry of one cadaver (i.e. cadaver K2) with respect to its global coordinate system. Considering the inertia, geometry and muscle
contraction parameters this cadaver seemed a more or less median cadaver (Veeger et al., 1991\textsuperscript{a}). More data for other cadavers are available on request. It should be kept in mind that the parameters presented are those of the position in which the cadaver was measured, not the anatomical or another well defined position. However, since the position of the bones can be used as input for the model this undefined position is not important. For each subsequent position the coordinates can be calculated again. In Tables 2 to 5 the geometry parameters of the right shoulder of cadaver K2 are shown for the thorax, clavicle, scapula and humerus, respectively. The morphological structures are clustered with respect to their geometric forms, in sequence a plane, line, point, sphere, ellipsoid and cylinder.

The articular surfaces of the glenohumeral joint are modelled as a sphere. Parameters of the sphere are the coordinates of its center and its radius. As shown, the center and radius are different for the humeral head and the glenoid, due to method artefacts (estimated transformation functions!) and restrictions on the set of data points (small area of the glenoid). Therefore, another fitting procedure has been performed for the center of the glenoid, using the radius of the humeral head as radius. It can be argued that the center of this sphere is the rotation center of the humerus with respect to the scapula (Van der Helm et al., 1989), see Fig. 2.

The articular surfaces of the sternoclavicular and acromioclavicular joints have been modelled as a plane. Neglecting (small) translations in the joint, the shift of the rotation center from one edge of the articular surface to the other (maximal ± 20 mm) and the effect of an articular disc (Fig. 3),

![Glenohumeral Joint Diagram](image)

**Figure 2:** The glenohumeral joint consists of the glenoid and the humeral head. The rotations center of the humerus with respect to the scapula is the center $M_{gl}$ of the sphere fitted to the glenoid, using the radius $R_h$ of the humeral head as radius.

- $M_h$: center of the sphere fitted to the humeral head.
- $R_h$: radius of the sphere fitted to the humeral head.
- $M_{gl}$: center of the sphere fitted to the glenoid with $R_h$ as radius.
the centroid of the plane fitted to the proximal joint surface can be assumed to be the rotation center of the joint. The orientation of the plane is important for decomposing joint reaction forces into a compressing component (perpendicular to the articular surface) and shearing components (parallel to the articular surface).

In the finite element model the scapulothoracic gliding plane is represented as a surface element: a point of the scapula following the surface of the thorax. In this study the scapulathoracic gliding plane has been modelled by an ellipsoid with its axes parallel to the main anatomical axes. The parameters of this ellipsoid are then the coordinates of the center of the ellipsoid and the length of the x-, y- and z-axis (Fig. 4). For stability of the iteration process the x-coordinate of the center of the ellipsoid is in the midsagittal plane ((x=0)-plane).

A flat attachment site of a muscle or ligament is modelled as a plane. Parameters describing an attachment site are:
- the normal vector to the plane \((n_x, n_y, n_z)\)
- the centroid of the attachment projected on the plane \((c_x, c_y, c_z)\)
- the distribution (first-order moment) of the attachment site in x-, y- and z-direction \((w_x, w_y, w_z)\)
- the mean residual error \(\bar{e}\) of the least squares fit of the data points to the plane.

**Example 1:** Using normal vector \(\mathbf{N}(n_x, n_y, n_z)\) and centroid \(\mathbf{C}(c_x, c_y, c_z)\) an attachment plane can be described by
\[
n_x c_x + n_y c_y + n_z c_z + D = 0;
\]
from which \(D\) can be calculated. Distributions \(w_x, w_y\) and \(w_z\) show the size of the attachment site in x-, y- and z-direction, respectively. The mean residual error is an indication of the correctness of the chosen model. Fig. 5 shows the results of modelling the origin of \(m.\ subscapularis\).

![Figure 3: The acromioclavicular joint between the clavicle and scapula. If the small translations in the joint are neglected, then the rotation center of the scapula with respect to the clavicle shifts from A to B, for upward and downward rotation, respectively. The error of the estimated rotation center \(C\), the centroid of the clavicular joint surface, will be at most half the diameter of the joint surface.](image-url)
Other muscle attachments are modelled with a line parameterized as a function of variable \( t \) (\( x(t), y(t), z(t), 0 \leq t \leq 1 \)). The parameters describing the polynomial in \( t \) are, respectively, the vectors \( t_x, t_y \) and \( t_z \). The order of the line equals vector dimension minus 1. When a muscle attachment is modelled as a point (e.g. in the case of a tendon) simply the mean of the recorded data points can be presented.

Example 2: Using vectors \( t_x, t_y \) and \( t_z \) an attachment line can be described as a function of \( t \) \((0 \leq t \leq 1)\):

\[
\begin{align*}
x &= t_x(1) + t_x(2).t + t_x(3).t^2 + \cdots + t_x(n).t^{n-1}; \\
y &= t_y(1) + t_y(2).t + t_y(3).t^2 + \cdots + t_y(n).t^{n-1}; \\
z &= t_z(1) + t_z(2).t + t_z(3).t^2 + \cdots + t_z(n).t^{n-1};
\end{align*}
\]

where \( t_x(1) \) is the first element of vector \( t_x \), etc. Start and end of the attachment line is given by \( t=0 \) and \( t=1 \), respectively. Fig. 6 shows the results of modelling the origin of m. serratus anterior.

Beads were used to mark origin and insertion of some muscle bundles. For a number of corresponding beads \( x \)-, \( y \)- and \( z \)-coordinates of origin and insertion were recorded. After a line was fitted to the attachment, a \( t \)-value has been assigned to each point at the attachment which was marked with a bead. Hence, for \( N \) beads \( t \)-values at the origin \( t_{ij} \) (\( j = 1, \ldots, N \)) and \( t \)-values at the insertion \( t_{ij} \) are known. Next, \( t_{uj} \) is fitted to \( t_{ij} \). This map can be used to interpolate corresponding \( t \)-values of all fibers in between. Illustrations of such maps are presented in Figs 7A and 7B.

![Diagram](image)

**Figure 4:** The parameters of the ellipsoid fitted to the scapulothoracic gliding plane are the coordinates of the center of the ellipsoid \((M_x, M_y, M_z)\) and the axes of the ellipsoid \((a_x, a_y, a_z)\).
maps like Figs 7A and 7B in combination with Tables 2 to 5, the attachment site of any muscle bundle can be reconstructed.

Example 3: Starting with a t-value \( t_{ok} \) (0 ≤ \( t_{ok} \) ≤ 1) at the origin of muscle bundle \( k \) the corresponding value \( t_{ik} \) at the insertion is found using the map from origin to insertion (Figs 7A and 7B). As explained in Example 2, \( t_{ok} \) and \( t_{ik} \) can be used to calculate the x-, y- and z-coordinates of origin and insertion of bundle \( k \), respectively, using the appropriate vectors \( t_x \), \( t_y \) and \( t_z \) as presented in Tables 2 to 5. Path of the bundle is determined by a straight line between origin and insertion, or, in case a bony contour intervenes (Table 1), the shortest path around the bony contour is calculated.

**Figure 5:** Origin m. subscapularis at the scapula.

a: projection frontal plane.
b: projection transversal plane.
c: projection sagittal plane.
x...x: recorded datapoints connected by straight lines.
+: centroid.
o: distribution with respect to centroid.
Total force of a muscle is the summation of all forces of its bundles. Using data of this study a general theory is developed for representing the mechanical effect of the whole muscle with an adequate number of muscle lines of action (Van der Helm & Veenbaas, 1991). Fig. 8A shows the position and direction of a large number of muscle bundles of m. trapezius, Fig. 8B shows the position and direction of six muscle lines of action which are necessary to represent the mechanical effect.

Some bony contours determine the muscle path (Table 1). The thorax is modelled as an ellipsoid,

Figure 6: Origin m. serratus anterior at the thorax (1 to 9: origin and insertion of distinct heads).
  a: projection frontal plane.
  b: projection transversal plane.
  c: projection sagittal plane.
  - - -: fitted to the datapoints.
  ---: connecting datapoints.
the tuberculum majus combined with the tuberculum minus as a sphere, the caput humeri also as a sphere and the humeral shaft as a cylinder. The parameters of the cylinder are the direction vector of the axis \((d_x, d_y, d_z)\), the position of the axis \((x_0, y_0, (z_0 = 0))\) and the radius \(r\) (Fig. 9). The muscle path is defined as the shortest line between origin and insertion which will be a straight line if no bony contour intervenes. Else, the shortest path around the bony contour can be calculated from the position coordinates of origin and insertion and the mathematical description of the surface of the bony contour.

In the finite element method each muscle line of action is represented by an active element. Origin and insertion are rigidly connected with any local coordinate system of the bone. Motion of the local coordinate system is determined by motion of bony landmarks which are also rigidly attached.

Figure 7: Mapping of the \(t\)-values of the origin \((t_0)\) to the \(t\)-values of the insertion \((t_i)\), in order to obtain the direction of the muscle bundles. \(x\): \(t\)-values of recorded beads.
A: \(m.\) serratus anterior.
B: \(m.\) pectoralis major.
Hence, starting from the position in which the cadaver was measured, new positions can easily be calculated using rigid body kinematics. In each new position muscle lines of action are recalculated whether or not wrapped around bony contours which in turn are rigidly attached to the local coordinate systems.

No explicit data about muscle length and moment arms are presented in this article, since these data will change with each position of the shoulder mechanism. It is more important to consider that muscle length and moment arms can be derived for any position, using a model of the shoulder mechanism.

IV Discussion

In order to obtain 3D coordinates the palpator has been used (Pronk & Van der Helm, 1991). This instrument permitted a new experimental method. Firstly, positions of bony segments (thorax, clavicula, scapula and humerus) were measured in a global coordinate system, by means of six screws fastened to the bones. After dissection and exarticulation a large number of data points at each morphological structure was measured in a local coordinate system. The configuration of screws has been used to calculate a transformation function from the local to the global coordinate system. This transformation introduces an additional error (residual error $\epsilon$ of the estimation of the transformation function) in repositioning the bony segments with respect to each other, about 1-3 mm for each segment (Veeger et al., 1991\(^a\)).

Using the palpator a large number of data points could be recorded. From the data points collected mathematical descriptions of each morphological structure can be derived. 3D geometric forms are

![Diagram](image.png)

**Figure 8:** Representation of the mechanical effect of the scapular part of m. trapezius by multiple muscle lines of action:

A: 20 muscle lines of action.

B: 6 muscle lines of action (as used in the finite element method).
used for mathematical description. The choice of a certain form is rather subjective, mainly
directed by an optical resemblance. The residual error of the fitting procedure which is small for
all structures, is the only justification afterwards (see Tables 2 to 5).
The experimental method of this study enabled recording of all important morphological structures:
muscle attachments, joint surfaces, bony contours, bundle orientations, ligament attachments and
bony landmarks. From position and shape of the articular surface, the position of joint rotation
centers as well as the direction of compression and shear joint reaction forces can be calculated.
In our finite element model the medial border of the scapula is constrained to follow the
scapulothoracic gliding plane which is modelled as an ellipsoid. For each muscle the position of
the whole attachment site of origin and insertion is modelled. Combined with the bundle orientation
and, when applicable, the bony contour underneath a muscle, a number of (curved) muscle lines
of action is calculated, which adequately represent the muscle action for each position of the
muscle. In this way the muscle moment arm can be calculated. The same approach as for muscles
is applicable for ligaments, except that dimensions of ligaments are so small that one line of action

\[ (x_0, y_0) \]

\[ (Z = 0) \text{ PLANE} \]

\[ d_x \]

\[ d_y \]

\[ R \]

Figure 9: The parameters of the cylinder fitted to the collum humeri are the normalized direction vector
\((d_x, d_y)\) of the central line through the cylinder, the intersection of the central line and the \((z = 0)\)-
plane \((x_0, y_0)\) and the radius \(r\) of the cylinder.
is considered to be sufficient. Combined with mass and moment of inertia of segments an inverse
dynamic analysis can be performed, using PCSA as an estimate of the maximal muscle force and
a suitable optimization criterion to calculate muscle force. In addition, a number of bony landmarks
is recorded. Motions of the shoulder mechanism were recorded using these bony landmarks (Pronk,
1987). The advantage of such an approach in musculoskeletal modelling is clear: For each new
position of the bony segments the position of the rotation center of the joints, the muscle
attachments and the bony contours can be calculated again. Hence, the new muscle lines of action
and moment arms can be assessed for each position of the shoulder mechanism. A disadvantage
of this method is that for curved muscle lines of action the diameter of muscles is neglected.
However, the muscles in the shoulder mechanism concerned are flat muscles, so the systematic
error is likely to be small.

The shoulder mechanism of each individual performs more or less the same functions. It could be
that the same morphology is a basis for these functions. It is however possible that the same
functions can be performed with different mechanisms. This would imply that the shoulder
mechanism can only be compared on a functional level, and not on the underlying morphology.
The geometry parameters derived in this study for both shoulders of seven cadavers, were not
sufficiently similar to allow them to be presented as averages over all cadavers.
Therefore, we incline towards the statement that the geometry of each muscle should be regarded
in comparison with the geometry of the whole shoulder, which means that only functional aspects
of morphology can be compared.

V. Concluding remarks

1. A complete parameter set, consisting of inertia, geometry and muscle contraction parameters,
   of the shoulder mechanism has been derived for seven cadavers. This enables comparisons
   between individuals on a morphological and functional level.

2. The data are mathematically described by geometric forms which are suited to be incorporated
   in a dynamic finite element model of the shoulder mechanism.

3. Through the assessment of muscle architecture, i.e. bundle direction, and the description of the
   whole attachment area an adequate number of force vectors can be positioned in order to
   represent muscle action.

4. Since the location of muscle attachments, rotation centers and bony contours can be recalculated
   after motion of the bones, moment arms and mechanical effects of muscles can be determined
   for any position of the shoulder mechanism.
Appendix A

Parameter estimation of a plane.

The general mathematical description of a plane in 3-D space is

$$Ax + By + Cz + 1 = 0 .$$  \hspace{1cm} (A1)

Fitting a plane to a number of measured data points is defined as minimizing the distance of the data points to the plane with a quadratic criterion. By virtue of the fact that the equation of a plane is linear, the total linear least squares (TLLS) technique for parameter estimation can be used very elegantly (van Huffel & Vandewalle, 1985). One important feature of this technique is that the noise will be distributed over all position coordinates of the data points, in contradiction to an ordinary linear least squares solution which will attribute all the measurement noise to one coordinate.

The equation mentioned above can be rewritten in matrix form as:

$$\begin{bmatrix}
  x_1 & y_1 & z_1 \\
  \vdots & \vdots & \vdots \\
  x_N & y_N & z_N \\
\end{bmatrix}
\begin{bmatrix}
  A \\
  B \\
  C \\
\end{bmatrix}
\begin{bmatrix}
  x_1 \\
  y_1 \\
  z_1 \\
\end{bmatrix}
\begin{bmatrix}
  10^6 \\
  10^6 \\
  10^6 \\
\end{bmatrix}
= \mathbf{0} \quad \text{or} \quad \mathbf{Z}_N \mathbf{\theta} = \mathbf{0} ; \hspace{1cm} (A2)
$$

where

$(x_i, y_i, z_i)$ are the position coordinates of data point $i$, $i = 1 \ldots N$;

$\mathbf{Z}_N$: matrix of $N$ data points;

$\mathbf{\theta}$: parameter vector $(A, B, C, 10^{-6})$.

In general, this equation will not have a solution for parameter vector $\mathbf{\theta}$. Therefore, $\mathbf{Z}_N$ will be slightly deviated to $\hat{\mathbf{Z}}_N$ to get a solution for $\mathbf{\theta}$.

$$\hat{\mathbf{Z}}_N \mathbf{\theta} = \mathbf{0} \quad \text{or} \quad \begin{bmatrix}
  \hat{x}_1 & \hat{y}_1 & \hat{z}_1 \\
  \vdots & \vdots & \vdots \\
  \hat{x}_N & \hat{y}_N & \hat{z}_N \\
\end{bmatrix}
\begin{bmatrix}
  A \\
  B \\
  C \\
\end{bmatrix}
\begin{bmatrix}
  10^6 \\
  10^6 \\
  10^6 \\
\end{bmatrix}
= \mathbf{0} ; \hspace{1cm} (A3)
The TLLS technique will minimize the Frobenius norm

\[ \| Z_N - \hat{Z}_N \|_F = \sqrt{\sum_{ij} (Z_{N_{ij}} - \hat{Z}_{N_{ij}})^2} . \]  \hspace{1cm} (A4)

The last column of \( Z_N \) is scaled to an arbitrarily large value, i.e. \( 10^6 \), to minimize the attribution of noise to this column. The parameter vector \( \varrho \) is obtained by the TLLS technique as follows:

The singular value decomposition (SVD) is denoted by

\[ \text{SVD}(Z_N) = U \cdot \Sigma \cdot V^T ; \]  \hspace{1cm} (A5)

where

- \( U \): left singular vectors;
- \( V \): right singular vectors;
- \( \Sigma \):

\[ \Sigma = \begin{bmatrix} \text{diag} (\sigma_1 & \ldots & \sigma_4) \\ 0 \end{bmatrix} . \]

\( \text{diag}(\sigma_1 \ldots \sigma_4) \): diagonal matrix with the singular values, \( \sigma_1 \geq \sigma_2 \geq \sigma_3 \geq \sigma_4 \).

To get a minimal deviation \( [\hat{Z}_N] \), \( \sigma_4 \) is replaced by zero. Then

\[ \hat{Z}_N = U \cdot \hat{\Sigma} \cdot V^T ; \]  \hspace{1cm} (A6)

\[ \hat{Z}_N \cdot \varrho = \Omega ; \]  \hspace{1cm} (A7)

where

\[ \hat{\Sigma} = \begin{bmatrix} \text{diag}(\sigma_1 & \ldots & \sigma_3 & 0) \\ 0 \end{bmatrix} . \]

The TLLS solution for \( \varrho \) is obtained by scaling the last row \( R_4 \) of \( V^T \) until its last component is \( 10^{-6} \).

\[ \varrho = \left( \frac{R_4}{\text{V(4,4)}} \right) \cdot 10^{-6} . \]  \hspace{1cm} (A8)
Appendix B

Parameter estimation of a sphere.

In order to model the glenohumeral joint surfaces as a perfect sphere, the measured positions of about fifty points on these surfaces were fitted to a mathematical description of a sphere, by minimizing the distance $e_i$ of each point $(x_i,y_i,z_i)$ to the sphere with radius $r$ and center $(m_x,m_y,m_z)$:

$$e_i = \sqrt{(x_i - m_x)^2 + (y_i - m_y)^2 + (z_i - m_z)^2} - r .$$

Using a quadratic criterion, this yields for $N$ data points the minimization of

$$V_N(\theta, Z_N) = \frac{1}{N} \sum_{i=1}^{N} \frac{1}{2} e_i^2(\theta) ;$$

where

$\theta$ : parameter vector, containing $m_x,m_y,m_z,r$ ;

$Z^N$: set of data points $(x_i,y_i,z_i)$, $i=1, \ldots , N$.

This is a non-linear least squares problem, which can be solved by an iterative Gauss-Newton method (Ljung, 1987).

To obtain a first estimate, $\hat{\theta}_N^{(0)}$ of the parameter vector, the following description of a sphere is used:

$$A(x^2 + y^2 + z^2) + Bx + Cy + Dz + 1 = 0 ,$$

which is linear in the parameters. The parameters of this equation can be solved by using the total linear least squares method (Van Huffel & Vandewalle, 1985), see appendix A.

Appendix C

Parameter estimation of an ellipsoid.

Former studies have shown that the thorax, and in particular the scapulothoracic gliding plane, can be modelled by an ellipsoid (Entken & Wijgergangs, 1984; Schultz, 1985; Pronk & Padt, 1986). In this study the scapulothoracic gliding plane as well as the muscle path of the m.serratus anterior are approximated by an ellipsoid. To find a best fit with the ellipsoid the distance between the measured data points and the ellipsoid is minimized using a quadratic criterion. For convenience matters in calculation the axes of the ellipsoid are placed in the coordinate planes. To obtain a
stable algorithm the x-coordinate of the center of the ellipsoid is chosen to be in the \((x = 0)\)-plane.

The equation of such an ellipsoid is:

\[
\left( \frac{x - m_x}{a_x} \right)^2 + \left( \frac{y - m_y}{a_y} \right)^2 + \left( \frac{z - m_z}{a_z} \right)^2 = 1;
\]

(C1)

where

\(x, y, z\): coordinates of points on the ellipsoid

\(m_x, m_y, m_z\): coordinates of the center of the ellipsoid \((m_x = 0)\)

\(a_x, a_y, a_z\): axes of the ellipsoid in resp the \(x, y\) and \(z\) direction

The shortest distance between a point \(i (x_i, y_i, z_i)\) and the ellipsoid is along a normal of the ellipsoid through \(i\), intersecting the ellipsoid at \(i' (x'_i, y'_i, z'_i)\). Then the line through the points \(i\) and \(i'\) is described as:

\[
\begin{bmatrix}
  x_i \\
  y_i \\
  z_i
\end{bmatrix}
= \begin{bmatrix}
  x_i' \\
  y_i' \\
  z_i'
\end{bmatrix} + \mu_i \begin{bmatrix}
  2(x'_i - m_x) \\
  2(y'_i - m_y) \\
  2(z'_i - m_z)
\end{bmatrix}.
\]

(C2)

Combining Eq. (C2) with Eq. (C1) results in function \(F(\mu_i)\)

\[
F(\mu_i) = \left( \frac{x_i - m_x}{a_x + \frac{2\mu_i}{a_x}} \right)^2 + \left( \frac{y_i - m_y}{a_y + \frac{2\mu_i}{a_y}} \right)^2 + \left( \frac{z_i - m_z}{a_z + \frac{2\mu_i}{a_z}} \right)^2 - 1.
\]

(C3)

For a given set of ellipsoid parameters and a data point \(i (x_i, y_i, z_i)\), \(\mu_i\) can be solved from this equation when \(F(\mu_i)\) is zero, and with Eq. C2 the coordinates of point \(i'\) can be calculated. Consequently, the distance between \(i\) and \(i'\) is the distance from \(i\) to the ellipsoid.

An initial approximation of \(\mu_i\) is achieved by a stepping routine. \(\mu_i\) is solved numerically by an iterative Taylor expansion:
\[ \mu_i^{(j+1)} - \mu_i^{(j)} + \left[ \frac{\delta F(\mu_i)}{\delta \mu_i} \right]^{-1} \cdot \Delta F(\mu_i) + \frac{1}{2} \left[ \frac{\delta^2 F(\mu_i)}{\delta \mu_i^2} \right]^{-1} \cdot \Delta F(\mu_i)^2; \]  

(C4)

where
\[ \Delta F(\mu_i) = [F_0(\mu_i) - F(\mu_i)]; \]
\[ F_0(\mu_i) = 0; \]

\( \mu_i^{(j)} \) : the j-th update of \( \mu_i \).

An estimation of the center and the axes of the ellipsoid is obtained by minimizing the criterion:

\[ V_N(\theta, Z_N) = \frac{1}{N} \sum_{i=1}^{N} \frac{1}{2} e_i^2(\theta); \]  

(C5)

where
\( \theta \) : parameter vector, containing \( m_y, m_z, a_x, a_y, a_z; \)
\( Z_N \) : set of data points.

The next procedure is a modification of the Gauss-Newton method as described in appendix B. The approximation of the Hessian \( V''(\hat{\theta}_N^{(j)}, Z_N) \) happens to be close to singular because the data are not informative enough (only data of one side of the ellipsoid are obtained!). One way to overcome this problem is the application of the Levenberg-Marquardt procedure (Levenberg, 1944; Marquardt, 1963). In addition, a turbo-parameter \( f \) is used to enforce a line search as long as the criterion \( V_N(\theta, Z_N) \) decreases. For each successful step, \( f \) is increased by a factor 2. This reduces the number of calculations of the gradient and thus improves the speed of the algorithm:

\[ \hat{\theta}_N^{(j+1)} - \hat{\theta}_N^{(j)} - f \left[ V''(\hat{\theta}_N^{(j)}, Z_N) + \alpha \cdot \text{diag}(V''(\hat{\theta}_N^{(j)}, Z_N)) \right]^{-1} \cdot V'(\hat{\theta}_N^{(j)}, Z_N); \]  

(C6)

where
\( \alpha \) : Marquardt-parameter, initially equal to 0.01, increased by a factor 4 if a step is not successful, decreased by a factor 8 if a step is successful.

\( V'(\hat{\theta}_N^{(j)}, Z_N) \) : gradient of the criterion with respect to parameter vector \( \hat{\theta}_N^{(j)} \) at the j-th iterate.

\( V''(\hat{\theta}_N^{(j)}, Z_N) \) : Hessian of the criterion.
To obtain a first estimate $\theta_{0}^{(0)}$ of the parameter vector the following description of an ellipsoid is used:

$$Ax^2 + By^2 + Cz^2 + Dx + Ey + Fz + 1 - 0,$$

which is linear in the parameters. This equation can be solved for the parameters using the total linear least squares (TLLS) technique (van Huffel & Vandewalle, 1985), see appendix A.

$$\begin{bmatrix} x_i^2 & y_i^2 & z_i^2 & x_i & y_i & z_i & 10^6 \\ \vdots & \vdots & \vdots & \vdots & \vdots & \vdots & \vdots \\ x_N^2 & y_N^2 & z_N^2 & x_N & y_N & z_N & 10^6 \end{bmatrix} \begin{bmatrix} A \\ B \\ C \\ D \\ E \\ F \end{bmatrix} = \Omega.$$  

(C8)

Eq. (C1) can be rewritten in the form of Eq. (C7)

$$\frac{1}{a_x^2}x^2 + \frac{1}{a_y^2}y^2 + \frac{1}{a_z^2}z^2 - \frac{2m_x}{a_x^2}x - \frac{2m_y}{a_y^2}y - \frac{2m_z}{a_z^2}z + \frac{m_x^2}{a_x^2} + \frac{m_y^2}{a_y^2} + \frac{m_z^2}{a_z^2} - 1 = 0.$$  

(C9)

Appendix D

Parameter estimation of a cylinder.

One common feature of all points on the surface of a cylinder is that they have equal distance to the central axis of the cylinder. If we characterize this central axis by position vector $[x_0 \ y_0 \ z_0]^T$ and unit direction vector $[d_x \ d_y \ d_z]^T$, i.e.:

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix} = \begin{bmatrix} x_0 \\ y_0 \\ z_0 \end{bmatrix} + \lambda \begin{bmatrix} d_x \\ d_y \\ d_z \end{bmatrix},$$  

(D1)

then the distance between a point $i$ $(x_i, y_i, z_i)$ and this line is given by

$$d_i^2 = [(x_i - x_0)d_x - (y_i - y_0)d_y - (z_i - z_0)d_z]^2 + [(y_i - y_0)d_z - (z_i - z_0)d_x]^2 + [(z_i - z_0)d_x - (x_i - x_0)d_y]^2.$$  

(D2)
Geometry parameters

The direction vector \([d_x, d_y, d_z]^T\) is a unit vector, thus

\[
d_z = \sqrt{1 - d_x^2 - d_y^2}.
\] (D3)

For the position vector \([x_0, y_0, z_0]^T\) the intersection of the central axis and the \((z = 0)\)-plane is chosen, thus \(z_0 = 0\).

The distance of point \(i\) to the surface of the cylinder is expressed by

\[
e_i = d_{\hat{r}} - r - \sqrt{[(x_i - x_0)d_y - (y_i - y_0)d_x]^2 + [(y_i - y_0)\sqrt{1 - d_x^2 - d_y^2} - z_id_z]^2 + [z_i d_z - (x_i - x_0)\sqrt{1 - d_x^2 - d_y^2}]^2 - r}
\] (D4)

An estimation of the central axis and the radius of the cylinder is obtained by minimizing the criterion

\[
V_N(\theta; Z_N) = \frac{1}{N} \sum_{i=1}^{N} \frac{1}{2} e_i^2(\theta)
\] (D5)

where

\(\theta\): parameter vector, containing \(d_x, d_y, x_0, y_0, r\);

\(Z_N\): set of data points.

This non-linear least squares problem can be solved by an interactive Gauss-Newton method (see appendix B) using turbo-parameter \(f\) (see appendix C).

To obtain a first estimate \(\theta_N^{(0)}\) of the parameter vector a straight line is fitted to the data points (see appendix E). This line is the first estimate of the central axis, the mean distance to the line is an estimate of the radius.

Appendix E

Parameter estimation of a 3-D polynomial

The \(x\)-, \(y\)- and \(z\)-coordinates of a (curved) line are expressed as a polynomial in variable \(t\) with \(t \in [0,1]\)

\[
x = a_0 + a_1t + a_2t^2 + \ldots + a_nt^n;
\]

\[
y = b_0 + b_1t + b_2t^2 + \ldots + b_nt^n;
\]

\[
z = c_0 + c_1t + c_2t^2 + \ldots + c_nt^n;
\] (E1)
or in matrix form

\[ X = T \cdot \theta ; \]

\[
\begin{bmatrix}
  x_1 & y_1 & z_1 \\
  x_2 & y_2 & z_2 \\
  \vdots & \vdots & \vdots \\
  x_N & y_N & z_N \\
\end{bmatrix}
= \begin{bmatrix}
  1 & t_1 & t_1^2 & \ldots & t_1^n \\
  1 & t_2 & t_2^2 & \ldots & t_2^n \\
  \vdots & \vdots & \vdots & \ddots & \vdots \\
  1 & t_N & t_N^2 & \ldots & t_N^n \\
\end{bmatrix}
\begin{bmatrix}
a_0 & b_0 & c_0 \\
a_1 & b_1 & c_1 \\
\vdots & \vdots & \vdots \\
a_n & b_n & c_n \\
\end{bmatrix} ;
\]

(E2)

where

\[ x_i, y_i, z_i: \text{coordinates of data points } i (i = 1 \ldots N) ; \]
\[ N: \text{number of data points} ; \]
\[ t_i: \text{t-value of data point } i, \ 0 \leq t_i \leq 1 ; \]
\[ a_j, b_j, c_j: \text{parameters of a 3-D polynomial in } t \text{ of order } n (j = 0 \ldots n) ; \]

First the data points \( i \) are sorted in the sequence of the attachment. The difference between \( t_i \) and \( t_{i+1} \) is proportional to the distance between the measured data points \( p_i \) and \( p_{i+1} \) with respect to the total length of the attachment site. The parameter matrix \( \theta \) can easily be obtained by minimizing the least squares criterion:

\[ V_N(\theta, X) = (X - T \cdot \theta)^T (X - T \cdot \theta) . \]

(E3)

leading to the solution:

\[ \theta = (T^T \cdot T)^{-1} \cdot T^T \cdot X . \]

(E4)
Chapter 4
Modelling the mechanical effect of muscles with large attachment sites:
Application to the shoulder mechanism.

F.C.T. van der Helm, R. Veenbaas.


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Abstract
A general theory is described for deriving the mechanical effect of muscles with large attachment sites. In a cadaver experiment the complete attachment sites and bundle distribution of 16 muscles of the shoulder mechanism were recorded. These data were used to calculate the mechanical effect, i.e. the resulting force and moment vector, for a large number (200) and a reduced number (maximal 6) of muscle lines of action. The resulting error between both representations is small. The number of muscle lines of action in the reduced representation depends on the shape of the attachment site and muscle architecture. Important feature of the method is that the necessary number of muscle lines of action is determined afterwards. In the often used centroid line approach the number of muscle lines of action and the partitioning of muscles is determined before recording the geometry, leading to unverifiable results.
I Introduction

In biomechanical models describing forces and movements of musculoskeletal systems, the mechanical effect of muscles is represented by force vectors (e.g. De Luca & Forrest, 1973; Poppen & Walker, 1978; Schultz et al. 1982). These force vectors are derived from the muscle lines of action which are defined either by using the straight line, the centroid line or the so-called bony contour line approach (Jensen & Davy, 1975; Van der Helm, 1991c). A number of assumptions is made using the concept based on muscle lines of action. The first assumption is that the force vector derived from the muscle line of action at a certain cross-section of the muscle is representative of the forces transmitted by the muscle at that particular cross-section (Andrews & Hay, 1983). A second assumption is that there are no moments exerted by the muscle around the muscle line of action. In addition, it is assumed that no shear forces exist in the particular cross-section.

In fact, these assumptions sometimes prohibit an adequate representation of the mechanical effect of a muscle. If the muscle is modelled by a single muscle line of action, only forces can be exerted on the bone. However, some broad pennate muscles with large attachment sites consist of muscle fibers which are not attached to a tendon but insert directly into the bony elements. The resulting forces of these fibers can exert a moment at the attachment site. An example of such a muscle in the shoulder region is the m.trapezius (Fig. 1). Muscle fibers of m.trapezius run directly from origin to insertion. Fibers of the lower part are angular to the fibers of the upper part. EMG-studies have demonstrated that both parts of the m.trapezius are functionally independent, i.e. one part can be inactive while another part is active (Inman et al., 1944). If individual force vectors of the muscle fibers are summed, the result is not only a resultant force vector representing the force exerted by the muscle at the bone, but also a resultant moment vector, representing the moment exerted by the muscle at the attachment site.

Obviously, if the muscle is represented by one muscle line of action this resulting moment will not be included in the dynamic analysis of the musculoskeletal system. If the muscle is represented by more than one muscle line of action, it is important to know how these lines of action are located with respect to each other. Small changes in direction or point of application of the force vectors (amplified by the moment arms of the vectors) can have a large effect on the resulting moment vector.

In present literature, criteria for selection of the number and position of the muscle lines of action and force vectors are hardly discussed. Mostly, broad muscles are divided a priori into 'convenient' parts based on functional or anatomical criteria (Hvorslev, 1927; De Luca & Forrest,
1973; Poppen & Walker, 1978; Schultz et al., 1982; Gracovetsky et al.; 1986; McGill & Norman, 1986; Hogfors et al., 1987; Wood et al., 1989ab). Each part is then modelled with one muscle line of action. Only Scholten et al. (1982) represented the m.deltoides by an arbitrary large number of elements.

Additionally, a methodological problem arises in assessing an adequate number of muscle lines of action from three-dimensional (3D) position measurements of muscle attachments. For a correct representation each small force vector of every muscle fiber should be considered in the localisation of the resulting muscle line of action. This implies that the complete attachment site of origin and insertion should be included. In addition, the fiber distribution from origin to insertion must be considered. If the muscle line of action is to be derived from the muscle contour, as is common practice in the centroid line approach (Jensen & Davy, 1975; Nemeth & Olsen, 1985; McGill & Norman, 1986; Bassett et al., 1990) all information concerning fiber distribution is lost, and oversimplifying assumptions as described above are needed.

Purpose of this study is to develop a method for representing muscles with large attachment sites in musculoskeletal models. A method is presented to determine the number and position of muscle force vectors, capable of representing the mechanical effect of such muscles. The attachment sites and fiber distribution of 16 shoulder muscles were measured on seven cadavers. In the proposed method the complete attachment site is described mathematically, and a map of the fiber

Figure 1: Dorsal view of m. trapezius. The muscle fibers are multidirectional and exert a force as well as a moment at the scapula.
distribution from origin to insertion is derived. This map is then used to define an arbitrarily large number (about 200) of force vectors which is at least large enough in order to represent the mechanical effect of the muscle. Because computation of the mechanical effect of a muscle using 200 force vectors requires much computing time, a procedure is developed to minimize the number of force vectors while keeping the resulting error in the mechanical effect negligible.

The results of this study have been implemented in a dynamic finite element model of the shoulder mechanism (Van der Helm & Pronk, 1989). In the shoulder region a number of broad fan-shaped muscles with functionally independent parts is present. In the finite element model each muscle line of action is represented as a (straight or curved) force generating line element capable of lengthening and shortening. Each muscle element generates a force vector at both attachment sites. However, the number of muscle elements has to be sufficiently small in order to reduce the computing time of the model while still producing reliable results.

II Method

II.1 General theory

Consider a bone connected to a fixed reference frame by a broad muscle. The attachment site can be modelled in three different ways. A relatively small area can be approximated by a point. An area with a relatively large length to width ratio is approximated by a straight or curved line. A large area is approximated by a surface.

If the bone is modelled as a free body in 3D space, it has six degrees of freedom (DOF): three translations and three rotations. Which of these DOF a muscle can influence depends on the shape of the attachment sites and the direction of the muscle fibers. If the bone is connected to the rigid world by e.g. a spherical joint, the number of independent DOF is reduced by three: Only three rotations are left. But the reaction forces at the joint are still affected by the way the muscle can influence the DOF of the bone.

If the attachment sites are subdivided according to shape into points, straight lines, curved lines and surfaces, nine different combinations of attachment sites are possible (Fig. 2). Curved-line-shaped attachment sites can be regarded as surfaces: they provide at least three non-collinear attachment points. Each combination can influence a specific number of DOF. A muscle with two point-shaped attachment sites can only influence a translation in the direction of the line between the attachment points. A muscle with two surfaces or curved lines as attachment sites can influence the maximum number of six DOF of the bone.

A muscle consists of a large number of muscle fibers, grouped into motor units. The distribution
Mechanical effect of muscles with large attachment sites

of motor units allows separate muscle parts to contract more or less independently. Muscle fibers can be represented by force vectors attached to the bone, pointing towards the fiber direction. The total effect of all muscle fibers on the DOF of the bone then can be analyzed.

The force vector $\mathbf{F}_i$ of muscle fiber $i$ at the attachment site can be divided into a unit direction vector $\mathbf{f}_i$ (\(\parallel \mathbf{f}_i \parallel = 1\)) and a scalar $u_i$ representing the magnitude of $\mathbf{F}_i$:

$$\mathbf{F}_i = f_i u_i.$$  \hspace{1cm} (1)

where $f_i = [f_{ix} f_{iy} f_{iz}]^T$, in which $^T$ means transposed.

The moment vector $\mathbf{M}_i$ is defined as the crossproduct or vector product of the moment arm $\mathbf{a}_i$ and force vector $\mathbf{F}_i$:

$$\mathbf{M}_i = \mathbf{a}_i \times \mathbf{F}_i,$$  \hspace{1cm} (2)

where $\mathbf{a}_i$ is the vector between reference point $\mathbf{r}$ and point of application $\mathbf{p}_i$ of $f_i$: $\mathbf{a}_i = \mathbf{p}_i - \mathbf{r}$. This moment vector can be divided into a moment vector $\mathbf{m}_i = \mathbf{a}_i \times f_i$ and a scalar $u_i$ as well:

![Diagram](image)

**Figure 2:** Number of degrees of freedom which can be influenced (= number independent force vectors) for several combinations of attachment sites. Lines of second or higher order can be regarded similarly as planes: They provide three non-collinear points.
\[ M_i = (a_i \times f_i) \cdot u_i - m_i \cdot u_i , \]  

(3)

The resulting force vector \( \mathbf{Y}_f \) and moment vector \( \mathbf{Y}_m \) for all \( n \) muscle fibers can be written as:

\[ \mathbf{Y}_f = \sum_{i=1}^{n} f_i \cdot u_i , \]  

(4)

\[ \mathbf{Y}_m = \sum_{i=1}^{n} (a_i \times f_i) \cdot u_i - \sum_{i=1}^{n} m_i \cdot u_i , \]  

(5)

in matrix notation

\[
\begin{bmatrix}
\mathbf{Y}_f \\
\mathbf{Y}_m
\end{bmatrix} =
\begin{bmatrix}
f_1 & \cdots & f_i & \cdots & f_n \\
m_1 & \cdots & m_i & \cdots & m_n
\end{bmatrix} \cdot
\begin{bmatrix}
u_1 \\
u_1 \\
u_1 \\
u_n
\end{bmatrix},
\]

or

\[ \mathbf{Y} = \mathbf{A} \cdot \mathbf{U} \]

where

\( \mathbf{Y}_f \): resultant force vector of the muscle, in equilibrium with translational inertia forces and external forces;

\( \mathbf{Y}_m \): resultant moment vector of the muscle, in equilibrium with rotational inertia forces and external moments.

The influence of muscle geometry is accounted for by matrix \( \mathbf{A} \). Vector \( \mathbf{U} \) represents the influence of the muscle force. Given the muscle geometry matrix \( \mathbf{A} \), the resulting force and moment vector can be calculated as a function of the vector of scalar forces \( \mathbf{U} \) in the fibers.

The minimum number of force vectors necessary to represent the muscle depends on the number of DOF the muscle can influence independently. This number of elements corresponds to the number of independent equations which describes the effect of the muscle on the bone at the insertion. The number of independent equations can be determined by the rank of matrix \( \mathbf{A} \). The minimum number of elements, equal to the rank of \( \mathbf{A} \), is shown in Fig. 2 for the nine elementary combinations of attachment sites mentioned before.
II.2 Data processing

Data were obtained from 14 shoulders of 7 cadavers. The complete method of data collection has been described elsewhere (Veeger et al., 1991a; Van der Helm et al., 1991). For each muscle the procedure of data processing is divided into five steps:

1) For each muscle, 3-D coordinates of a number of points at the origin and insertion have been recorded using a so-called palpator, a kind of spatial digitizer (Pronk & Van der Helm, 1991). A mathematical description of a geometric form (i.e. a (straight or curved) line or a plane) is fitted to the data points using a least squares criterion.

2) Before exartication and the coupled cutting of muscle bellies, the origin and insertion of a number of muscle bundles have been marked with beads of the same color. For these bundles the position of origin and relating insertion have been recorded. The position data are used to derive a map for the muscle bundle distribution.

3) This map is used to locate a large number of elements (about 200). With these elements matrix $A_{200}$ and vector $Y_{200}$ are calculated.

4) The same map is used to determine a number of elements equal to the rank of matrix $A_{200}$. For these elements a new matrix $A_r$ and vector $Y_r$ are calculated where $r$ is equal to the rank of matrix $A_{200}$. Theoretically, there are solutions for matrix $A_r$, subject to the constraints posed by the attachment sites and muscle bundle distribution, which exactly fit the mechanical effect as derived using $A_{200}$. However, this would require calculation of $A_{200}$ in each position of the mechanism which is inefficient in terms of computing time. Therefore, a solution for $A_r$ is chosen in which the origin and insertion of the muscle lines of action remain fixed at the bones.

5) The result of step 4 is compared to step 3. Using error $e$ between $Y_{200}$ and $Y_r$ the usefulness of the method can be evaluated.

Two combinations of geometric shapes of attachment sites exist in the shoulder region: muscles running from one line-shaped attachment to another line shaped-attachment, and muscles running from a plane-shaped attachment to a line-shaped attachment. Attachment sites represented by a single point are considered to be represented by a zero-order line. Each combination will be treated separately. Special attention will be paid to muscles with two origins or two insertions, like m.trapezius, m.pectoralis major and m.deltoideus.
II.2.1 Two line-shaped attachment sites

The x-, y- and z-coordinates of data points of the attachment sites are expressed as polynomials in variable \( t \ (t \in [0,1]) \):

\[
\begin{align*}
  x &= a_0 + a_1t + a_2t^2 + \cdots + a_nt^n, & - x &= x(t) ; \\
  y &= b_0 + b_1t + b_2t^2 + \cdots + b_nt^n, & - y &= y(t) ; \\
  z &= c_0 + c_1t + c_2t^2 + \cdots + c_nt^n, & - z &= z(t) ;
\end{align*}
\] (6)

![Diagram A](image1.png)

![Diagram B](image2.png)

![Diagram C](image3.png)

![Diagram D](image4.png)

**Figure 3A:** Description of the origin of m. trapezius by a 4th order curved line (dorsal view), values in mm.  
—-: line connecting measured data points.  
-----: reconstruction by 4th order curved line.  

N.B. Due to the good fit the two lines are hardly distinguishable from each other.  

**3B:** Map of t-values \( t_0 \) of origin to \( t_i \) of insertion, fitted with a polynome in variable m.  

**3C:** Representation of m. trapezius by a large number of muscle elements (= muscle lines of action).  
Twenty elements are shown (dorsal view), values in mm.  

**3D:** Representation of m. trapezius by a reduced number (6) of muscle elements, values in mm.
The difference between \( t_i \) and \( t_{i+1} \) is proportional to the distance between the measured data points \( p_i \) and \( p_{i+1} \) with respect to the total length of the attachment site. Parameters of these functions are estimated for each coordinate separately using a least squares criterion (Van der Helm et al., 1991), Fig. 3A. The bundle direction is described by mapping the origin of muscle bundles to the insertion. For each marked bundle \( b \) the value of variable \( t \) of origin \( t_0(b) \) and insertion \( t_1(b) \) is calculated. Next, \( t_0 \) and \( t_1 \) are fitted with a polynomial in variable \( m \) (\( m \in [0,1] \)):

\[
\begin{align*}
    t_0 &= t_0(m) ; \\
    t_1 &= t_1(m) . 
\end{align*}
\]

(7)

For a certain \( m_k \) the corresponding values \( t_{0k} \) and \( t_{1k} \) can be calculated (Fig. 3B). These values can be used in Eq. (6) to calculate the coordinates of the origin and insertion. Between both attachments a straight or curved muscle line of action can be located. Thus, the combination of Eqs (6) and (7) yields a complete description of the muscle architecture.

The mechanical effect of each bundle can be represented by a line of action from origin to insertion. The summation of the force vectors associated with these lines of action is a good representation of the mechanical effect of the entire muscle. For reasons of efficient computation a large but finite number (200) of lines of action is chosen (Fig. 3C). These 200 lines of action are equally distributed over the line described by polynomial \( m \) (Fig. 3B) and thus not equally distributed over either origin or insertion. If a number of muscle bundles arising from the origin converges at the insertion, the polynomial \( m \) in Fig. 3B will be more horizontal. Equal distribution of lines of action over \( m \) will result in a wide-spread distribution over the origin and a bunch of lines at the insertion. The maximum force \( u_{\text{max}} \) attributed to each line of action is proportional to the relative length over the polynomial multiplied by the physiological cross-sectional area (PCSA) of the whole muscle and the maximal force per unit area (= 30 N/cm²), i.e.

\[
    u_{\text{max}} = \frac{1}{200.\text{PCSA}.30} .
\]

(8)

Subsequently, for each bundle the direction of force vector \( f_i \) and moment vector \( m_i \) can be calculated to establish matrix \( A \):

\[
    Y_{200} = A_{200}.U_{200} .
\]

(9)
with vector \( U \) chosen to be uniformly distributed: \( U_{200} = [1/200 \ldots 1/200]^T \).

The moment vector \( m_i \) has to be calculated with respect to a reference point. We have chosen the centroid \( c \) of the attachment site. This choice is discussed in Appendix A. 

As argued in Section II.1 the rank \( r \) of \( A \) is decisive for the minimum number of elements necessary to represent the muscle. The elements are equally distributed over the line described by polynomial \( m \) (Fig. 3B). Analogously to Eq. (9) a representation of the muscle with a reduced number of elements can be calculated (Fig. 3D):

\[
Y_r = A_r U_r ,
\]

with \( U_r = [1/r \ldots 1/r]^T \).

Finally, the approximation error \( e \) can be calculated between \( Y_{200} \) and \( Y_r \). For this reason both vectors are split into the force vector \( Y_f \) and moment vector \( Y_m \):

\[
Y_{200} = \begin{bmatrix} Y_{f200} \\ Y_{m200} \end{bmatrix} ; \quad Y_r = \begin{bmatrix} Y_{fr} \\ Y_{mr} \end{bmatrix} .
\]

Error \( e \) can thus be calculated:

\[
e_f = \| Y_{f200} - Y_{fr} \| ;
\]

\[
e_m = \| Y_{m200} - Y_{mr} \| .
\]

II.2.2 One line-shaped and one plane-shaped attachment site

The coordinates of a plane-shaped attachment site can be expressed mathematically by the equation:

\[
\begin{bmatrix}
x_1 & y_1 & z_1 & 1 \\
x_2 & y_2 & z_2 & 1 \\
\vdots & \vdots & \vdots & \vdots \\
x_n & y_n & z_n & 1 \\
\end{bmatrix}
\begin{bmatrix}
d_1 \\
d_2 \\
d_3 \\
d_0 \\
\end{bmatrix}
= Q ; \quad \text{or} \quad P \cdot d = Q ;
\]
with \( \mathbf{P} = [x \ y \ z \ 1] \).

The parameters describing the plane can be estimated using the total linear least squares (TLLS-)method (Van Huffel & Van de Walle, 1985; Van der Helm et al., 1991), in order to derive the projection of the data points on the plane:

\[
[\hat{x} \ \hat{y} \ \hat{z} \ 1] \cdot \mathbf{d} = 0, \quad \hat{\mathbf{P}} \cdot \mathbf{d} = 0. \tag{15}
\]

The projected data points are transformed to a local coordinate system \((u,v)\) with unit vectors arbitrarily oriented in the plane (Fig. 4A):

\[
\mathbf{R}_{uv} \cdot [\hat{x} \ \hat{y}]^T = [u \ v \ \Omega]^T; \tag{16}
\]

where \(\mathbf{R}_{uv}\) is the rotation matrix from the global coordinate system to the local coordinate system \((u,v)\).

For a number of muscle bundles the corresponding attachments at the line-shaped attachment site has been recorded. The value of the polynomial \(t\) describing the line is assigned to the data points in the \((u,v)\)-plane, and thus a \((u,v,t)\)-space is created. In this space the relation between the data points \((u,v)\) and the polynomial \(t\) is chosen to be fitted with a plane with parameter vector \(\mathbf{f} = [f_1 \ f_2 \ f_3 \ f_0]\) using the TLLS-method (Eq. 14):

\[
\begin{bmatrix}
  u_1 & v_1 & t_1 & 1 \\
  u_2 & v_2 & t_2 & 1 \\
  \vdots & \vdots & \vdots & \vdots \\
  u_n & v_n & t_n & 1 \\
\end{bmatrix}
\begin{bmatrix}
  f_1 \\
  f_2 \\
  f_3 \\
  f_0 \\
\end{bmatrix}
= \mathbf{0}. \tag{17}
\]

Thus, for each point \((u,v)\) in the plane the \(t\)-value of the corresponding point at the line-shaped attachment can be calculated. Fig. 4A shows parallel lines in the \((u,v)\)-plane consisting of points with equal \(t\)-values and thus connected to the same point on the line-shaped attachment.

A large number of force vectors can now be placed along the lines from origin to insertion. The most obvious way is to place force vectors by means of laying a grid over the plane. However, in our experience the resulting \(\mathbf{Y_m}\) is very sensitive to the position of the grid, because especially the vectors close to the contour of the attachment have a large moment arm with respect to the centroid. Slightly shifting the grid causes these vectors to be just inside or outside the contour,
largely affecting the resulting moment vector. Therefore, another approach was chosen. Each muscle fiber from a certain t-line inserts into the same point of the other (line-shaped) attachment. Hence, this is a line-point combination of attachments which can affect two DOF (see Fig. 2). Two force vectors are sufficient to represent this combination. Force vectors are positioned according to Section II.2.1. Thus, a large number of t-lines is chosen and two force vectors are placed on

Figure 4A: Mapping of t-values to the (u,v)-plane. Each line connects points at the attachment site from which the muscle fibers are inserting at the same point of the line-shaped attachment site.

4B: Reconstruction of muscle lines of action from the t-lines to the corresponding t-values at the insertion. At each t-line two muscle lines of action are connected. Total number of muscle lines of action is 200 (only 16 are shown).

4C: The plane-shaped attachment site is divided into three equal parts. Each part is represented by two muscle lines of action, attached to the centroidal t-line. Total number of muscle lines of action is 6.

4D: Representation of m. infraspinatus by a large number of muscle elements (= muscle lines of action). Twenty elements are shown (dorsal view), values in mm.

4E: Representation of m. infraspinatus by a reduced number (6) of muscle elements, values in mm.
Mechanical effect of muscles with large attachment sites

each t-line. The force attributed to each force vector is proportional to the length of the t-line. The sum of all forces is one. In this way the muscle is adequately represented by a large number of force vectors (Fig. 4B).

Some problems must be solved in order to reduce the number of force vectors. If the rank of matrix \( \mathbf{A} \) is less than six, a reduced number of muscle lines of action would be sufficient. However, the representation of the plane by t-lines requires two force vectors per t-line. Hence, an even number of force vectors is necessary in order to maintain the balancing around the center of the t-lines. Therefore, we have chosen to represent these muscles always by six force vectors, despite the rank of matrix \( \mathbf{A} \). The attachment area is thus partitioned in three equal parts with boundary lines parallel to the t-lines (Fig. 4C). The centroid t-line of each part is used to locate a cluster of two force vectors. Due to irregularities in the attachment area the length of this t-line can be extremely small or large. Therefore, the mean length of the t-lines in each part is used as length of the centroid t-line. Finally, both force vectors are placed on this t-line, according to Section II.2.1, and \( \mathbf{Y}_r \) can be calculated. Figs. 4D and 4E show the results of the procedure described above, for a large and for a reduced number of elements, respectively.

Calculation of the errors \( e_f \) and \( e_m \) between \( \mathbf{Y}_{200} \) and \( \mathbf{Y}_r \) is analogous as described is Section II.2.1.

II.2.3 Two origins or two insertions

Three muscles in the shoulder region, i.e. m.trapezius, m.pectoralis major and m.deltoides, are attached each to three bony structures, constituting either a double origin or a double insertion. Morphologically these muscles are regarded as an entity. But in order to preserve a correct force and moment calculation between the bony structures these muscles should be considered consisting of two independent parts. The procedure described in Sections II.2.1 and II.2.2 is followed considering the combined attachment site as a whole. After mapping the origin to the insertion the muscle is split into two parts using the t-value \( t_s \) ([0 \leq t_1 \leq t_s] for part one, \([t_s \leq t_2 \leq 1]\) for part two). Thus, \( t_s \) is the point where the line-shaped muscle attachment is intersected by the articulation (the acromioclavicular joint for the m. trapezius and m. deltoideus, the sternoclavicular joint for the m. pectoralis major). For each part a separate comparison between a large and a reduced number of force vectors has been made. Using \( t_s \) the corresponding m-value \( m_s \) of the map from origin to insertion has been calculated. This value \( m_s \) was used to distribute the recorded physiological cross-sectional area over both parts.
III Results

Table 1 shows the combination of line-shaped and plane-shaped attachments used for each muscle of the shoulder mechanism. In addition, the order of the line-shaped attachments, the order of the map from origin to insertion and the number of elements to represent each muscle are presented. Results are presented for all sixteen muscles of the right shoulder region of one cadaver. Additional data for the same cadaver have been published in Veeger et al. (1991) and Van der Helm et al. (1991). The mathematical description of the origin as well as the insertion, and the mean estimation error, are presented in Van der Helm et al. (1991). When fitting a (curved) line to an attachment site the mean residual error should be smaller than 4.5 mm., otherwise the order of the line is increased. Visual inspection revealed that further decreasing the maximum mean residual error causes high order curves which mainly represent irregularities in measuring muscle attachments and which do not affect the resulting mechanical effect. The same argument is true for mapping the origin to the insertion of the muscle. The mean residual error (in t-value) of the last fit is maximally 0.034 for line-line combinations, and 0.15 for line-plane combinations. The latter error is partly due to the assumption that the t-lines of the attaching fibers are parallel to each other.

The real mechanical effect of muscles with large attachment sites is impossible to measure. Therefore, we established a 'gold standard' by representing muscles by a large number (200) of force vectors, each representing more or less the effect of individual muscle bundles. A further increase in the number of force vectors did not alter the results. The mechanical effect of the whole muscle is represented as a resultant force vector $Y_f$ and a resultant moment vector $Y_m$. These vectors are not shown because they only represent the mechanical effect of a muscle with respect to its centroid which is difficult to interpret.

The modeling error $e_f$ of the resultant force vector, i.e. the difference between $Y_f^{200}$ and $Y_f$, is less than 1% for all muscles. In Figs 5A and 5B the error $e_m$, i.e. the difference between $Y_m^{200}$ and $Y_m$, is presented in relation to the magnitude of the moment vector $Y_m$. The relative error is usually less than 15%. Only for some muscles with a small $Y_m$ the relative error is larger, but the absolute contribution of this error can be neglected.

The representation of muscles with a reduced number of muscle lines of action results in a fairly accurate approximation of the mechanical effect of these muscles. For representation of the mechanical effect of twenty muscles and muscle parts of the shoulder mechanism a total of 95 muscle lines of action is needed. Position data of origin and insertion of these muscle lines of action are presented in Table 2. In combination with position data of bony contours (Van der Helm et al., 1991), the location of each muscle line of action can be calculated. Due to the complex
Table 1: Modelling of origin and insertion of shoulder muscles by line or plane.

<table>
<thead>
<tr>
<th>muscle</th>
<th>shape origin</th>
<th>shape insertion</th>
<th>order origin&lt;sup&gt;1&lt;/sup&gt;</th>
<th>order insertion&lt;sup&gt;2&lt;/sup&gt;</th>
<th>order map&lt;sup&gt;3&lt;/sup&gt;</th>
<th># elements&lt;sup&gt;4&lt;/sup&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>biceps, c. breve</td>
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<td>line</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>biceps, c. longum</td>
<td>line</td>
<td>line</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>coracobrachialis</td>
<td>line</td>
<td>plane</td>
<td>1</td>
<td>-</td>
<td>-</td>
<td>6</td>
</tr>
<tr>
<td>deltoideus, scap.</td>
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<td>plane</td>
<td>3</td>
<td>-</td>
<td>-</td>
<td>6</td>
</tr>
<tr>
<td>deltoideus, clav.</td>
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<td>plane</td>
<td>1</td>
<td>-</td>
<td>-</td>
<td>6</td>
</tr>
<tr>
<td>infraspinatus</td>
<td>plane</td>
<td>line</td>
<td>-</td>
<td>1</td>
<td>-</td>
<td>6</td>
</tr>
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<td>line</td>
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<td>line</td>
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<td>0</td>
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<td>2</td>
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</table>

1. order of polynome describing the line-shaped origin.
2. order of polynome describing the line-shaped insertion.
3. order of polynome describing the map of origin to insertion (line-shaped attachments).
4. number of elements (= muscle lines of action) used to represent the muscle in the finite element model.
Table 2: Position coordinates of combined origin and insertion of the reduced number of muscle elements (= muscle lines of action) for 16 muscles of the shoulder mechanism (in mm). Origin of the coordinate system: Incisura Jugularis, X-axis: medial-lateral, Y-axis: caudal-cranial, Z-axis: ventral-dorsal (right shoulder).

<table>
<thead>
<tr>
<th>origin</th>
<th>x</th>
<th>y</th>
<th>z</th>
<th>x</th>
<th>y</th>
<th>z</th>
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<td>71.6</td>
<td>195.5</td>
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<td>44.6</td>
<td>195.5</td>
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Figure 5: Error $e_m$ of the representation of muscle moment vectors by a reduced number of elements with respect to a large number of elements.

A: Muscle origins.

B: Muscle insertions.

\[ \square: \|Y_m\|, \text{length of muscle moment vector.} \]

\[ \Box: e_m, \text{norm of the difference between } Y_{m200} \text{ and } Y_{mr}: e_m = \|Y_{m200} - Y_{mr}\|. \]
muscle architecture in the shoulder region, it seems that an extensive musculoskeletal model is necessary.

IV Discussion
In the field of biomechanics two phases in assessing the mechanical behavior of a system can be distinguished: deriving an adequate model of the system and using the model to calculate forces, stresses and motions of the system. Once the model has been derived, i.e. the magnitude and point of application of force vectors, rotation centers, degrees of freedom of the system etc. have been determined, the subsequent calculation is fairly straightforward. Very often only the results of these calculations are discussed. However, an important item of discussion should be how system characteristics are represented by the model.

A muscle is a complex force-generating structure. Generally in a musculoskeletal model, force vectors are positioned to model the mechanical effect of a muscle. As stated in the introduction, this procedure is crucial to the results of the calculation, but thus far no theory has been presented to assure the correctness of the procedure.

In this article a general theory is proposed for derivation of the mechanical effect of muscles with large attachment sites. It is based on the recording and description of the complete attachment sites of origin and insertion, and the distribution of muscle bundles between origin and insertion. An important feature of this method is that the force vectors are positioned afterwards. The usual method is to determine the line(s) of action of a muscle, and to make assumptions about different parts of the muscle and their function before the actual recording of the muscle geometry takes place. Afterwards it is impossible to check the correctness of the assumptions.

It is important that some kind of muscle architecture, in this case the distribution of bundle directions, is included when modelling the mechanical effect of muscles. In the centroid line approach (Jensen & Davy, 1975) the angle between muscle fibers and the centroid line is ignored, violating assumptions of the approach itself (see introduction) and neglecting the fact that due to this angle, moments are exerted on the attachment site. Thus, modelling the mechanical effect of muscles with large attachment sites by the centroid line approach is prone to large and not verifiable errors. It is our opinion that in modelling the lower extremity an omission of the muscle architecture may be acceptable, due to smaller attachment sites and parallel fibered muscles, but in modelling the shoulder muscles this omission is not acceptable.

Reducing the number of force vectors is important for computational reasons. To assess the effect of this reduction we have calculated the euclidean norm of the difference between the force and
moment vectors of the gold standard and the reduced representation. The euclidean norm for force vectors, i.e. the error \( e_f \), is very small (\(< 1\%\)). Figs 5A and 5B show that the error for the moment vectors is somewhat larger, but it remains below 15\%. Smaller muscles have the largest relative error, but this error will hardly have any effect on the resulting motion of the bony element. Muscles with large attachment sites, fan-shaped muscle architecture and hence large moment vectors, e.g. m. deltoideus, m. pectoralis major and m. serratus anterior, show the largest absolute errors. However, relative to their large moment vector the error is small, indicating the success of the approximation.

If the attachment site is a straight line parallel to the x-axis, only moments around the y- and z-axis are possible. If in addition the muscle fibers are positioned in the xy-plane, only moments around the z-axis are possible. If the muscle fibers are all parallel to each other, only forces can be exerted and no moments. Thus, moments which are calculated around certain axes are only due to the angle between muscle fibers.

Naturally, the maximum number of DOF that a muscle can influence is six, equal to the full rank of matrix \( A \). The rank of \( A \) depends on the shape of the attachment sites. Fig. 2 shows that if the attachment site is modelled by a straight line, or even a single point, the rank of \( A \) is reduced. A remarkable property of the map from origin to insertion (Fig. 3B) is that if this map is approximated by a straight line, i.e. \( t_0(m) \) is equal to \( t_1(m) \) for every \( m \), the rank of \( A \) is also reduced by one, despite the order of the line-shaped attachments. If the muscle has a plane-line combination of attachment sites, the number of elements used to represent the muscle is always six, irrespective of the rank of matrix \( A \). The number of elements per muscle, equal to the rank of matrix \( A \), is presented in table 1.

Fig. 6 shows the effect of decreasing the number of elements of m. trapezius on error \( e_m \). In our method the scapular part of m. trapezius is represented by six elements, resulting in a relative error \( e_m \) of 2\%. A further increase in the number of elements is hardly effective. In contrast, a further decrease in the number of elements, e.g. to a number of two elements (like pars ascendens and pars transversalis) results in an increase in the relative error to 12\%. The effect depends on the condition of \( A \): If \( A \) is nearly singular, the effect will be small. However, it should be noted that \( A \) is position-dependent. When the attachments move with respect to each other, the full number of elements may be necessary. So it is advisable, even if \( A \) is almost singular, to use the number of elements as one can depict from Fig. 2 and the map from origin and insertion, as explained before.

Theoretically, if the number of force vectors is equal to the rank of \( A \), there is a solution for the
position and direction of these vectors which exactly represents the mechanical effect of the muscle. Thus, $e_f$ and $e_m$ will be zero. However, this solution is position-dependent, so for every subsequent position renewed elaborate calculations of the position of the force vectors would be necessary. Therefore, the position and direction of force vectors are chosen to be solely dependent on the attachment site and bundle distribution, resulting in fixed points of application of the force vectors at the bone. In many cases not all higher-order complex properties of attachment sites and bundle distribution are included which results in errors $e_f$ and $e_m$. The magnitude of these errors which is generally small, is a justification for this approach.

In this study, a uniform distribution of the muscle force over vector $\mathbf{U}$ is chosen. The results for the uniform distribution can be extrapolated to non-uniform distributions, e.g. in the case that only part of the muscle is active, since each realized mechanical effect can be simulated with the reduced number of force vectors. If the muscle lines of action are wrapped around a bony contour, matrix $A$ is affected and therefore its rank could be changed. However, it is advised to use the number of elements dependent on the shape of the attachment sites and muscle architecture, in case that during motion the bony contour moves out of the way and the muscle lines of action become straight. Errors $e_f$ and $e_m$ remain small using curved muscle lines of action.

Obviously, the choice of the rotation point is very important in assessing the moment vector. We have chosen the centroid of the attachment site, because this point is most sensitive to errors in reducing the number of vectors (see Fig. 7). Determining the moment around another point of

![Figure 6](image)

**Figure 6**: Effect of number of muscle elements (= muscle lines of action) on the calculated $e_m$ (as percentage of the length of the moment vector). Results of m. trapezius are presented (in our model represented by 6 elements).
rotation is simply done by adding a term composed of the cross product of the resultant muscle force and the position vector of the centroid with respect to the defined rotation point (Appendix A).

The results of this approach are implemented in a dynamic finite element model of the shoulder mechanism. The points of application of the force vectors at the origin and insertion (Table 2) are connected with a force-generating muscle element, comparable to a muscle line of action. This element can be either straight, if no morphological structure intervenes between origin and insertion, or curved around bony contours. The force generated by this element is directed along the element. If the bony structure moves from one position to another, the new positions of the muscle elements are calculated. Thus, for every subsequent position the mechanical effect is correctly calculated. This approach is valid for inverse dynamic as well as dynamic analysis of the mechanism. It is of special importance to this method that the elements can be considered as independent parts of the muscle.

![Diagram](image)

**Figure 7:** Effect of the choice of reference point $\mathbf{r}$ on the calculated $e_m$ (as percentage of the length of the moment vector) for m. trapezius.

- $\mathbf{r}_{t=0}$, centroid of the attachment site.
- $\mathbf{r}_{t=1}$, one end of the attachment site.
- $\mathbf{r}$, origin of global coordinate system (Incisura Jugularis).
Appendix A

The moment vector $m_i$ has to be calculated with respect to a certain reference point $r$. The difference $e$ between the true moment vector $Y_m$ and estimated moment vector $Y_m'$ is then

$$ e = |Y_m - Y_m'| = |\sum_{i=1}^{n}(a_i \times f_i).u_i - \sum_{i=1}^{n}(a'_i \times f'_i).u'_i|, $$  

(A1)

where

$a_i$ : moment arm vector of force $F_i$ : $a_i = p_i - r$ ;
$p_i$ : point of application of force $F_i$.

If we assume that the force vectors are equal ($F_i = F'_i$), only a position error in $p_i$ remains. Eq. (A1) reduces then to

$$ e = |\sum_{i=1}^{n}((a_i - a'_i) \times f_i).u_i|, $$  

(A2)

which is independent of reference point $r$.

The largest relative error $e_{rel}$ occurs if the length of $Y_m$ is as small as possible

$$ e_{rel} = \frac{|Y_m - Y'_m|}{|Y_m|}, $$  

(A3)

which is the case when the reference point $r$ is chosen to be the point where the sum of moments is zero: the force centroid point $c_f$.

$$ Y_m - \sum_{i=1}^{n}(p_i - c_f) \times F_i = 0; $$  

(A4)

$$ c_f \times \sum_{i=1}^{n}F_i = \sum_{i=1}^{n}p_i \times F_i; $$  

(A5)

$$ c_f \times F = \sum_{i=1}^{n}p_i \times F_i; $$  

(A6)
which is only solvable in exceptional circumstances in 3D space when all lines of action of $E_i$ intersect. Hence, our objective is to minimize $\| Y_m \|$ by minimizing $\Sigma p_i - r$, which is clearly the case when $r$ is chosen to be the centroid of the attachment site.
Chapter 5

A finite element musculoskeletal model of the shoulder mechanism.

F.C.T. van der Helm.


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Abstract

The finite element method described in this study provides an easy method to simulate the kinetics of multibody mechanisms. It is used in order to develop a musculoskeletal model of the shoulder mechanism. Each relevant morphological structure has been represented by an appropriate element. For the shoulder mechanism two special purpose elements have been developed: A SURFACE element representing the scapulothoracic gliding plane and a CURVED-TRUSS element to represent muscles which are wrapped around bony contours. The model contains 4 bones, 3 joints, 3 extracapsular ligaments, the scapulothoracic gliding plane and 20 muscles and muscle parts. The kinematic and (inverse) dynamic analysis of a complex mechanism as the shoulder is described. In the model input variables are the positions of the shoulder girdle and humerus and the external load on the humerus. Output variables are muscle forces subject to an optimization procedure in which the mechanical stability of the glenohumeral joint is one of the constraints. Four different optimization criteria are compared. For twelve muscles, surface EMG is used to verify the model. Since the optimum muscle length and force-length relationship are unknown, and since maximal EMG amplitude is length dependent, verification is only possible in a qualitative sense. Nevertheless, it is concluded that a detailed model of the shoulder mechanism has been developed which provides good insight in the function of morphological structures.
I Introduction

The shoulder mechanism which consists of thorax, clavicular, scapula and humerus is because of its complexity one of the most challenging systems in musculoskeletal modelling. Motions of the upper arm are the result of the simultaneous motions in the sternoclavicular, acromioclavicular and glenohumeral joints. External load and inertia forces of the upper extremity are transmitted to the thorax through the muscles crossing these joints and through the bones of the shoulder girdle. The scapula provides a moveable but stable base for a large range of motion of the humerus. Motion studies have revealed that a fixed and reproducible relation exists between motions of scapula and humerus, the so-called spinohumeral rhythm (Inman et al., 1944; Pronk, 1987).

Motions of the scapula are constrained on the one hand by the so-called scapulothoracic gliding plane which connects the medial border of the scapula to the thorax, and on the other hand by the clavicula which allows the acromion to move more or less on a sphere around the sternoclavicular joint. Due to these constraints, the shoulder girdle is a closed-chain mechanism. For this reason, models of the shoulder mechanism, or even only the glenohumeral joint, should not be restricted to two dimensions.

Furthermore, motions are constrained by three extracapsular ligaments, i.e. the costoclavicular, conoid and trapezoid ligaments. Seventeen muscles are crossing the three joints of the shoulder mechanism. They are listed in Table 1. Muscles running from thorax to scapula are crossing two joints, and muscles running from thorax directly to the humerus, like m.latissimus dorsi and m.pectoralis major, are even crossing three joints. Most of these muscles have large attachment sites, are broad and flat, and have multidirectional muscle bundles.

Due to the complexity of the shoulder mechanism, there is little or no insight in the mechanics constituting its motion behavior and the function of muscles and ligaments. Hence, interpretation of complaints in the shoulder region is hampered and does often not result in a detailed diagnosis. Surgery in the shoulder region is often based on a misjudgement of the underlying mechanics. The main motive of this study was to develop a model of the shoulder mechanism by which it will be possible to gain insight in the function of morphological structures. Results of this analysis can for instance be applied to predict optimal fusion angles for a glenohumeral arthrodesis, to improve the application of an endoprosthesis, or to develop therapies for patients with a habitual glenohumeral subluxation.

The complexity of the shoulder mechanism is probably the reason that thus far hardly any three-dimensional models were developed. Mollier (1899), Shiino (1913) and Hvorslev (1927) built physical models based on dissected specimen. Attempts for biomechanical models are scarce and
Table 1: Muscles of the shoulder mechanism.
1: partly inserting at clavicle; 2: inserting at clavicle; 3: partly originating at clavicle;
4: inserting at forearm bones.

<table>
<thead>
<tr>
<th>thoracoscapular muscles</th>
<th>thoraco humeral muscles</th>
<th>scapulo humeral muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>m. trapezius(^1)</td>
<td>m. latissimus dorsi</td>
<td>m. deltoideus(^3)</td>
</tr>
<tr>
<td>m. levator scapulae</td>
<td>m. pectoralis major(^3)</td>
<td>m. coracobrachialis</td>
</tr>
<tr>
<td>m. rhomboideus</td>
<td>m. teres major</td>
<td>m. teres minor</td>
</tr>
<tr>
<td>m. pectoralis minor</td>
<td>m. infraspinatus</td>
<td></td>
</tr>
<tr>
<td>m. subclavius(^2)</td>
<td>m. supraspinatus</td>
<td></td>
</tr>
<tr>
<td>m. serratus anterior</td>
<td>m. subscapularis</td>
<td></td>
</tr>
<tr>
<td></td>
<td>m. triceps, c. longum(^4)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>m. biceps(^4)</td>
<td></td>
</tr>
</tbody>
</table>

incomplete. Generally, two approaches to solve the mechanical problem can be distinguished. Firstly, in the Newton-Euler approach for each rigid body, i.e. the clavicle, scapula and humerus, six motion equations are derived, totalling 18 equations (Karlsson et al., 1989; Karlsson, 1990). Apparent problems in this approach are caused by the closed-chain nature of the mechanism, due to the scapulothoracic gliding plane and poly-articular muscles. Complete results of this modelling approach are not yet published. Secondly, Wood et al. (1989\(^3\)) proposed an approach based on the Lagrangian method. They included all three joints but omitted the constraint of the scapulothoracic gliding plane. The complex derivations in order to obtain motion equations are not elaborated by these authors, and are expected to be too difficult.

Nowadays a schism is encountered in deriving motion equations (Sol, 1983). Biomechanical models have grown so complex that analytical solutions are very hard to obtain, so numerical solutions are preferred with the appearance of more and more powerful computers. In order to model the shoulder mechanism a computer program SPACAR is used, based on the finite element method and specially suited for the dynamic analysis of multibody mechanisms. At each position of the mechanism motion equations are numerically derived based on deformations of the elements and displacements of its nodes. A special feature of this finite element method is that the subsequent position of the mechanism, as determined by generalized coordinates, is calculated iteratively.

In this article a summary of the finite element method will be presented. Detailed information can
be found in Werff (1977), Werff & Jonker (1983), Schwab (1983) and Jonker (1988). In the following paragraph its basic elements are described, followed by the kinematic, dynamic and inverse dynamic analysis. Subsequently, a few elements specially developed for the shoulder mechanism are introduced and discussed in more detail. The shoulder model in terms of finite elements is presented. Results of this study will be compared with EMG recordings of the shoulder.

II Finite Element Method
The gross morphological structures of the shoulder mechanism can be represented by finite elements of a simple geometry. Since the mechanical properties of these elements are well defined, the kinematic and dynamic behavior of the whole mechanism can be simulated. Characteristic for this finite element method is that deformation modes of the elements are expressed as a function of the displacements of its endpoint nodes. There are two types of nodes: position nodes and orientation nodes. Each position node has three coordinates: x, y and z. Orientation nodes have four coordinates, the Euler parameters λ₀, λ₁, λ₂ and λ₃. There are three basic types of elements: TRUSS elements, HINGE elements and BEAM elements. The TRUSS element has two position nodes and one deformation mode: elongation, calculated by the change in distance between both position nodes. The HINGE element has two orientation nodes and three deformation modes: torsion around its initial axis and two bending deformations of this axis. The BEAM element has two position nodes and two orientation nodes, and six deformation modes: elongation, torsion around the length axis of the BEAM, and two bending deformations of each endpoint of the BEAM. Contact between elements is obtained by letting them share common nodes. For instance, two BEAM elements rotate around a HINGE defined in between if the orientation nodes of the HINGE are identical to the orientation nodes of connected endpoints of the BEAM elements. The exact formulation of the deformation modes as function of the nodal coordinates is presented in Appendix A.

In addition to these elements new elements can be integrated in the finite element approach. Special purpose elements for the shoulder mechanism are the SURFACE element and the CURVED-TRUSS element. The SURFACE element has one position node and one deformation mode: the distance of the position node to a predefined surface. The CURVED-TRUSS element has three position nodes and one deformation mode: the elongation of the shortest distance between two of its nodes around a predefined surface.

With these five elements the shoulder mechanism can be modelled. In the model, bones are
represented by rigid BEAM elements, joints by spherical joints, composed of three orthogonal HINGE elements, ligaments by flexible TRUSS elements, and muscles by force-generating 'active' TRUSS or CURVED-TRUSS elements, comparable with hydraulic actuators in robots. The scapulothoracic gliding plane is modelled by two SURFACE elements, constraining two points of the medial border of the scapula to the thorax.

Motion of the mechanism is expressed in motion of generalized coordinates, comparable to the Lagrangian approach. These generalized coordinates can be position coordinates or deformation modes. The model can be used for an inverse dynamic as well as a dynamic analysis. In the inverse dynamic analysis the most common input variables of biomechanical models can be used: Either the position of bony landmarks and/or the angle between two segments which is expressed as torsion around the initial axis of a HINGE element. Output variables are muscle forces calculated using an optimization criterion like minimization of the sum of squared muscle forces or minimization of the sum of squared muscle stresses.

For dynamic simulations in the near future, a third-order muscle model based on the work of Winters & Stark (1985) has been developed linking the input motor program of a muscle to the generated force of an active element. These muscle forces in combination with inertia and external load of the system result in the motion of the mechanism.

II.1 Kinematic analysis
The position of the mechanism is fully determined by independent generalized coordinates. Central point in the kinematic analysis is the calculation of the new position of the mechanism as a function of these generalized coordinates. The number of generalized coordinates is equal to the number of degrees of freedom (DOF) of the mechanism. Generalized coordinates can consist of position coordinates and deformation modes. Then, starting from position i the next position (i+1) is approximated by a second-order Taylor expansion:

$$\mathbf{x}_{i+1} - \mathbf{x}_i = D\mathbf{F}^x \cdot (\Delta \mathbf{x}^m, \Delta \mathbf{g}^m) + \frac{1}{2} (D^2 \mathbf{F}^x \cdot (\Delta \mathbf{x}^m, \Delta \mathbf{g}^m)) \cdot (\Delta \mathbf{x}^m, \Delta \mathbf{g}^m),$$

where

$\mathbf{x}$: vector with position and orientation coordinates;

$(\Delta \mathbf{x}^m, \Delta \mathbf{g}^m)$: column vector describing the change of generalized coordinates, position coordinates $(\Delta \mathbf{x}^m = \mathbf{x}_{i+1}^m - \mathbf{x}_i^m)$ and deformation modes $(\Delta \mathbf{g}^m = \mathbf{g}_{i+1} - \mathbf{g}_i)$, respectively;

$D\mathbf{F}^x$: matrix of first derivatives of coordinate vector $\mathbf{x}$ with respect to the column vector of
generalized coordinates: $\delta x/(\delta x_m, \delta z_m)$, in which $(\delta x_m, \delta z_m) = [\delta x^{mT}, \delta z^{mT}]^T$ (superscript $T$ means transposed);

$D^2F^x$: matrix of second derivatives of coordinate vector $\chi$ with respect to the vector of generalized coordinates: $\delta^2 x/((\delta x^m, \delta z^m), (\delta x^m, \delta z^m)^T)$. $D^2F^x$ is a matrix with three dimensions.

At each position $i$ the matrices $DF^x$ and $D^2F^x$, the first and second order geometric transfer function, respectively, can be calculated from the position and deformation of the elements. First the zero-order geometric transfer function is established. This is the non-linear function $D$ of the vector of deformations $\varepsilon$ and the vector of position and orientation coordinates $\chi$ (Appendix A):

$$\varepsilon = D(\chi) \quad (2)$$

Example: A TRUSS element is defined by two position nodes $P(x, y, z)$ and $Q(x, y, z)$ of its endnodes. The only deformation of the TRUSS, i.e. elongation, is calculated as the distance between the nodes and its initial length $l_0$:

$$\varepsilon_1 = \sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2} - l_0.$$

The function in Eq. (2) can be differentiated with respect to the generalized coordinates yielding the first-order geometric transfer function:

$$DF^e = DD \cdot DF^x \quad (3),$$

where $DD$ is a matrix containing derivatives of vector of deformations $\varepsilon$ with respect to the vector of coordinates $\chi$: $\delta \varepsilon/\delta \chi^T$. $DF^c$ is a matrix of first derivatives of the vector of deformations $\varepsilon$ with respect to generalized coordinates $(\delta \varepsilon/(\delta x^m, \delta z^m))$.

Example: Matrix $DD$ can be obtained by differentiating deformation modes of each element with respect to position and orientation coordinates, e.g. the deformation $\varepsilon_1$ of the TRUSS element can be differentiated with respect to the coordinates of its endnodes: $x_p, y_p, z_p, x_q, y_q, z_q$:

$$\begin{align*}
\delta e_1 &= -\frac{(x_q - x_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}}; \\
\delta x_p &= \frac{1}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}}; \\
\delta e_1 &= \frac{-(y_q - y_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}}; \\
\delta y_p &= \frac{1}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}}; \\
\delta e_1 &= \frac{-(z_q - z_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}}; \\
\delta z_p &= \frac{1}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}};
\end{align*}$$
\[
\begin{align*}
\frac{\delta e_1}{\delta x_q} &= \frac{(x_q - x_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}} ; \\
\frac{\delta e_1}{\delta y_q} &= \frac{(y_q - y_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}} ; \\
\frac{\delta e_1}{\delta z_q} &= \frac{(z_q - z_p)}{\sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2}} ; \\
DF^e &= DD \cdot DF^x \\
&= \left[ \begin{array}{cccc}
\delta e_1 & \delta e_1 & \delta e_1 & \delta e_1 \\
\delta x_p & \delta y_p & \delta z_p & \delta e_1 & \delta e_1 & \delta e_1 & \delta e_1 \\
\end{array} \right] \cdot [DF^x].
\end{align*}
\]

Depending on the constraint conditions and the choice of the generalized coordinates the vector of position coordinates \( \bar{x} \) and element deformations \( \bar{e} \) can be partitioned in three parts:

\( \bar{x}^0 \): vector of fixed support coordinates;

\( \bar{x}^c \): vector of dependent nodal coordinates;

\( \bar{x}^m \): vector of global generalized coordinates;

\( \bar{e}^0 \): vector of fixed prescribed deformation mode coordinates;

\( \bar{e}^m \): vector of relative generalized coordinates;

\( \bar{e}^c \): vector of dependent deformation mode coordinates;

and analogously to this partitioning

\[
\begin{bmatrix} 
DF^{x_0} \\
DF^{x_m} \\
DF^{x_c} 
\end{bmatrix} = 
\begin{bmatrix} 
D^{x_0} & D^{x_0} & D^{x_0} \\
D^{x_m} & D^{x_m} & D^{x_m} \\
D^{x_c} & D^{x_c} & D^{x_c} 
\end{bmatrix} \begin{bmatrix} 
DF^{x_0} \\
DF^{x_m} \\
DF^{x_c} 
\end{bmatrix}.
\tag{4}
\]

The only unknown matrices are \( DF^{x_c} \) and \( DF^{x_c} \). The other matrices are

\[
\begin{align*}
DF^{x_0} &= \begin{bmatrix} \delta e_1^o, \delta e_1^o \\
\delta e_1^m, \delta e_1^m \end{bmatrix} = [0, 0] ; \\
DF^{x_m} &= \begin{bmatrix} \delta e_1^m, \delta e_1^m \\
\delta e_1^m, \delta e_1^m \end{bmatrix} = [0, 1] ;
\end{align*}
\tag{5}
\]
\[ DF^{xo} = \begin{bmatrix} \frac{\delta x^o}{\delta x^m} & \frac{\delta x^o}{\delta \varepsilon^m} \end{bmatrix} = [0,0]; \]
\[ DF^{xm} = \begin{bmatrix} \frac{\delta x^m}{\delta x^m} & \frac{\delta x^m}{\delta \varepsilon^m} \end{bmatrix} = [1,0]. \]

The number of dependent nodal coordinates, i.e. the dimension of \( x^c \), is equal to the number of prescribed deformation mode coordinates, i.e. the dimension of \([\varepsilon^o]^T [\varepsilon^m]^T]^T\). Then, the partitioned matrix

\[ \begin{bmatrix} D^c D^o \\ D^c D^m \end{bmatrix}, \]

is a square matrix and if in addition the mechanism is not in a singular position, the unknown first-order geometric transfer function can be calculated by its inverse

\[ DF^{xe} = \left[ D^c D^o \right]^{-1} \left( \begin{bmatrix} DF^{eo} \\ DF^{em} \end{bmatrix} - \begin{bmatrix} D^m D^o \\ D^m D^m \end{bmatrix} . DF^{xm} \right), \]

or with Eq. (5)

\[ DF^{xe} = \left[ D^c D^o \right]^{-1} \begin{bmatrix} D^m D^o & 0 \\ D^m D^m & 1 \end{bmatrix}, \]

and

\[ DF^{ce} = D^c \cdot DF^{xe} + D^m \cdot DF^{xm}. \]

The second-order geometric transfer function can be obtained by differentiating Eq. (3) with respect to the generalized coordinates once again:

\[ D^2 F^c = (D^2 D . DF^e) . DF^x + DD . D^2 F^x, \]

where \( D^2 D \) consists of the second partial derivative of the vector of element deformations with respect to the nodal coordinates \( (\delta \varepsilon^e/(\delta x^o, \delta x^m)) \), and it is therefore a three-dimensional matrix. \( D^2 F^e \) and \( D^2 F^x \) are the second-order geometric transfer function, \( \delta \varepsilon^e/(\delta x^m, \delta \varepsilon^m) . (\delta x^m, \delta \varepsilon^m)^T \) and \( \delta x^e/(\delta x^m, \delta \varepsilon^m) . (\delta x^m, \delta \varepsilon^m)^T \), respectively.

Then, after partitioning
\[
\begin{bmatrix}
D^2F_{\text{co}} \\
D^2F_{\text{cm}} \\
D^2F_{\text{cm}}
\end{bmatrix} = \begin{bmatrix}
(D^2D^0 \cdot D F^0).D F^x \\
(D^2D^m \cdot D F^m).D F^x \\
(D^2D^c \cdot D F^c).D F^x
\end{bmatrix} + \begin{bmatrix}
D^0D^0 & D^0D^0 & D^mD^0 \\
D^0D^m & D^0D^m & D^mD^m \\
D^0D^c & D^0D^c & D^mD^c
\end{bmatrix} \begin{bmatrix}
D^2F_{\text{co}} \\
D^2F_{\text{cm}} \\
D^2F_{\text{cm}}
\end{bmatrix},
\]

(9)

where
\[
D^2F_{\text{co}} = D^2F_{\text{cm}} = 0;
\]
\[
D^2F_{\text{cm}} = D^2F_{\text{cm}} = 0.
\]

Hence, we obtain
\[
D^2F_{\text{cm}} = - \begin{bmatrix}
D^0D^0 \\
D^0D^m \\
D^0D^c
\end{bmatrix}^{-1} \begin{bmatrix}
(D^2D^0 \cdot D F^0).D F^x \\
(D^2D^m \cdot D F^m).D F^x \\
(D^2D^c \cdot D F^c).D F^x
\end{bmatrix};
\]

(10)

\[
D^2F_{\text{cm}} = (D^2D^c \cdot D F^c).D F^x + D^0D^c.D^2F_{\text{cm}}.
\]

(11)

After the first approximation of the new position \(i+1\) of the mechanism is obtained, an iteration process based on the Newton-Raphson method is applied to guarantee that ultimately (Jonker, 1988):
\[
\varepsilon_{i+1} = D(\xi_{i+1}).
\]

(12)

Finally the velocity and acceleration of the nodal coordinates can be calculated from the velocity and acceleration of the generalized coordinates:
\[
\dot{\xi} = D F^x.(\dot{\xi}^m, \dot{\xi}^m);
\]
\[
\ddot{\xi} = D F^e.(\dot{\xi}^m, \dot{\xi}^m);
\]

(13)

\[
\ddot{\xi} = (D^2F^x.(\dot{\xi}^m, \dot{\xi}^m)).(\dot{\xi}^m, \dot{\xi}^m) + D F^x.(\ddot{\xi}^m, \ddot{\xi}^m);
\]
\[
\dddot{\xi} = (D^2F^e.(\dot{\xi}^m, \dot{\xi}^m)).(\dot{\xi}^m, \dot{\xi}^m) + D F^e.(\dddot{\xi}^m, \dddot{\xi}^m).
\]

(14)

II.2 Dynamic analysis

In the dynamic analysis motion equations of the mechanism are derived by using Lagrange’s form of d’Alembert’s principle. In this approach, a virtual power approach, the computational
advantages of the Newton-Euler and Lagrange's equations are combined, i.e. the advantage of eliminating non-working constraint forces without the need of elaborate differentiation of the energy equations. The final number of equations is equal to the number of DOF of the mechanism. These differential equations can be integrated using an appropriate integration algorithm.
To obtain the motion equations the principle of virtual power for the vectors \( f \) of external forces and \( f_{in} \) for inertia forces, and for vector \( g \) of stress associated with the deformation of the elements can be written in the form:

\[
<\dot{\mathbf{x}}, \mathbf{a}> - <\dot{\mathbf{x}}, (f - f_{in})> \rightarrow \dot{\mathbf{x}}^T \cdot \mathbf{a} = \dot{\mathbf{x}}^T (f - f_{in}) ,
\]  

where the notation \(<\cdot, \cdot>\) is the scalar product of two vectors and where

\[
\dot{x} = DF_{xT}(\dot{x}^m, \dot{\mathbf{x}}^m);
\]

\[
\dot{\mathbf{x}} = DF_{xT}(\dot{x}^m, \dot{\mathbf{x}}^m).
\]

Since Eq. (15) is true for every value of \((\dot{x}^m, \dot{\mathbf{x}}^m)\) it can be reduced to

\[
DF_{xT} \cdot \mathbf{a} = DF_{xT} (f - f_{in}) .
\]  

(16)

Inertia in the system is included by subtracting inertia forces \((f_{in})\) using d'Alembert's principle. Since in the shoulder model only rigid BEAM elements are used and the attribution of muscle mass to inertia forces is neglected, inertia in the system can be included using a lumped mass representation at the position and orientation nodes, representing translational and rotational inertia, respectively. Then, with Eq. (14),

\[
f_{in} = M \ddot{x} - M \{(D^2 F_x(\dot{x}^m, \dot{\mathbf{x}}^m)) \cdot (\dot{x}^m, \dot{\mathbf{x}}^m) + DF_{xT}(\dot{x}^m, \dot{\mathbf{x}}^m)\} ,
\]  

(17)

where

\(M\): system mass matrix describing the lumped rotational and translational inertia at the orientation and position nodes, respectively.

Substitution of Eq. (17) in Eq. (16) results in the motion equations:

\[
DF_{xT} M DF_{x}.(\dot{x}^m, \dot{\mathbf{x}}^m) - DF_{xT} \mathbf{a} + DF_{xT} f - DF_{xT} M \{(D^2 F_x(\dot{x}^m, \dot{\mathbf{x}}^m)) \cdot (\dot{x}^m, \dot{\mathbf{x}}^m)\} .
\]  

(18)
Eq. (18) describes a number of differential equations equal to the number of generalized coordinates as depicted by the vector $(Δx^m, Δq^m)$. The motion equations can be solved numerically by using one of the standard numerical integration routines. In this finite element method a predictor/corrector method with variable order and stepsize is used (Shampine & Gordon, 1975). However, in the inverse dynamic analysis as used in the shoulder mechanism model motion equations are reduced to algebraic equations.

After solving the motion equations a kinetostatic analysis is performed in order to calculate the internal stress distribution and the unknown reaction forces of the mechanism. After substituting of Eq. (3) in Eq. (16) the next equation is obtained:

$$DD^T \cdot g = f - f_m.$$  \hspace{1cm} (19)

Analogously to Eq. (4) this can be split into

$$\begin{bmatrix} D^oD^{oT} & D^oD^mT & D^oD^cT \\ D^mD^{oT} & D^mD^mT & D^mD^cT \\ D^mD^mT & D^mD^mT & D^mD^cT \end{bmatrix} \begin{bmatrix} g^o \\ g^m \\ g^c \end{bmatrix} = \begin{bmatrix} f^o \\ f^m - f_m^m \\ f^c - f_m^c \end{bmatrix},$$ \hspace{1cm} (20)

where

$g^0$: vector of stresses attributed to fixed deformations;

$g^m$: vector of driving stresses attributed to generalized deformations;

$g^c$: vector of stresses calculable from dependent deformation mode coordinates;

$f^0$: vector of reaction forces at fixed coordinates;

$f^m$: vector of driving forces at generalized coordinates;

$f^c$: vector of prescribed (external) forces.

The stress components $g^0$ and $g^m$ can be calculated

$$\begin{bmatrix} g^o \\ g^m \end{bmatrix} = [D^oD^{oT} D^oD^mT]^{-1} \cdot (f^o - f_m^c - D^oD^cT \cdot g^c),$$ \hspace{1cm} (21)

where vector $g^c$ is calculated from constitutive equations relating the deformations of the element to stress:
\[ q^c = C \cdot \varepsilon^c + K \cdot \dot{\varepsilon}^c, \]  

(22)

in which \( C \) is the system stiffness matrix and \( K \) is the system damping matrix. Any other user supplied (non-linear) constitutive equation can be used in this equation. Finally the vector of reaction forces \( f^0 \) and the vector of driving forces \( f^m \) can be calculated from Eq. (22).

II.3 Inverse dynamic analysis

Input variables of the model are the prescribed motion of the mechanism, expressed in position, velocity and acceleration of generalized coordinates \((\Delta x^m, \Delta \dot{x}^m)\). In the kinematic analysis length, velocity and acceleration of muscles can be calculated. In the inverse dynamic analysis, force vector \( f^m \) and stress vector \( q^m \) are calculated as driving forces, attributed to the generalized coordinates. In fact, driving forces for this motion are generated by muscles. In the inverse dynamic analysis, driving forces \( f^m \) and \( q^m \) are distributed over the stress vector associated with the stress in the active TRUSS and CURVED-TRUSS elements representing the muscles. Subsequently, in the kinetostatic analysis this muscle stress will result in forces at the endnodes in the direction of the element. Thus, the course of the muscle element can in fact be viewed as the muscle line of action, exerting force at its attachments on the bone. One unit of stress in the muscle element results in one unit of force at both attachments.

In the kinematic analysis deformation modes of the active elements are part of the deformation vector \( \varepsilon^c \). The associated stress vector \( q^c \) is split into two parts: \( q^{c1} \) for the passive dependent stresses and \( q^{c2} \) for the active stresses, to be calculated in the inverse dynamic analysis. However, since in general there are more active elements than DOF, the solution for \( q^{c2} \) is indeterminate. Therefore, this vector \( q^{c2} \) is calculated in an optimization procedure with an adequate optimization criterium, whereas equality constraints from the motion equations have to be considered. From the motion equations (Eq. 20) the relation between active stresses \( q^{c2} \) on the one hand and driving forces \( f^m \) and driving stresses \( q^m \) of the generalized coordinates on the other hand can be derived:

\[ DF^{\infty T} \cdot q^{c2} = -DF^{\infty m} \cdot q^m + DF^{x m T} \cdot (f^m - \bar{f}^m_h). \]  

(23)

Substituting Eq. (5) in Eq. (23) results in:

\[ DF^{\infty T} \cdot q^{c2} = \begin{bmatrix} (f^m - \bar{f}^m_h) \\ -\bar{q}^m \end{bmatrix}. \]  

(24)
Eq. (24) describes the equality constraints for the optimization procedure due to the motion equations.

Example: In this equation the matrix $D\dot{\epsilon}c_{2T}$ is the derivative of the elongation of the muscle elements with respect to the degrees of freedom $(\delta \epsilon^2/(\delta x^m, \delta \epsilon^m))$. If the rotations $\phi$ of the joint are the DOF, then $\gamma^m$ corresponds to the vector of net joint moments and $\delta \epsilon^2/(\delta x^m, \delta \epsilon^m)$ is analogous to the definition of An et al. (1984) for moment arm $a$ of a muscle:

$$a = \delta l/\delta \phi.$$

Then Eq. (24) can be compared with

$$\sum_i a_i \cdot F_i = M,$$

where

- $a_i$ : moment arm of muscle $i$ ;
- $F_i$ : force attributed to muscle $i$ ;
- $M$ : net joint moment.

Furthermore, additional constraints to the optimization procedure can be imposed, e.g. the constraint that at the SURFACE elements of the scapulothoracic gliding plane only compression forces can occur and that only traction forces in the ligaments exists. Special constraints to the joint reaction forces in the glenohumeral joint will be explained later.

In the kinetostatic equations these constraints exist in the form of limitations on particular items of the $\sigma^0$ vector. If $\sigma^0$ is zero no stresses occur, if $\sigma^0$ is smaller than zero compression forces occur and if $\sigma^0$ is greater than zero traction forces occur. The relation between these items of $\sigma^0$ and the variables $\sigma^2$ in the optimization procedure can be established according to Eq. (23):

$$[D\sigma^0 \sigma^0T]^{-1}D\sigma^2 \sigma^2 - \frac{\sigma^0}{\sigma^m} - \frac{\sigma^0}{\sigma^m} \sigma^{cl} = (f^c - f^c_{in} - D\sigma^0 \sigma^0T \sigma^2).$$

Finally, each element of the vector $\sigma^2$ is constrained to be greater than zero (only traction forces) and smaller than the maximal force which can be exerted by the muscle element depending on the physiological cross-sectional area of the muscle.

II.4 SURFACE element

The scapulothoracic gliding plane imposes constraints on the scapular motions which turns the shoulder girdle into a closed-chain mechanism. Reaction forces exerted by the thorax at the scapula are important for the force and moment balance of the sternoclavicular and acromioclavicular joint. In the shoulder model the scapulothoracic gliding plane is represented by two SURFACE elements,
one near the Trigonum Spinae (TS) and one near the Angulus Inferior (AI). In this way it is assured that the medial border of the scapula is connected to the thorax. The SURFACE element has been described by Pronk (1989) for kinematic purposes. Of late, this element was adapted such that it could be used for dynamic analysis as well. The SURFACE element has one position node and in addition a parameter description of the surface area to which the node is attached. It has one deformation mode: The distance between the node and the surface.

Pronk & Padt (1986), Van der Helm et al. (1991) and Pronk (1991) showed that the thorax can be modelled by an ellipsoid. The distance from a point to the ellipsoid is calculated iteratively along the normal vector to the ellipsoid through this point (Appendix B). The first and second derivative of this deformation is calculated from the normal vector. To ensure that the point is attached to the surface, the deformation should be zero. The stress in this SURFACE element, i.e. the force exerted by the node at the surface, will be directed along the normal vector. During the optimization procedure this force will be constrained such that only compression can occur.

### II.5 CURVED-TRUSS element

A number of muscles in the shoulder region wraps around underlying morphological structures, i.e. bony contours. There are basically two methods to represent these muscles (Van der Helm, 1991). The first method is to use the centroid line for estimation of the muscle line of action. The main disadvantage of this method is that it is only applicable to one position of the model. The method used in the current model is the bony contour method: The muscle line of action is the shortest distance between origin and insertion around the bony contour which is in between. The effective moment arm of the muscle can be calculated from the straight line between one attachment and the tangent point at the bony contour. For any position of the mechanism this line can be calculated and is represented by a special element, the CURVED-TRUSS element. The CURVED-TRUSS element has three nodes: the origin, insertion and tangent point constituting the effective muscle line of action. After each step the tangent points are calculated again. The only deformation is elongation of the element, calculated from the length of the element around the bony contour. The first and second derivative of the element are calculated from the straight line between one attachment and its tangent point. Three representations of bony contours are used in the shoulder model: A sphere (caput humeri; the combined tuberculum majus and minus), a cylinder (collum humeri) and an ellipsoid (thorax). Appendix C describes the calculation of the deformation and derivatives of the CURVED-TRUSS element.
III MODEL OF THE SHOULDER MECHANISM

In Fig. 1 the model of the shoulder mechanism is depicted in a block diagram. Input variables to the model are position, velocity and acceleration of generalized coordinates and external forces. Output variables of the model consist of muscle forces calculated by using an optimization criterium. The model itself consists of motion equations describing the mechanical behavior of the shoulder mechanism, derived by using the finite element method.

Parameters for the model have been derived in an extensive cadaver study consisting of both shoulders of seven cadavers. There are three types of parameters: Inertia parameters describing the translational and rotational inertia of the segments, muscle contraction parameters describing the maximal force attributed to a muscle element depending on the Physiological Cross-Sectional Area (PCSA) of the muscle and geometry parameters describing the position of joint rotation centers, muscle and ligament attachments, shape and position of bony contours and the position of bony landmarks to link the recorded motion of subjects to motions of the model. In this study parameters derived at the right shoulder of a more or less median cadaver are used. The complete cadaver experiment, including all geometric data used in the present model, has been described elsewhere (Veeger et al., 1991a; Van der Helm et al., 1991; Van der Helm & Veenbaas, 1991).

The sternoclavicular (SC-)joint between sternum and clavica is represented as a spherical joint by three orthogonal HINGE elements. A BEAM representing the clavica connects this joint with the acromioclavicular (AC-)joint between clavica and scapula, also modelled as a spherical joint. The scapula is represented by two rigidly connected BEAM elements: One from the AC-joint to trigonum spinae (TS) at the medial border of the scapula and the other from TS to angulus inferior

**Figure 1:** Block diagram of the model of the shoulder mechanism. Input variables are the position (and velocity and acceleration) of the bones ($x^m, \dot{x}^m$ and derivatives) and the external load ($F^e$). Output variables are muscle forces ($\sigma^{c2}$). Model parameters describe the shoulder morphology.
(AI). TS and AI are connected with SURFACE elements to the thorax. The glenohumeral (GH-)joint is connected to the AC-joint by another BEAM rigidly connected with the scapula. The humerus, and the rest of the arm, is represented by a BEAM from the GH-joint to the midpoint between the medial and lateral epicondyle. The mechanism consisting of clavicula, scapula, humerus and scapulothoracic gliding plane, with three spherical joints, has seven DOF: Four at the shoulder girdle and three for the humerus. Ligaments are modelled as flexible TRUSS elements. Attachment points of these elements are rigidly connected with BEAM elements to the respective bones. Three extracapsular ligaments are accounted for in the model: The costoclavicular ligament crossing the SC-joint and the conoid and trapezoid ligaments crossing the AC-joint. The mechanism described so far is shown in Fig. 2., with a few muscle elements added.

Muscles are modelled as active elements: TRUSS or CURVED-TRUSS elements. Mostly more than one element is necessary for adequately representing the mechanical effect of muscles with large attachment sites. Each muscle is represented by one to six elements where each element can

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**Figure 2**: The shoulder mechanism described by finite elements. Only one ligament and a few muscle elements are shown.

- **BEAM** (bone)
- **HINGE** (joint)
- **TRUSS** (ligament)
- **SURFACE** (scapulothoracic gliding plane)
- **TRUSS** or **CURVED-TRUSS** (muscle).
be considered as a single independent muscle line of action. The actual number of elements depends on the shape of origin and insertion and the muscle architecture, as recorded in the cadaver experiment. A general theory for this partitioning of the complete muscle into muscle elements and the coordinates of the attachments has been presented in Van der Helm & Veenbaas (1991). An equal share of the PCSA recorded for the whole muscle is attributed to each element. In order to achieve a realistic force balance in the shoulder mechanism two other constraints are necessary. Motion constraints of the scapulothoracic gliding plane are modelled by two SURFACE elements. In reality, this constraint can only occur if the scapula is pressed against the thorax. Therefore, the stress in SURFACE elements is constrained to zero or negative stress (compression).

In the glenohumeral joint the resultant force vector should point to the glenoid cavity otherwise it can not be neutralized by the joint reaction force and the joint will dislocate (Van der Helm et al., 1989), see Fig. 3. Restraining the force vector to point just anywhere to the glenoid cavity leads to a non-linear constraint which is difficult to take into account. Therefore, if the force vector is not already pointing to the glenoid cavity, the direction of the force vector is limited to pointing to the nearest point at the glenoid cavity (Appendix D). This is achieved by implementation of an extra BEAM between the intersection of the glenoid cavity and the desired reaction force vector, and the rotation center of the joint. This BEAM should only be subject to compression stress, not to bending stress.

![Diagram](image)

**Figure 3**: Resultant force vector \( F_h \) should point to the glenoid, otherwise it can not be neutralized by the joint reaction force \( F_{rh} \) and the joint will dislocate (GL: Glenoid cavity; HH: Humeral head).
Input variables to the model are positions and motions of the bones. Three-dimensional motion recording of the shoulder girdle is difficult because the scapula and clavicle move underneath the skin and are not accessible for optical methods but roentgenography. Wallace and coworkers (Wallace, 1982; Wallace & Johnson, 1982) made 3D roentgenographic recordings but unfortunately did not present a 3D description of the motions. For an accurate recording the implementation of markers is probably necessary as is done by Peterson et al. (1985). However, results of these experiments are not published in detail. Pronk (1987; 1991) and Van der Helm & Pronk (1991a) used a palpation technique to record 3D positions of bony landmarks of the scapula and humerus. The position of bony landmarks is used to reconstruct the position and orientation of the bones.

A disadvantage of this method is that only static positions of the shoulder mechanism can be recorded. The data published in Van der Helm & Pronk (1991a) are used for simulations with the model. It should be noted that the limitation to static simulations is caused by available data on scapular positions. The finite element model of the shoulder mechanism allows complete inverse dynamic simulations including velocity and acceleration of the segments.

The choice of adequate generalized coordinates for input is very important. The number of generalized coordinates is equal to the number of DOF. The shoulder model has seven degrees of freedom: four at the shoulder girdle and three at the humerus. In most inverse dynamic simulation studies input variables are the orientation angles of the segments. However, due to the complex motion constraints imposed by the scapulothoracic gliding plane and the clavicle, prescription of the orientation of the scapula results in irrational positions on the thorax. Therefore, position coordinates of bony landmarks are chosen for generalized coordinates, i.e. the y- and z-coordinate of the bony landmark AC (most dorsal point on the AC-joint) and the x-coordinate of the bony landmark trigonium spinae (TS). Rotations of the humerus are decomposed as Euler angles in the rotation order pole angle around the global Y-axis, elevation around the local z-axis and axial rotation around the local y-axis (Van der Helm & Pronk, 1991a). All positions and rotations are described with reference to a global coordinate system with the origin at the incisura jugularis (II) and the axes along the anatomical axes (X-axis: medial-lateral; Y-axis: caudal-cranial; Z-axis: ventral-dorsal; right shoulder). The position of the bony landmarks was measured in the cadaver experiments mentioned and thus the recorded motion can be related to the geometry of the cadaver. Position nodes of the bony landmarks are rigidly connected by BEAM elements to the respective bones. Hence, a position change of the bony landmarks will cause a position change of the bones and of all muscle and ligament attachments rigidly connected to these bones.

Axial rotation of the clavicle could not be measured in the palpation experiment. In order to
compensate for this missing input variable of the clavicular motion, the number of DOF has been reduced by one through the assumption that the conoid ligament is rigid. Simulations have revealed that only the axial rotation of the clavicle is affected, and that the resulting axial rotation is close to the optimal axial rotation, obtained by minimizing rotations in the AC-joint (Pronk, 1991; Van der Helm & Pronk, 1991a).

Coordinates of the bony landmarks were recorded at 10 subjects during unloaded abduction. The mean orientations of the clavicula and scapula were derived. Using these mean orientations and the bony dimensions of the cadaver while omitting the motion constraints of the thorax, the desired position coordinates of the bony landmarks were obtained. However, using three of these position coordinates as input variables, the resulting scapular position was slightly changed from the mean orientation, due to the imposed motion constraints of the thorax in the model.

Output of the model consists of forces generated by the active elements, subject to an optimization criterion. In addition, joint reaction forces and reaction forces at the scapulothoracic gliding plane are calculated. Validation of the model is severely hampered because none of these forces can be measured directly. Instead, surface EMG recorded at twelve locations is used as an indirect measure of muscle activity.

IV Results
IV.1 Kinematics

Figs 4A and 4B show the excursion of the scapula over the thorax during humeral abduction, projected on the frontal and the sagittal plane. At 30 degrees abduction, the scapula is oriented towards the plane of humeral elevation by retraction and slightly medial rotation. When the humerus is further abducted, the main rotation of the scapula is lateral rotation in the scapular plane.

Motions of the scapula are important for the excursions of the muscle lines of actions, i.e. the position of the muscle elements. Fig. 5A shows the position of the six muscle elements of the scapular part of m. trapezius at 0 degrees humeral abduction. The location of these elements is derived from measurements of origin and insertion including the muscle architecture. Fig. 5B shows the same six elements at 180 degrees abduction. The elements are decreased in length and, more clearly, the orientation of the elements with respect to each other and to the articulations have been changed. This implies that the moment arms with respect to the rotation centers of the joints have been changed as well.
IV.2 Inverse dynamics

As stated in Section III an optimization procedure is applied to derive a unique vector of muscle stresses $\mathbf{g}^2$ from the motion equations. Stress in a muscle element results in traction forces at its endnodes, exactly as is the case during muscle contraction. One unit of muscle stress results then in one unit of force. The optimization criterion is composed of the stresses $\mathbf{g}^2$ in the active TRUSS and CURVED-TRUSS elements, the results are hereafter presented as muscle forces at the attachments. Four different optimization criteria have been used:

a. Minimization of the sum of quadratic muscle forces: $\min \Sigma F^2$.

Since in the finite element model one unit of stress in the muscle element corresponds to one unit of force exerted at the attachment, this criterion has been implemented as: $\min \mathbf{g}^2^T \mathbf{g}^2$.

b. Minimization of the sum of quadratic muscle stresses: $\min \Sigma (F/\text{PCSA})^2$, where PCSA is the physiological cross-sectional area attributed to each muscle element. Analogously to criterion a this criterion has been implemented as $\min (\mathbf{g}^2/\text{PCSA})^T (\mathbf{g}^2/\text{PCSA})$.

c. Minimization of the sum of quadratic muscle forces, normalized to the maximal muscle force $F_{i\text{max}}$ which is a function of PCSA and length of the muscle: $\min \Sigma (F_i/F_{i\text{max}})^2$.

$d. Minimization$ of the maximal muscle stress in the entire mechanism:

![Diagram A](image)

![Diagram B](image)

**Figure 4:** Position of the scapula (bony landmarks AC, TS and AI) and clavícula (bony landmarks IJ and AC) during humeral abduction.

A: Projection frontal plane.

B: Projection sagittal plane.
\[ \min \max \left( \frac{F_1}{\text{PCSA}_1}, \frac{F_2}{\text{PCSA}_2}, \ldots, \frac{F_i}{\text{PCSA}_i}, \ldots, \frac{F_N}{\text{PCSA}_N} \right) \], \text{ which has been implemented as } \min \max \left( \frac{\sigma_1}{\text{PCSA}_1}, \frac{\sigma_2}{\text{PCSA}_2}, \ldots, \frac{\sigma_i}{\text{PCSA}_i}, \ldots, \frac{\sigma_N}{\text{PCSA}_N} \right). \\

As an example, Figs 6A to 6O show results for all muscles using criterion b. For comparison, all results are presented with the same vertical axis. However, the number of elements per muscle should be noted in order to get an impression of the total force exerted by the whole muscle (Table 2).

Special attention is paid to the force and moment balance of the glenohumeral joint. The resulting force vector is constrained to point from the rotation center of the glenohumeral joint to the glenoid cavity. If the constraint is working, in this simulation at 0 and 30 degrees of humeral abduction, especially the antagonistic muscles of the rotator cuff become active. Force output of all other muscles is hardly affected, but the location and the small moment arms of the rotator cuff favours these muscles to deliver the force to keep the humeral head in its socket.

For a few positions of the shoulder mechanism the weight of the arm and counterbalancing muscle forces would result in a dilatation of the trigonum spinae from the scapulothoracic gliding plane.

The combined action of the m. rhomboideus and the m. serratus anterior preserves the scapula from loosing contact from the thorax which would be a very unstable situation. This constraint becomes more important with higher loads and during other motions like anteflexion.

Verification of such a complex model of the shoulder mechanism is a cumbersome item.

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**Figure 5:** Position of six TRUSS elements describing the mechanical effect of m. trapezius, scapular part (dorsal view).

A: 0 degrees humeral abduction (rest position).
B: 180 degrees humeral abduction.
Figure 6: Muscle force during humeral abduction from zero to 180 degrees, calculated by minimizing the sum of quadratic muscle stresses. Each line represents the force in a muscle element.
Table 2: Physiological cross-sectional area (PCSA) in cm$^2$, number of 'active' TRUSS or CURVED-TRUSS elements and PCSA per element (cm$^2$) for 20 muscles and muscle parts of the shoulder mechanism.

<table>
<thead>
<tr>
<th>muscle</th>
<th>PCSA</th>
<th># elements</th>
<th>PCSA/element</th>
</tr>
</thead>
<tbody>
<tr>
<td>m. trapezius, scapular part</td>
<td>14.34</td>
<td>6</td>
<td>2.39</td>
</tr>
<tr>
<td>m. trapezius, clavicular part</td>
<td>3.12</td>
<td>6</td>
<td>0.52</td>
</tr>
<tr>
<td>m. levator scapulae</td>
<td>3.44</td>
<td>3</td>
<td>1.15</td>
</tr>
<tr>
<td>m. pectoralis minor</td>
<td>3.43</td>
<td>4</td>
<td>0.86</td>
</tr>
<tr>
<td>m. rhomboideus</td>
<td>7.57</td>
<td>3</td>
<td>2.52</td>
</tr>
<tr>
<td>m. serratus anterior</td>
<td>11.43</td>
<td>6</td>
<td>1.91</td>
</tr>
<tr>
<td>m. deltoideus, scapular part</td>
<td>16.58</td>
<td>6</td>
<td>2.76</td>
</tr>
<tr>
<td>m. deltoideus, clavicular part</td>
<td>8.07</td>
<td>6</td>
<td>1.35</td>
</tr>
<tr>
<td>m. coracobrachialis</td>
<td>3.20</td>
<td>6</td>
<td>0.53</td>
</tr>
<tr>
<td>m. infraspinatus</td>
<td>8.17</td>
<td>6</td>
<td>1.36</td>
</tr>
<tr>
<td>m. teres minor</td>
<td>3.10</td>
<td>6</td>
<td>0.52</td>
</tr>
<tr>
<td>m. teres major</td>
<td>12.56</td>
<td>6</td>
<td>2.09</td>
</tr>
<tr>
<td>m. supraspinatus</td>
<td>4.69</td>
<td>6</td>
<td>0.78</td>
</tr>
<tr>
<td>m. subscapularis</td>
<td>14.99</td>
<td>6</td>
<td>2.50</td>
</tr>
<tr>
<td>m. biceps, caput longum</td>
<td>1.72</td>
<td>1</td>
<td>1.72</td>
</tr>
<tr>
<td>m. biceps, caput breve</td>
<td>1.78</td>
<td>1</td>
<td>1.78</td>
</tr>
<tr>
<td>m. triceps, caput longum</td>
<td>6.24</td>
<td>2</td>
<td>3.12</td>
</tr>
<tr>
<td>m. latissimus dorsi</td>
<td>8.52</td>
<td>5</td>
<td>1.70</td>
</tr>
<tr>
<td>m. pectoralis major, thor. part</td>
<td>8.68</td>
<td>5</td>
<td>1.74</td>
</tr>
<tr>
<td>m. pectoralis major, clav. part</td>
<td>3.55</td>
<td>5</td>
<td>0.71</td>
</tr>
</tbody>
</table>

Obviously, direct measurement of muscle forces is impossible. Generally, EMG is used for assessing muscle activity. In twelve subjects, eight of them also acted in the palpation experiment, surface EMG has been measured at twelve well-defined spots (Table 3). For each spot EMG amplitude has been compared with the force output of the nearby muscle element as calculated using the four optimization criteria. First inspection of EMG patterns revealed for almost all muscles a monotonously increasing amplitude during abduction, in contrast to the muscle forces which peaked at 90 degrees abduction, as would be expected. In literature, similar EMG patterns are found during humeral abduction (e.g. Inman et al., 1944; Saha, 1961). Kronberg (1989)
presented parabolic curves but still EMG amplitude is much higher at 180 degrees abduction than at 0 degrees. Clearly, for EMG amplitude not only exerted muscle force is a determining factor, but also the muscle length at which EMG is recorded (Heckathorne & Childress, 1981).

Two ways exist to include this length dependency in verification of the model. Firstly, it can be assumed that maximal EMG does not depend on muscle length and that it is related with the maximal force output for each muscle length. The ratio between the recorded EMG and overall maximal EMG can be compared with the ratio between the calculated muscle force and the maximal muscle force at the specified muscle length. Secondly, at each abduction angle in the EMG experiment maximal force output and maximal EMG were recorded. This value will be referred to as local $EMG_{\text{max}}$. But now the problem arises that at maximal force output not necessarily all active muscles are maximally active. During humeral abduction many muscles are solely active in stabilizing the scapula and humerus and not in contributing to the maximal force output. However, with the shoulder model the maximal force output can be simulated as well. The ratio between recorded EMG and local $EMG_{\text{max}}$ can be compared with the ratio between calculated muscle force and the muscle force calculated at maximal force output.

Both methods have been applied in the shoulder model, but neither of them gave the desired results. Despite normalization, EMG amplitude curves were still monotonously increasing, whereas the normalized muscle forces had a peak around 90 degrees abduction. Maximal EMG depends on muscle length (Heckathorne & Childress, 1981), the optimum muscle length is probably not at the initial length and the width of the force-length relation is unknown. It is concluded that no EMG-force relation can be assessed without accounting for change of muscle length which hampers the use of surface EMG (Mouton et al., 1991). Apparently, only on-off patterns of recorded EMG can be used for verification of the model. However, humeral abduction is not a cyclic motion like walking or cycling, and most muscles are active throughout the motion. Fig. 7 shows the results of the four different optimization criteria for twelve muscles. Fig. 8 shows the EMG patterns of these muscles as a percentage of maximal EMG. It is concluded that results of the model simulations reasonably correspond to recorded EMG patterns, based on a on-off comparison, but that the verification is merely qualitative and no distinction between the four criteria could be made.

Criterion d, minimization of the maximal muscle stress, appeared to be a numerically unstable criterion. At 0 and 90 degrees humeral abduction, no solution was found for the indeterminacy problem. At 90 degrees humeral abduction the muscle stress needed around the glenohumeral joint exceeded the muscle stress needed for the thoracoscapular muscles. No load sharing between these
Table 3: Electrode positions for surface EMG recordings.

<table>
<thead>
<tr>
<th>muscle</th>
<th>electrode position</th>
</tr>
</thead>
<tbody>
<tr>
<td>m. trapezius, p. descendens</td>
<td>in a vertical line above Trigonum Spinae (TS), 2 cm below the edge of the muscle.</td>
</tr>
<tr>
<td>m. trapezius, p. transversalis</td>
<td>1 cm medial and 4 cm cranial to TS.</td>
</tr>
<tr>
<td>m. trapezius, p. ascendens</td>
<td>2/3 on the line between TS and the eighth thoracic vertebra (T8), 4 cm from the edge of the muscle.</td>
</tr>
<tr>
<td>m. deltoideus, p. anterior</td>
<td>2 cm below the anterior rim of the acromion.</td>
</tr>
<tr>
<td>m. deltoideus, p. medialis</td>
<td>2 cm below the lateral rim of the acromion.</td>
</tr>
<tr>
<td>m. deltoideus, p. posterior</td>
<td>2 cm below angulus acromialis.</td>
</tr>
<tr>
<td>m. infraspinatus</td>
<td>in the midst between TS and angulus Inferior (AI), 2 cm of the medial border of the scapula.</td>
</tr>
<tr>
<td>m. serratus anterior</td>
<td>roughly the sixth head at the level of the nipple, just lateral to the m. pectoralis major.</td>
</tr>
<tr>
<td>m. latissimus dorsi</td>
<td>6 cm below AI.</td>
</tr>
<tr>
<td>m. pectoralis major, p. clav.</td>
<td>in the midst between sternoclavicular joint and processus coracoideus, 2 cm below the clavica.</td>
</tr>
<tr>
<td>m. pectoralis major, p. thor.</td>
<td>6 cm above the nipple.</td>
</tr>
<tr>
<td>m. biceps, caput breve &amp; longum</td>
<td>in the midst of the muscle</td>
</tr>
</tbody>
</table>

two muscle groups could be provided by e.g. a muscle with a suitable location which crosses all three joints. Hence, the load sharing problem between the thoracoscapular muscles is still unresolved, and no solution for the entire system has been found. In the initial position (0 degrees humeral abduction) a similar problem occurred, but in this position the stress in the thoracoscapular muscles exceeds the stress in the glenohumeral muscles.

V DISCUSSION

Even if a model is considered to be a simplification of the real system, a model of such a complex system as the shoulder mechanism still remains a very complex model. This complexity is probably the reason that hardly any models exist. Mollier (1899), Shiino (1913) and Hvorslev (1927) used human specimen to build physical models of the shoulder mechanism (Fig. 9). They replaced the muscles by cords and connected them to a keyboard. Then, by pressing the keys a motion could be established and in reverse, when moving the bones the deviation of the keys could be observed
Figure 7: Muscle forces (as percentage of maximal muscle force) calculated with four different optimization criteria.

- : $\min \Sigma F^2$.
- : $\min \Sigma (F/PCSA)^2$.
- : $\min \Sigma (F_i/F_{i,max})^2$.
- : $\min (\max(F_1/PCSA_1 \ldots F_i/PCSA_i \ldots F_N/PCSA_N))$
Figure 8: Surface EMG (percentage of maximal EMG) of a number of muscles of the shoulder mechanism.
to measure the length changes of the muscles. These models were essentially kinematic.

Things went wrong when two-dimensional biomechanical models started to dominate the research of the shoulder mechanism. These models merely described the motion of the humerus with respect to a non-moving scapula (DeLuca & Forrest, 1973; Poppen & Walker, 1978; Dul, 1987). These models did not add much to the comprehension and understanding of the entire system.

Three-dimensional models are scarce. Karlsson et al. (1989) and Karlsson (1990) used a Newton-Euler approach and 3D roentgenographic motion recording for modelling of the shoulder mechanism. Wood et al. (1989a) described a model based on a Lagrangian approach, but did not elaborate on the necessary equations. Both modelling approaches did not present much details thus far, so it is not clear if important features of the shoulder mechanism are included, e.g. motion of both scapula and humerus, change of moment arms due to this motion, constraints of the scapulothoracic gliding plane and the stability of the glenohumeral joint.

The finite element method as presented in this paper is a very powerful tool for musculoskeletal modelling, allowing the development of all kinds of mechanisms. Elements are easily added or removed. The choice of input variables can be changed as well as the position coordinates of morphological structures. The basic elements combined with the special purpose elements provide modelling of all important morphological structures to be represented.

Building an integral model of the shoulder mechanism is essential for analyzing the function of

---

**Figure 9:** Shoulder 'organ', built by Mollier (1899). Muscles have been replaced by cords. The mechanism can be moved by pressing the keys which result in 'contraction' of the muscles.
morphological structures. It gives a complete picture of the shoulder motion and the muscle action, instead of isolating small parts and neglecting the influence of separate parts on each other. In fact, the finite element model of the shoulder mechanism as presented in this paper very much resembles the physical model as built by Mollier (1899), see Fig. 9. Length changes of the muscles can be calculated from motion of the bones, and in reversed order, when the length of a muscle is prescribed, the resulting motion can be calculated. But in addition the finite element model is (inverse) dynamic, i.e. the force in the muscles can be calculated subject to the motion and external load of the mechanism. A new theory was developed for adequately representing the mechanical effect of muscles with large attachment sites with a limited number of muscle lines of action (Van der Helm & Veenbaas, 1991). In the model each muscle line of action is represented by an active, force-generating element between origin and insertion. These elements can be straight or curved around a bony contour (Van der Helm, 1991).

In the inverse dynamic analysis no stiffness is attributed to ligaments, since the recorded input motion was not accurate enough for assessing the deformation of the ligaments which incorrect deformations could consequently result in extremely large stresses in the ligaments and hence in the whole mechanism. Thus, only the length change of ligaments can be analysed. The conoid ligament was chosen to have infinite stiffness (rigid). Evidently, there is no length change, but forces can be transmitted from scapula to clavicle through the ligament. Included in the constraints of the optimization procedure, only traction forces are allowed in this ligament.

Furthermore, important aspects are included as the constraint imposed by the scapulothoracic gliding plane, rendering the system a closed-chain mechanism, and the necessary stability of the glenohumeral joint. The restriction to compression forces in the gliding plane results in additional forces in mainly the m. rhomboideus and m. serratus anterior, pulling the scapula to the thorax. In fact, when the constraint is working, activity of these muscles is still underestimated because at zero compression force the scapula is not sufficiently stabilized: external and internal disturbances could easily cause the scapula to loose contact which would result in a limited positioning ability for the entire upper extremity.

Due to the lax capsule and small articulating surface of the glenoid cavity, the glenohumeral joint can easily dislocate. Muscle forces are essential for controlling the stability of the joint. The resultant force vector is constrained to pass from the rotation center through the articular surface of the glenoid cavity. If this vector would point outside the glenoid cavity it could not fully be counteracted by the joint reaction force vector, and a dislocating force would result, see Fig. 3 (Van der Helm et al., 1989). This constraint mainly influences the force generated by muscles of
the rotator cuff. Without this constraint these muscles are hardly active, due to optimization criteria which favour muscles with large moment arms. If the constraint is working the rotator cuff becomes active, because pointing the resultant force vector requires antagonistic muscle forces and then muscles with smaller moment arms are favoured. As a matter of fact, in the model activity of the rotator cuff muscles is underestimated because the resultant force vector is allowed to point anywhere within the glenoid cavity. It is doubtful whether a stable position of the glenohumeral joint is achieved when the joint reaction force vector points towards the rim of the glenoid. Adjusting the joint reaction force vector towards the midst of the glenoid cavity requires large forces of the rotator cuff muscles, proportional to the magnitude of the joint reaction force vector. Nevertheless, thus far in all musculoskeletal models it is assumed that calculated joint reaction forces are counteracted by joint structures as ligaments and the shape of articular surfaces. It is not likely that this is always true. In the glenohumeral joint, with its lax capsule and spherical shape of the articular surface, the orientation of the joint reaction force which prevents dislocation, is easily calculated. Sometimes additional muscle force is required to maintain stability. It can be concluded that for all joints in the human body this stability requirement should be considered which will probably shed new light on the role of muscles with small moment arms.

Any model is as good as its validation. Validation generally requires comparison of the predicted output of the model with measured output of the system. One problem of validating musculoskeletal models is that the predicted output, muscle forces, can not be measured directly. Normally only the external moment delivered by a number of muscles can be measured, which only can be decomposed into individual muscle forces using a musculoskeletal model. Therefore, the strict requirements for validation of the model are not fulfilled. Only verification of the model is possible, i.e. determine whether the predicted behavior of the model agrees qualitatively with the recorded output. One method to assess the individual contribution of muscles is to compare predicted muscle force with recorded EMG. In this study surface EMG was recorded at twelve locations in the shoulder mechanism. Formally, verification of the force output of 95 muscle elements with a subset of 12 recorded signals would not be sufficient. However, if these 12 signals agree, the other 83 signals will be assumed to be similarly accurate due to the large interactions in the system.

In this study four different optimizing criteria are used to calculate muscle forces. Comparison of the results with EMG amplitude should offer the opportunity to distinguish between these criteria. However, the comparison is hampered by the fact that EMG is length dependent (Heckathorne & Childress, 1981). Length changes range from 52% to 167% with respect to initial length at 0
degrees humeral abduction, contrasting with published force-length relationships ranging from about 60 to 140% of the optimum length (Woittiez, 1984). To eliminate length dependency one method is to normalize the calculated muscle force to maximal force estimated using the force-length relationship. Another method is to normalize the EMG to the local maximal EMG. In both cases comparison failed because optimum muscle length in the force-length relation is unknown. Despite normalization with length dependent measures, the relation between normalized EMG and muscle force remains obscure. More fundamental EMG studies have to be performed to analyse length-dependency and in addition optimum muscle length in vivo needs to be assessed. Meanwhile, verification of the shoulder model is limited to specific features, more based on on-off comparison than on quantitative comparison with the EMG amplitude. As shown in Figs 7 and 8, similarity between force patterns and EMG is for most muscles reasonable. In our opinion, results of the model calculations are better predictions of the muscle force than the EMG amplitude which is subject to measurement errors and all kind of unknown influences.

Since more muscle elements are present than motion equations, the system is indeterminate. An optimization criterion is necessary which preferably reflects the physiological criterion the central nervous system uses to control muscle forces. Criterion a, minimization of quadratic muscle forces, gives incorrect results, because this criterion does not account for the stress in the muscle, denoted by the PCSA, which leads to an overuse of favourably located muscles. In addition, this criterion is very sensitive to the number of muscle lines of action used to represent the muscle, just because the PCSA is not included in the criterion. Criterion b, minimization of the sum of squared muscle stresses, distributes the necessary muscle forces partly according to the PCSA. However, even using this criterion, some muscle elements which have an excellent moment arm will be favoured to a large extent, e.g. m. deltoideus, scapular part, and m. subscapularis. Criterion c includes the effect of the force-length relationship. A disadvantage is that the muscle optimum length is unknown in vivo, and in addition, considering the large length changes of the muscles (52% to 167%), the force-length relationship needs to be re-evaluated. The last criterion, minimization of the maximal muscle stress in the mechanism, prevents the calculation of extremely high muscle stresses. A disadvantage of this criterion is that it is numerically unstable, since the DOF of the scapula and humerus are sometimes not sufficiently related to each other.

Criteria b, c and d do not show much differences. With help of EMG it could not be distinguished which of these optimization criteria provides the best muscle force predictions. Considering the computational efficiency criterion b, minimization of the sum of squared muscle stresses, has our preference.
Conclusions

1) The finite element method as used in the computer program SPACAR is very much suited for complex musculoskeletal models as the shoulder mechanism.

2) A detailed model of the shoulder mechanism has been developed which can provide excellent insight in the function of morphological structures.

3) Constraints imposed by the scapulothoracic gliding plane are important for the motions and force balance of the shoulder girdle.

4) EMG amplitude is a poor measure to validate musculoskeletal models. Only on/off patterns can be compared.

5) Rotator cuff muscles are active in controlling the stability of the glenohumeral joint by pointing the resultant force vector to the articular surface of the glenoid.

Acknowledgements - The help of Hein Daanen and Monique van der Hoeven in conducting the EMG experiment is greatly acknowledged.
Appendix A

A.1 TRUSS element

Nodes of the TRUSS element are defined by position coordinates: $P(x,y,z)$ and $Q(x,y,z)$, Fig. A1. One deformation mode is associated with this element: elongation $\varepsilon_1$. This elongation can be expressed in the momentary values of the element position coordinates and the original length $l_0$:

$$\text{elongation: } \varepsilon_1 = \sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2} - l_0$$  \hspace{1cm} (A1)

![Figure A1: TRUSS element, reference and deformed position (Jonker, 1988).](image)

A.2 HINGE element

A HINGE element has two orientation nodes: $P(\lambda_0^p,\lambda_1^p,\lambda_2^p,\lambda_3^p)$ and $Q(\lambda_0^q,\lambda_1^q,\lambda_2^q,\lambda_3^q)$. An orientation node is defined by the orientation of a local coordinate system. To avoid singularity problems the orientation of the local coordinate system is described by orientation matrix $R$ expressed in terms of Euler parameters: $\lambda = (\lambda_0, \lambda_1, \lambda_2, \lambda_3)^T$:

$$R = \begin{bmatrix}
\lambda_0^2 + \lambda_1^2 - \lambda_2 - \lambda_3^2 & 2(\lambda_1 \lambda_2 - \lambda_0 \lambda_3) & 2(\lambda_1 \lambda_3 + \lambda_0 \lambda_2) \\
2(\lambda_1 \lambda_2 + \lambda_0 \lambda_3) & \lambda_0^2 - \lambda_1^2 + \lambda_2^2 - \lambda_3^2 & 2(\lambda_2 \lambda_3 - \lambda_0 \lambda_1) \\
2(\lambda_1 \lambda_3 - \lambda_0 \lambda_2) & 2(\lambda_2 \lambda_3 + \lambda_0 \lambda_1) & \lambda_0^2 - \lambda_1^2 - \lambda_2^2 + \lambda_3^2
\end{bmatrix}$$  \hspace{1cm} (A2)

Vectors $n_x$ of the hinge joints coincide with the hinge axis, the directions of the vectors $n_y$ and $n_z$ of both hinge nodes coincide initially (Fig. A2). A HINGE has three deformation modes: torsion around the initial axis, and two orthogonal bending deformations of this initial axis:
rotation: \( \varepsilon_1 = \text{ATAN2}(\langle R^p \vec{n}_y, R^q \vec{n}_z \rangle, \langle R^p \vec{n}_y, R^q \vec{n}_y \rangle) \) ;

bending: \( \varepsilon_2 = \langle R^p \vec{n}_y, R^q \vec{n}_x \rangle \) ;

\( \varepsilon_3 = \langle R^p \vec{n}_x, R^q \vec{n}_x \rangle \).

where \( \langle , \rangle \) denotes the scalar product of two vectors and \( \text{ATAN2}\{\sin(\phi), \cos(\phi)\} \) denotes the computer \( \text{ATAN2} \) function of two arguments.

By definition the Euler parameters satisfy the constraint equation

\( \lambda_0^2 + \lambda_1^2 + \lambda_2^2 + \lambda_3^2 = 1 \).

This constraint is included in the system by adding one deformation for each orientation node, the so-called LAMBDA element:

\( \varepsilon_1 = \lambda_0^2 + \lambda_1^2 + \lambda_2^2 + \lambda_3^2 - 1 \).

A.3 BEAM element

The BEAM element has four nodes: Two orientation nodes \([P (\lambda_0^P, \lambda_1^P, \lambda_2^P, \lambda_3^P)]\) and \([Q (\lambda_0^Q, \lambda_1^Q, \lambda_2^Q, \lambda_3^Q)]\) and two position nodes \([P (x^P, y^P, z^P); Q (x^Q, y^Q, z^Q)]\). Six deformation modes are associated with the BEAM element: elongation, torsion around the axis of the BEAM and two orthogonal bending deformations of each endpoint of the BEAM (Fig. A3).
Figure A3: BEAM element, initial and deformed state (Jonker, 1988).

\[
\epsilon_1 = \sqrt{(x_q - x_p)^2 + (y_q - y_p)^2 + (z_q - z_p)^2} - l_0
\]  (A8)

\[
\text{torsion} : \epsilon_2 = (\langle R_p n_x, R_q n_y \rangle - \langle R_p n_y, R_q n_z \rangle) \cdot l / 2 ;
\]  (A9)

\[
\text{bending} : \epsilon_3 = -\langle R_p n_z, l \rangle ;
\]  (A10)

\[
\epsilon_4 = \langle R_q n_x, l \rangle ;
\]  (A11)

\[
\epsilon_5 = \langle R_q n_y, l \rangle ;
\]  (A12)

\[
\epsilon_6 = -\langle R_q n_z, l \rangle .
\]  (A13)

where \( l = [x_q-x_p, y_q-y_p, z_q-z_p]^T \) and \( l_0 \) is the original length of the BEAM.

Appendix B

The SURFACE element has one position node and one deformation mode: The distance between the position node and the predefined surface, i.e. an ellipsoid representing the thorax with its axes
along the main axes of the coordinate system. The equation of such an ellipsoid with center \( M (m_x, m_y, m_z) \) and axes \( a_x, a_y \) and \( a_z \) for each point \( P (x, y, z) \) on the surface is

\[
\left( \frac{x - m_x}{a_x} \right)^2 + \left( \frac{y - m_y}{a_y} \right)^2 + \left( \frac{z - m_z}{a_z} \right)^2 = 1 .
\]  
(B1)

The distance is calculated along the normal vector \( N \) of the ellipsoid through the position node \( P (P_x, P_y, P_z) \), intersecting the ellipsoid at \( P' (P'_x, P'_y, P'_z) \). The line through the points \( P \) and \( P' \) is described by

\[
\begin{bmatrix}
P_x \\
P'_x \\
P_y \\
P'_y \\
P_z \\
P'_z
\end{bmatrix}
= \mu \begin{bmatrix}
P_x \\
P'_x \\
P_y \\
P'_y \\
P_z \\
P'_z
\end{bmatrix} + \mu \begin{bmatrix}
a_x^2 \\
2(P'_x - m_x) \\
a_y^2 \\
2(P'_y - m_y) \\
a_z^2 \\
2(P'_z - m_z)
\end{bmatrix}.
\]  
(B2)

Combining Eqs. (B1) and (B2) results in function \( F(\mu) \):

\[
F(\mu) = \left( \frac{P_x - m_x}{a_x - \frac{2\mu}{a_x}} \right)^2 + \left( \frac{P_y - m_y}{a_y - \frac{2\mu}{a_y}} \right)^2 + \left( \frac{P_z - m_z}{a_z - \frac{2\mu}{a_z}} \right)^2 - 1.
\]  
(B3)

The parameter \( \mu \) can be solved numerically from this equation when \( F(\mu) \) is zero, and with Eq. (B2) the coordinates of \( P' \) can be calculated. Consequently, the distance between \( P \) and \( P' \) is the distance from \( P \) to the ellipsoid:

\[
\varepsilon_1 = \| P - P' \|.
\]  
(B4)

The 3x1 vector of first derivatives is the normal vector \( N \) to the ellipsoid through \( P' \), normalised to a unit vector:

\[
\frac{\delta \varepsilon_1}{\delta P} = \frac{N}{\| N \|}.
\]  
(B5)
and the 3x3 matrix of second derivatives is

$$\frac{\delta^2 \epsilon_1}{\delta \mathbf{P} \cdot \delta \mathbf{P}^T} - \mathbf{I} - \left( \frac{\mathbf{N}}{\|\mathbf{N}\|} \cdot \frac{\mathbf{N}^T}{\|\mathbf{N}\|} \right). \quad (B6)$$

Appendix C

The CURVED-TRUSS element describes the mechanical effect of a TRUSS element between two position nodes \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \), which is wrapped around a predefined surface. \( \mathbf{P}_2 \) is attached to the same rigid body as the surface. The tangent points \( \mathbf{R}_1 \) and \( \mathbf{R}_2 \) of the shortest line between \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \) around the surface can be calculated (Fig. C1). The CURVED-TRUSS element has three position nodes, \( \mathbf{P}_1, \mathbf{P}_2 \) and \( \mathbf{R}_1 \), and one deformation mode: Elongation. The deformation is calculated analogously to the TRUSS element:

$$\epsilon_1 = \|\mathbf{P}_1 - \mathbf{R}_1\| + a(\mathbf{R}_1, \mathbf{R}_2) + \|\mathbf{R}_2 - \mathbf{P}_2\| - l_0, \quad (C1)$$

where \( a(\mathbf{R}_1, \mathbf{R}_2) \) denotes the distance between \( \mathbf{R}_1 \) and \( \mathbf{R}_2 \) over the surface and \( l_0 \) is the initial length of the element. Shortening (or lengthening) and force exertion of the CURVED-TRUSS element virtually takes place between \( \mathbf{P}_1 \) and \( \mathbf{R}_1 \), since \( \mathbf{R}_1, \mathbf{R}_2 \) and \( \mathbf{P}_2 \) are at the same rigid body. Then, by definition, \( \delta \epsilon_1/\delta \mathbf{P}_2 = 0 \), and the behavior of the CURVED-TRUSS element becomes

![Figure C1: The CURVED-TRUSS element. \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \) are the endnodes, \( \mathbf{R}_1 \) and \( \mathbf{R}_2 \) are the tangent points to the surface \( A \). The line through \( \mathbf{P}_1, \mathbf{R}_1, \mathbf{R}_2 \) and \( \mathbf{P}_2 \) is the shortest line between \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \) over surface \( A \). \( \mathbf{P}_2 \) and \( \mathbf{R}_1 \) are connected with BEAM elements with the local coordinate system with origin at the position node \( Q \) and the orientation defined by the orientation node \( \lambda \). Surface \( A \) is fixed in the local coordinate system.](image)
similar to a straight TRUSS element between $P_1$ and $R_1$ (see Appendix A).

At each step of the mechanism $R_1$ and $R_2$ are calculated. The surface is described with respect to the local coordinate system of the bone. If the bone is moving, the surface changes position and orientation accordingly. $P_2$ and $R_1$ are connected with rigid BEAM elements to the same local coordinate system as the surface. The BEAM connecting $R_1$ is repositioned each step according to the motion of $R_1$ with respect to the local coordinate system. In the trivial case that the line through $P_1$ and $P_2$ does not intersect the surface, the CURVED-TRUSS element behaves similar to a straight TRUSS element between $P_1$ and $P_2$.

Currently, three types of surfaces are used: a sphere, a cylinder and an ellipsoid. Calculation of the tangent points $R_1$ and $R_2$ at these surfaces will be presented in Section C.1, C.2 and C.3, respectively.

C.1 Sphere

In the special case of a sphere as intervening object, tangent points $R_1$ and $R_2$ are in the same plane as $P_1$, $P_2$ and center $M$ of the sphere, and can be calculated analytically (see Fig. C2). Angle $\alpha_1$ between $R_1$, $P_1$ and $M$ is

$$\alpha_1 = \arcsin(r/|M-P_1|)$$

(C2)

where $r$ is the radius of the sphere. The vector $(R_1 - P_1)$ can be calculated by rotating vector

Figure C2: Calculation of the shortest line between $P_1$ and $P_2$ around a sphere.

$M$: center of the sphere.
$r$: radius
$R_1, R_1'$: Tangent points of $P_1$.
$R_2, R_2'$: Tangent points of $P_2$. 
(M - P₁) with rotation matrix R(n, α₁), i.e. a rotation with angle α₁ around axis n (perpendicular to the plane through P₁, P₂, M), and scaling with cos(α):

\[(R₁ - P₁) = R(n, α₁) \cdot (M - P₁) \cdot \cos(α₁)\]  \hspace{1cm} (C3)

If (M - P₁) is rotated with angle -α₁, the other tangent point R₁' in the plane is found. Which one (R₁ or R₁') provides the shortest distance can be calculated afterwards. Similar calculations are performed for R₂ and the shortest distance can be calculated.

C.2 Cylinder

The shortest line over a surface between two points on that surface is called the geodesic line. Property of each point of the geodesic line is that the normal vector to the surface is situated in the plane of curvature of the line (Grossman, 1976). However, for second-order surfaces like a cylinder and ellipsoid the orientation of the plane of curvature is changing along the line. Hence, calculation of the shortest line between P₁ and P₂, which consists of calculation of tangent points R₁ and R₂ and the geodesic line between them, will be a time consuming iterative procedure. Since for each muscle element and in each position the geodesic line is recalculated, a faster algorithm is used to approximate the geodesic line by assuming that the whole line is in one plane. Afterwards, the angle between the normal vectors of each point of the approximated geodesic line and the plane of curvature in which this line is situated, is calculated in order to check the correctness of the assumption.

In order to obtain the shortest line between P₁ and P₂ around a cylinder, tangent points R₁ and R₂ must be calculated (see Fig. C3). The cylinder is defined by its central axis with position vector O and unit direction vector d, and by its radius r. The shortest line, including P₁, P₂, R₁ and R₂, is assumed to be in one plane. Normal vector n to this plane can be calculated by the double vector or cross product:

\[n = (P₁ - P₂) \times d \times (P₁ - P₂)\]  \hspace{1cm} (C4)

Vector N₁, perpendicular to the central axis and through R₁, can be calculated by the following three equations:

1) It is perpendicular to the direction vector d of the central axis, i.e. the scalar product of both vectors is zero:
2) It is the normal vector to the tangent plane of \( P_1 \) to the cylinder:

\[
P_{1x} \cdot N_{R1x} + P_{1y} \cdot N_{R1y} + P_{1z} \cdot N_{R1z} + 1 = 0.
\]  

(C6)

3) A parallel plane which contains the central axis of the cylinder is at distance \( r \) (radius) from the previously described tangent plane, i.e. position vector \( Q (Q_x, Q_y, Q_z) \) of the central axis should be on this plane:

\[
Q_x \cdot N_{R1x} + Q_y \cdot N_{R1y} + Q_z \cdot N_{R1z} + r + 1 = 0.
\]  

(C7)

\( R_1 \) is the single intersection point of the tangent line of \( P_1 \) and the cylinder. The tangent line is

---

**Figure C3:** Calculation of the shortest line between \( P_1 \) and \( P_2 \) around a cylinder.

- \( d \): direction vector of central axis.
- \( Q \): position vector of central axis.
- \( n \): normal vector to the plane through \( P_1, R_1, R_2, \) and \( P_2 \).
- \( N_{R1} \): vector perpendicular to \( d \) and through \( R_1 \).
- \( N_{R2} \): vector perpendicular to \( d \) and through \( R_2 \).
- \( R_1 \): Tangent point of \( P_1 \).
- \( R_2 \): Tangent point of \( P_2 \).
Finite element musculoskeletal model

described by

\[ \mathbf{R}_1 = \mathbf{P}_1 + \lambda_1 \cdot \mathbf{L}_1. \]  \hspace{1cm} (C8)

Direction vector \( \mathbf{L}_1 \) is perpendicular to \( \mathbf{n} \) and \( \mathbf{N}_{R1} \):

\[ \mathbf{L}_1 = \mathbf{n}_1 \times \mathbf{N}_{R1} \]  \hspace{1cm} (C9)

\( \mathbf{R}_1 \) is on the surface of the cylinder which is described by

\[ \left[ (R_x - O_x) \cdot d_x - (R_y - O_y) \cdot d_y \right]^2 + \left[ (R_y - O_y) \cdot d_x - (R_z - O_z) \cdot d_y \right]^2 + \left[ (R_z - O_z) \cdot d_x - (R_x - O_x) \cdot d_y \right]^2 - r^2 = 0. \]  \hspace{1cm} (C10)

\( \lambda_1 \) is calculated by substituting Eq. (C8) in Eq. (C10), and subsequently \( \mathbf{R}_1 \) is calculated. \( \mathbf{R}_2 \) can be calculated similarly.

When the assumption is checked if normal vectors to the geodesic line are approximately in the plane of curvature, it was revealed that during humeral motions the cylinder hardly plays a role as bony contour. Whenever a muscle line of action is wrapped around it, normal vectors to the cylinder are angular to the plane of curvature only up to 2 degrees.

C.3 Ellipsoid

An ellipsoid with its axes \( a_x, a_y, a_z \) in the main planes of the coordinate system and center \( \mathbf{M} (m_x, m_y, m_z) \) is defined for each point \( \mathbf{I} (x, y, z) \) on the surface by (Fig. C4):

\[ \left[ \frac{x - m_x}{a_x} \right]^2 + \left[ \frac{y - m_y}{a_y} \right]^2 + \left[ \frac{z - m_z}{a_z} \right]^2 = 1. \]  \hspace{1cm} (C11)

To solve for the tangent points \( \mathbf{R}_1 \) and \( \mathbf{R}_2 \) at the ellipsoid of respectively \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \) four equations can be posed:

1) It is assumed that the geodesic line is in one plane. \( \mathbf{P}_1, \mathbf{P}_2 \) and one point of the geodesic line are needed to define the plane. The line through \( \mathbf{P}_1 \) and \( \mathbf{P}_2 \) is intersecting one of the main planes of the ellipsoid in \( \mathbf{C} \). The normal vector to the ellipsoid through \( \mathbf{C} \) is one of the normal vectors of the geodesic line. Then, the projection \( \mathbf{C}' \) of \( \mathbf{C} \) to the surface of the ellipsoid along the normal vector is the third point of the plane with normal vector \( \mathbf{n} (n_x, n_y, n_z) \) defined by:
\[ n = (P_1 - C') \times (P_2 - P) , \]  \hspace{1cm} \text{(C12)}

and scaled to a unit vector. \( R_1 \) is also in this plane:

\[ n_x R_{1x} + n_y R_{1y} + n_z R_{1z} + D_1 = 0 , \]  \hspace{1cm} \text{(C13)}

where \( D_1 \) is defined by the fact that e.g. \( P_1 \) is in the same plane:

\[ D_1 = - n_x P_{1x} - n_y P_{1y} - n_z P_{1z} . \]  \hspace{1cm} \text{(C14)}

2) \( R_1 \) is on the surface of the ellipsoid:

\[ \left( \frac{R_{1x} - m_x}{a_x} \right)^2 + \left( \frac{R_{1y} - m_y}{a_y} \right)^2 + \left( \frac{R_{1z} - m_z}{a_z} \right)^2 = 1 . \]  \hspace{1cm} \text{(C15)}

3) The normal vector \( N_{R1} \) through \( R_1 \) is the normal to a tangent plane of \( P_1 \) to the ellipsoid. Hence, for \( R_1 \):

\[ \]  

Figure C4: Calculation of the shortest line between \( P_1 \) and \( P_2 \) around an ellipsoid.  
\( M \): center of the ellipsoid.  
\( A_x, A_y, A_z \): Axes of the ellipsoid.  
\( C \): Intersection of \( P_1 \) and \( P_2 \) through the main plane of the ellipsoid.  
\( C' \): Projection of \( C \) to the ellipsoid, intersection of the geodesic line and the main plane of the ellipsoid.  
\( n \): normal vector of the ellipsoid through \( C \).
Finite element musculoskeletal model

\[ N_{R1x} \cdot R_{lx} + N_{R1y} \cdot R_{ly} + N_{R1z} \cdot R_{lz} + D_2 = 0 \]  \tag{C16}

where

\[ N_{R1} = \begin{bmatrix} \frac{2(R_{lx} - m_x)}{a_x^2} \\ \frac{2(R_{ly} - m_y)}{a_y^2} \\ \frac{2(R_{lz} - m_z)}{a_z^2} \end{bmatrix}, \]  \tag{C17}

4) and for \( P_1 \):

\[ N_{R1x} \cdot P_{lx} + N_{R1y} \cdot P_{ly} + N_{R1z} \cdot P_{lz} + D_2 = 0 \]  \tag{C18}

These four non-linear equations can be solved for \( R_1 \) and \( D_2 \) using a Gauss-Newton approximation. In the same way \( R_2 \) can be calculated. Checking afterwards, all normal vectors to the points of the geodesic line are approximately (within one degree) in the plane of curvature.

Appendix D

Constraint of the glenohumeral joint

In the cadaver study a sphere was fitted to data points recorded at the articular surface of the glenoid (Van der Helm et al., 1991). Data points at the rim of the glenoid were indicated. The direction vector from the center of the sphere to these data points can be described by pole coordinates: A rotation \( \phi \) around the global Y-axis (angle with the frontal plane) and subsequently a rotation \( \theta \) around the rotated \( z' \)-axis (angle with the transversal plane). An ellips with center \( M \) (\( \phi_m, \theta_m \)) and axes \( a_\phi \) and \( a_\theta \) can be fitted to the obtained rotation angles using a least squares criterion, describing the orientation and shape of the glenoid (Fig. D1). The orientation of the glenoid changes with the motions of the scapula.

Then, the optimization procedure is applied without the constraint that the resultant force vector at the humerus should point into the glenoid. After the optimization procedure the direction of the
resultant force vector is described by the rotation angles $\phi_f$ and $\theta_f$. If the point $F(\phi_f, \theta_f)$ is outside the ellipsoid (Fig. D1), this point is projected to the ellips along the normal vector, analogously to Appendix B. Hence, the projection $F'$ is obtained and the direction of the resultant force vector at the humerus will be constrained to point from the rotation center of the glenoid to $F'$ in a renewed application of the optimization procedure.

**Figure D1:** An ellipsoid with center $M(\phi_m, \theta_m)$ and axes $a_\phi$ and $a_\theta$ is fitted to the pole coordinates $\phi$ and $\theta$ of the datapoints ('x') on the rim of the glenoid cavity with respect to the rotation center of the GH-joint. If the pole coordinates of the resultant muscle force vector $F(\phi_f, \theta_f)$ are outside the ellips, the muscle force vector is constrained to point through $F'(\phi_f, \theta_f)$ which is projected on the ellips along the normal through $F$. 
Chapter 6
Analysis of the kinematic and dynamic behavior of the shoulder mechanism.

F.C.T. van der Helm


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Abstract
A finite element musculoskeletal model of the shoulder mechanism consisting of the thorax, clavicle, scapula and humerus, has been used for analysis of the kinematic and dynamic behavior. The model includes 16 muscles, 3 joints, 3 extracapsular ligaments and the motion constraints of the scapulothoracic gliding plane which turns the shoulder girdle into a closed-chain mechanism. Simulations are inverse dynamic. Input variables are the positions of the shoulder girdle and humerus which have been recorded in 10 subjects during unloaded and loaded humeral abduction and anteflexion. Comparisons of muscle force predictions and EMG recordings are hampered by the unknown force-length relationship and the length dependency of EMG amplitude. It is concluded that EMG amplitude can not be used for validation of complex musculoskeletal models. Muscle function is analysed with help of a force and moment balance of the three joints. The moment balance includes the contributions of ligaments and the reaction forces at the scapulothoracic gliding plane. The scapulothoracic gliding plane is very important for the motions and the stabilization of the shoulder girdle. The direction and magnitude of joint reaction forces are calculated as well. It is concluded that the model provides good insight in the mechanics of the shoulder mechanism and that it enables an analysis of the function of morphological structures.
I. Introduction

The shoulder mechanism is a chain of bones connecting the upper extremity to the trunk. In a thorough analysis the following four bony elements should be included: Thorax (assumed to be rigid), clavicle, scapula and humerus. The clavicle and scapula form the shoulder girdle. In contrast to the pelvic girdle, a considerable range of motion is available for the shoulder girdle. Motions of the shoulder girdle, i.e. motions of the sternoclavicular (SC-) and acromioclavicular (AC-) joint, contribute to the large range of motion of the upper extremity. The shoulder mechanism is a multifunctional joint. The number of functions of the upper extremity is almost infinite. It ranges from manipulating objects on a desk top to throwing a ball and lifting heavy objects. The existence of these tasks are feasible through the large range of motion and the number of degrees of freedom (DOF) available in the shoulder mechanism (Fig. 1).

In addition, if the hand is in a certain position, it is possible to exert forces in almost any direction. Muscle forces around the glenohumeral (GH-) joint result in net joint moments, counterbalancing external load, gravitational and inertial forces. Moments and forces are transmitted to the trunk through the bones of the shoulder girdle and muscles crossing its joints. The force analysis is complicated due to poly-articular muscles and the connection between scapula and thorax, the so-called scapulothoracic gliding plane, which turns the shoulder mechanism into a closed-chain mechanism.

Muscles in the shoulder mechanism can be divided into four groups: Prime movers of the GH-joint, rotator cuff muscles, bi-articular upper arm muscles and thoracoscapular muscles. Around

Figure 1: Motions of the humerus are the result of the simultaneous motions of the sternoclavicular, acromioclavicular and glenohumeral joint, dorsal view (after Meijers, 1961).
the GH-joint four large muscles with large moment arms can be discerned (Fig. 2A). They are considered as the prime movers of the humerus. M. deltoideus, divided in a mono-articular scapular part and a bi-articular clavicular part, originates from a half circle formed by the scapular spine and clavicle, and inserts into the tuberculum deltoideum at the lateral side of the humerus. M. pectoralis major and m. latissimus dorsi are poly-articular muscles, crossing all three joints of the shoulder mechanism. M. pectoralis major is in front of the body, partly originating from the clavicular and partly from the thorax. M. latissimus dorsi has its origin at the spine and the pelvic girdle. M. teres major can functionally be regarded as a mono-articular short head of m. latissimus dorsi, originating near the angulus inferior of the scapula and sharing a tendon with m. latissimus

Figure 2: Muscles of the shoulder mechanism:
A: Prime movers of the glenohumeral joint (lateral view).
B: Rotator cuff muscles (lateral view).
C: Thoracoscapular muscles (dorsal view, m. pectoralis minor and m. subclavius not shown).
dorsi at its insertion at the medioventral side of the humeral shaft.

The four muscles of the rotator cuff insert as a half circle close to the GH-joint (Fig. 2B). M. subscapularis is in front, m. supraspinatus is above and m. infraspinatus and m. teres minor are dorsal to the joint. Their main function is generally assumed to stabilize the joint in any position. M. biceps and m. triceps are two bi-articular muscles crossing the elbow joint and GH-joint. M. triceps has two mono-articular parts crossing only the elbow joint. M. biceps is divided in a short head, originating at the processus coracoideus of the scapula and a long head, originating at the top of the glenoid cavity by a tendon which passes through the sulcus bicipitis. M. coracobrachialis originates at the same location as the short head of m. biceps, and inserts at the medial side of the humeral shaft. Innervated by the same nerve as m. biceps, it can be regarded morphologically and functionally as a real mono-articular short head of the biceps.

The scapula and clavícula are connected to the thorax by six muscles (Fig. 2C). The largest muscles are m. serratus anterior, originating at the front of the thorax and inserting at the medial border of the scapula, and m. trapezius, originating at the spine and the head and inserting partly at the scapular spine and partly at the clavícula. Smaller muscles at the back side are m. rhomboideus, originating at the spinous processus and inserting at the medial border of the scapula, and m. levator scapulae, running from the cervical spine to the angulus superior of the scapula. M. pectoralis minor is situated at the front side, connecting the processus coracoideus to the thorax. One small muscle, m. subclavius, runs almost parallel to the clavícula from thorax to the clavícula.

Summarizing, two mono-articular muscles are crossing the SC-joint, the clavicular part of m. trapezius and m. subclavius. No mono-articular muscles are crossing the AC-joint and seven mono-articular muscles are crossing the GH-joint: the scapular part of m. deltoideus, m. teres major, m. coracobrachialis and the four muscles of the rotator cuff. All other muscles are poly-articular (see Table 1).

Three extracapsular ligaments restrict the motions and redistribute muscle forces. The costoclavicular ligament, crossing the SC-joint, originates at the first rib and inserts at the medial curvature of the clavícula. The conoid ligament originates at the dorsal side of the lateral clavicular curvature and inserts at the processus coracoideus. The trapezoid ligament, almost perpendicular to the conoid ligament, originates at the most lateral point of the clavícula and inserts near the conoid ligament. The latter two ligaments are crossing the AC-joint.

Generally, muscle function is derived based on a kinematic analysis (Fick & Weber, 1877; Braune & Fischer, 1888; Fick, 1911). Starting from the anatomical position, bones are imaginary set in
Table 1: Origin and insertion of the muscle at the bones of the shoulder mechanism (*: insertion at the forearm bones).

<table>
<thead>
<tr>
<th></th>
<th>Insertion clavica</th>
<th>Insertion scapula</th>
<th>Insertion humerus</th>
</tr>
</thead>
<tbody>
<tr>
<td>Origin thorax</td>
<td>trapezius, clav. part subclavius</td>
<td>trapezius, scap. part serratus anterior levator scapulae rhomboideus pectoralis minor</td>
<td>pect. major, thor. part latissimus dorsi</td>
</tr>
<tr>
<td>Origin clavicula</td>
<td></td>
<td></td>
<td>deltoideus, clav. part pect. major, clav. part</td>
</tr>
<tr>
<td>Origin scapula</td>
<td></td>
<td></td>
<td>deltoideus, scap. part subscapularis supraspinatus infraspinatus teres minor teres major coracobrachialis biceps, caput breve * biceps, caput longum* triceps, caput longum*</td>
</tr>
</tbody>
</table>

motion and if a muscle is considered to shorten, its function is named after the motion. In the shoulder region lateral rotators, medial rotators, protractors, retractors, elevators and depressors are found. Since no forces are included in the analysis, results of this kinematic approach are limited. For each motion a number of muscles are co-operating. The contribution of each individual muscle and the significance for the whole motion is not revealed. In addition, during the motion muscle moment arms are changed, especially for the shoulder mechanism with its large range of motion, and hence the functional role of muscles will change. Therefore, not only the anatomical position is of interest, but the whole range of motion. The combined action of muscles and simultaneous motion at several joints are especially important to assess the function of poly-articular muscles.

During the last fifty years electromyography (EMG) has been used for assessing muscle function (e.g. Inman et al., 1944; Saha; 1961). Mainly surface EMG was involved which evidently restricts the analysis to superficial muscles. Only a small number of muscles was analysed simultaneously and the performed motion as well as electrode position were in general insufficiently standardized (Van der Hoeven, 1988). Since the EMG-force relationship is unknown for shoulder muscles and
EMG amplitude is length dependent (Heckathorne & Childress, 1981), EMG analysis is merely qualitative from a mechanical point of view.

Thus far, a biomechanical analysis of muscle function in the shoulder mechanism is not yet published. Hogfors et al. (1987), Karlsson et al. (1989), Karlsson (1990), Wood et al. (1989ab) and Basset et al. (1990) proposed a biomechanical modelling approach. They were bogged down in derivation of the parameters describing the geometry of the musculoskeletal system and did not present an actual model. From the nature of the geometrical measurements performed, i.e. centroid line assessment for muscle function, the models of Wood et al. (1989ab) and Bassett et al. (1990) will be restricted to a static analysis in one position. However, a thorough analysis should depart from the change of position of the shoulder girdle during humeral motions. Position changes will cause change of muscular moment arms which is likely to cause a change in the role of the muscles during the motion. In order to obtain a realistic picture of the function of morphological structures, an integral model including all morphological structures present is necessary. Then, a complete force and moment balance can be derived. Especially the role of the scapulothoracic gliding plane must be stressed, since it is neglected in the models presented until now. The thorax not only restricts scapular motions, but in addition it exerts forces at the scapula.

In this study a finite element model of the shoulder mechanism is used (Van der Helm, 1991a). All morphological structures are represented by elements of which the mechanical behavior is well-known. Parameters describing the position and behavior of the elements have been derived in an extensive cadaver study (Veeger et al., 1991a; Van der Helm et al., 1991; Van der Helm & Veenbaas, 1991). Positions used as input for the model have been recorded in ten healthy subjects during loaded and unloaded abduction and anteflexion of the humerus (Van der Helm & Pronk, 1991a). Muscle forces calculated by the model are compared with EMG-signals recorded in twelve subjects of which eight also participated in the motion recording experiment. The model allows an analysis of forces and moments exerted by individual muscles to maintain the desired humeral position and counterbalance the external load.

II. Finite element model

An extensive description of the finite element model of the shoulder mechanism is published in Van der Helm (1991a). In the finite element model each relevant morphological structure is represented by an appropriate element. Since the mechanical behavior of these elements is well-known, the mechanical behavior of the whole mechanism can be assessed by simply connecting the elements. Here, a brief outline of the model will be presented.
Analysis of the kinematic and dynamic behavior

In the model the clavícula and humerus are each represented by a single rigid BEAM element. The scapula is represented by two tightly connected BEAM elements, providing two points at the medial border. Joints are represented as spherical joints by three perpendicular HINGE elements. Motion constraints of the scapulothoracic gliding plane are represented by two SURFACE elements, each restricting a point of the medial border of the scapula to follow an ellipsoid representing the thorax. Extracapsular ligaments are represented by flexible TRUSS elements. Muscles are represented by force-generating 'active' TRUSS and CURVED-TRUSS elements which connect origin and insertion and exert pulling forces on its attachments. TRUSS elements model a straight muscle line of action, CURVED-TRUSS elements model muscle lines of action which are wrapped around underlying morphological structures such as bony contours. In order to adequately model the mechanical effect each muscle is represented by one to six elements (Van der Helm & Veenbaas, 1991).

Parameters of the model are the inertia parameters, describing the mass and rotational inertia of the upper extremity, muscle parameters, i.e. the physiological cross-sectional area (PCSA) which provides an estimate of the maximal muscle force (37 N/cm², Weijis & Hillen, 1985), and geometry parameters, describing the position coordinates of muscle and ligament attachments, joint rotation centers and bony landmarks, and the position and shape of bony contours determining the muscle lines of action. All parameters are derived in an extensive cadaver study (Veeger et al., 1991a; Van der Helm et al., 1991; Van der Helm & Veenbaas, 1991).

Motion equations of the mechanism are numerically derived using generalized variables, comparable with the Lagrangian approach. The number of generalized variables is equal to the number of degrees of freedom (DOF). The shoulder mechanism model has seven DOF: four at the shoulder girdle and three at the GH-joint. Since axial rotation of the clavícula could not be recorded in the motion recording study (Van der Helm & Pronk, 1991a), it is decided to represent the conoid ligament by a rigid TRUSS element, thereby only affecting the axial rotation and no other motions of scapula and clavícula and decreasing the number of DOF by one.

All input and output variables are presented with respect to the coordinate system fixed in the thorax. Origin of the coordinate system is Incisura Jugularis (IJ), with the X-axis medial to lateral, Y-axis caudal to cranial and Z-axis ventral to dorsal (Fig. 3). Input variables were derived in a motion recording study (Van der Helm & Pronk, 1991a). Bony landmarks of the shoulder mechanism were palpated and their position recorded using the palpator, a kind of spatial digitizer (Pronk & Van der Helm, 1991). With the position of the bony landmarks the orientation of bones could be reconstructed. The mean clavicular, scapular and humeral orientations were derived for
ten healthy subjects. The mean clavicular and scapular orientations were used in combination with the bony dimensions of the cadaver to calculate the position of the bony landmarks. However, in the model this position could not be reached for all bony landmarks without violating the constraints imposed by the thorax. Therefore, some position coordinates, i.e. the y- and z-coordinate of the most dorsal palpable point of the AC-joint, resp. AC_y and AC_z, and the x-coordinate of trigonum spinae (TS_x), were used as input variables, and the clavicular and scapular positions were calculated according to the formulated motion constraints. The humeral position is not constrained and thus the mean rotation angles could be used as input. Position measurements of the shoulder mechanism could only be performed statically. Hence, this simulation study is essentially quasi-static. It should be noted that the finite element model used is capable of dynamic and inverse dynamic simulations.

Four types of humeral motion have been analyzed in this simulation study: abduction unloaded (ABU), abduction loaded (ABL), anteflexion unloaded (ANU) and anteflexion loaded (ANL). Loaded humeral motion consisted of carrying a weight of 750 grams in the hand during the motion recording. Though analysis of loaded vs. unloaded motions revealed that no significant change of motion occurred, the actually recorded positions for each condition were used as input. It is important to consider that scapular motions are closely related to humeral motions (Inman et al., 1944; Pronk, 1987; Van der Helm & Pronk, 1991b). Analysis of the motion and force balance at one joint, e.g. the GH-joint, is not very fruitful without considering the motions of the proximal bone, i.e. the scapula. Muscle function, and especially the function of poly-articular muscles, depends on rotations of all joints crossed. Therefore, the simultaneous motion of the complete

Figure 3: Global coordinate system. Origin at Incisisura Jugularis, X-axis pointing from medial to lateral, Y-axis from caudal to cranial and Z-axis from ventral to dorsal (right shoulder).
shoulder mechanism was used as input. Output variables of the model are the resultant orientations of clavícula and scapula, and the muscle forces necessary to counterbalance the external load on the upper extremity (and inertial load in case of dynamic simulations). Generally, more force generating muscle elements are present than the number of DOF. A solution can be obtained using an optimization procedure. In literature, a number of optimization criteria has been proposed (Seirig & Arvick, 1973; Hardt, 1978; Dul et al., 1984ab; Davy & Audu, 1987). Non-linear criteria are preferred in order to enforce synergism between agonistic muscles. In a recent study results of several criteria were compared with EMG recordings, and preference has been given to minimization of the sum of squared muscle stresses in the entire system, though other criteria as minimization of the maximal muscle stress or squared quotient between muscle force and maximal, length dependent, muscle force did not lead to clearly different results (Van der Helm, 1991a). If all muscle forces in the system are known, the resultant joint reaction forces, ligament forces and forces acting at the scapulothoracic gliding plane can be calculated.

III Analysis and results
III.1 Kinematic analysis
As a typical example, Figs 4A, 4B and 4C show stick diagrams of the recorded motion of the shoulder girdle during unloaded humeral anteflexion (ANU). A line connecting IJ and AC represents the clavícula, whereas the scapula is represented by a triangle connecting AC, TS and AI. Throughout the motion the clavícula is rotating upwards and backwards, except for 180 degrees anteflexion. TS and AI move caudally and laterally immediately from the start of the motion, resulting in a lateral rotation of the scapula. Then, AI moves ventrally, until it ends besides the thorax, underneath and ventral to AC.

Rotation of the bones has been described using Euler angles, i.e. three subsequent rotations around perpendicular axes, starting from the anatomical position. Local coordinate systems are used to define the orientation of the bones. The \( x_c \)-axis of the clavicular coordinate system is along its length axis, the \( z_c \)-axis is perpendicular to the \( x_c \)-axis and the global Y-axis, pointing dorsally, and the \( y_c \)-axis is defined by the \( x_c \)- and \( z_c \)-axis. The \( x_s \)-axis of the scapula is along the scapular spine (AC-TS), pointing to TS, the \( y_s \)-axis is in the scapular plane defined by AC-TS-AI, pointing caudally, and the \( z_s \)-axis is perpendicular to the scapular plane, pointing dorsally. The \( y_h \)-axis of the humerus is along its length axis, pointing from the rotation center of the GH-joint to the midpoint between the lateral and medial epicondyle. The \( z_h \)-axis is perpendicular to the \( y_h \)-axis and
the line through both epicondyles, pointing ventrally and the $x_h$-axis is defined by the $y_h$- and $z_h$-axis, roughly pointing from the medial to the lateral epicondyle.

Rotation of the bones has been described with respect to a virtual reference position (Van der Helm & Pronk, 1991\textsuperscript{a}). In the virtual reference position the clavicle is along the frontal axis, with its local coordinate system identical to the global coordinate system. The clavicular rotations are defined as: First a rotation around the global Y-axis (ventral/dorsal rotation or pro/retraction), then rotation around the local $z'_c$-axis (cranial/caudal rotation or elevation/depression) and finally rotation around the local $x'_c$-axis (axial rotation). Quotation marks ' and '' indicate that these axes are not the original axes but local axes rotated by one or two prior rotations, respectively. The virtual reference position of the scapula is defined by the spine ($x_s$-axis) along the frontal axis and the scapular plane ($x'_s$- and $y'_s$-axis) parallel to the frontal plane. Scapular rotations are defined as subsequently around the global Y-axis (pro/retraction), around the local $z'_s$-axis (lateral/medial.

![Figure 4: Stick diagram of the recorded motion of the clavicle (IJ-AC) and scapula (AC-TS-AI) during unloaded humeral anteflexion at 0 and 180 degrees. Small circles indicate intermediate positions. A: Projection frontal plane. B: Projection sagittal plane. C: Projection transversal plane.](image-url)
rotation) and finally around the local $x_s^e$-axis (tipping forward/backward). The virtual reference position of the humerus is defined by its length axis ($y_h$-axis) along the vertical axis and the $x_h$-axis in the frontal plane. Rotations are defined as subsequently around the Y-axis (pole angle), the local $z_h^e$-axis (elevation) and the local $y_h^e$-axis (axial rotation).

Figs 5A to 5E show recorded and simulated rotations of the bones of the shoulder girdle. The rotation angles calculated in the model, as a result of the input variables, are compared to the mean as measured at the subjects, for the four simulated conditions (ABU, ABL, ANU, ANL). The scapular position is very sensitive to the scapular orientation. Due to the motion constraints imposed by the clavicle and the thorax, a small change in orientation results in irrational positions of the scapula. Therefore, position coordinates were used as input variables. Still, the resultant rotation angles deviate up to 10 degrees from the mean. Mainly, the lateral rotation of the scapula is somewhat out of range (Fig. 5C), whereas the pro/retraction and tipping forward/backward show smaller deviations.

Since ACy and ACz were derived using the mean rotation angles, obviously the simulated pro/retraction and elevation of the clavicle are almost identical to the recorded motion (Figs 5A and 5B, respectively). It is more surprising that the simulated axial rotation of the clavicle closely follows the mean axial rotation. In the motion recording study, axial rotation of the clavicle could not be recorded, since only two bony landmarks on the clavicle could be distinguished. Axial rotation was estimated by minimizing rotations in the AC-joint. In the simulation study axial rotation is caused by the conoid ligament which was assumed to be rigid. Thus, it can be concluded that the conoid ligament very much acts to minimize the rotations in the AC-joint, resulting in rotations of less than 10 degrees per axis.

Humeral rotations, as shown in Fig. 5E, were used as input for the simulation. Rotation axes were chosen in order to achieve a well interpretable motion description. The pole angle around the global Y-axis defines the plane in which the humerus is elevated and is very sensitive near 0 and 180 degrees elevation. The elevation angle around the local $z_h$-axis describes the angle with the vertical axis. The remaining rotation is the axial rotation around the length axis of the humerus. In the anatomical position axial rotation is identical with endo/exorotation and ranges from 60 degrees endorotation to 90 degrees exorotation. However, when the humerus is elevated and in a position more perpendicular to the glenoid cavity, it is important to include a good estimate of the axial rotation. In that position axial rotation causes merely a torsion on the rotator cuff muscles, thereby changing the direction of the muscle moment vector and thus to a large extent of muscle function.
Figure 5: Comparison of simulated and recorded rotations. A: Simulated rotations clavicle; B: Recorded and estimated rotations clavicle; C: Simulated rotations scapula; D: Recorded rotations scapula; E: Recorded rotations humerus, which are identical to the simulated rotations.

- - - - - : abduction unloaded (ABU).
- - - - - - : anteflexion unloaded (ANU).
- - - - - - - : abduction loaded (ABL).
- - - - - - - - : anteflexion loaded (ANL).
Table 2: Number of elements, physiological cross-sectional area (PCSA) per element and location order of the elements for future reference in figures, for the muscles of the shoulder mechanism.

<table>
<thead>
<tr>
<th>muscle</th>
<th># elements</th>
<th>PCSA/element (cm²)</th>
<th>element order</th>
</tr>
</thead>
<tbody>
<tr>
<td>trapezius, scap. part</td>
<td>6</td>
<td>2.39</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>trapezius, clav. part</td>
<td>6</td>
<td>0.52</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>levator scapulae</td>
<td>3</td>
<td>1.15</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>pectoralis minor</td>
<td>4</td>
<td>0.86</td>
<td>medial - lateral</td>
</tr>
<tr>
<td>rhomboideus</td>
<td>3</td>
<td>2.52</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>serratus anterior</td>
<td>6</td>
<td>1.91</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>deltoideus, scap. part</td>
<td>6</td>
<td>2.76</td>
<td>medial - lateral</td>
</tr>
<tr>
<td>deltoideus, clav. part</td>
<td>6</td>
<td>1.35</td>
<td>lateral - medial</td>
</tr>
<tr>
<td>coracobrachialis</td>
<td>6</td>
<td>0.53</td>
<td>lateral - medial</td>
</tr>
<tr>
<td>infraspinatus</td>
<td>6</td>
<td>1.36</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>teres minor</td>
<td>6</td>
<td>0.52</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>teres major</td>
<td>6</td>
<td>2.09</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>supraspinatus</td>
<td>6</td>
<td>0.78</td>
<td>ventral - dorsal</td>
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<tr>
<td>subscapularis</td>
<td>6</td>
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<td>cranial - caudal</td>
</tr>
<tr>
<td>biceps, cap. longum</td>
<td>1</td>
<td>1.72</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>biceps, cap. breve</td>
<td>1</td>
<td>1.78</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>triceps</td>
<td>2</td>
<td>3.12</td>
<td>medial - lateral</td>
</tr>
<tr>
<td>latissimus dorsi</td>
<td>5</td>
<td>1.70</td>
<td>cranial - caudal</td>
</tr>
<tr>
<td>pect. major, thor. part</td>
<td>5</td>
<td>1.74</td>
<td>caudal - cranial</td>
</tr>
<tr>
<td>pect. major, clav. part</td>
<td>5</td>
<td>0.71</td>
<td>medial - lateral</td>
</tr>
</tbody>
</table>

Each muscle is represented by one to six elements representing muscle lines of action. Table 2 presents for each muscle the number of elements, location of numbered elements and PCSA. Figs 6A to 6T show the length of the muscle elements representing the muscle lines of action during unloaded humeral abduction (ABU). For a number of muscles the length difference between elements of the same muscle is remarkable, especially for muscles with two large line-shaped attachment sites, like m. trapezius and m. serratus anterior, and for muscles with plane-shaped attachment sites like m. subscapularis and m. infraspinatus. If muscle length in the initial position (0 degrees humeral elevation) is defined as 100%, length changes vary between 52% and 167%.
This is much larger than length changes in e.g. leg muscles during walking (Frigo & Pedotti, 1977; Pierrynowski, 1982). Optimal muscle length, i.e. the length at which a muscle can exert maximal force, is difficult to assess in vivo, and certainly for shoulder muscles. However, it can be assumed that optimal muscle length is near the length on which the muscle is active most of the time (Goldspink, 1988) which means for shoulder muscles probably at small humeral elevation angles. Assuming a normal force-length relationship ranging from 60% to 140% of optimum muscle length (Woittiez, 1984), this would imply that some muscles, e.g. m. serratus anterior and m. deltoideus, medial part, could not exert any force at 180 degrees elevation angle. In the same position, some muscles, e.g. m. rhomboideus and m. pectoralis minor, are stretched up to 167% of their initial length, normally resulting in a huge amount of passive elastic force. The large range of motion of the bones of the shoulder mechanism, in combination with the fact that especially all thoracoscapular and thoracohumeral muscles are at quite a distance from the joint rotation centers, causes these extraordinary length changes. Therefore, it is expected that the active force-length relationship covers a wider range of muscle length, and that passive muscle forces are negligible in the normal physiological range of shoulder motions.

III.2 Dynamic analysis

In the model of the shoulder mechanism each muscle is represented by 1 to 6 muscle elements, adequately representing the mechanical effect of muscles with large attachment sites (Van der Helm & Veenbaas, 1991). The total number of muscle elements is 95. Because of the enormous amount of data, some choices are necessary to present the results. Firstly, the force of a subset of muscle elements is compared with mean EMG-signals, recorded at twelve well-defined muscle spots. Secondly, the exerted force for all 20 muscles and muscle parts is analysed during humeral elevation. In the third place, exerted moments are analysed for all muscles and other structures. Finally, the joint reaction forces and scapulothoracic reaction forces are presented.

III.2.1 Force-EMG comparison

Surface EMG was recorded for twelve muscles during ABU, ANU, ABL and ANL. Twelve subjects participated in the experiment, eight of them also participated in the motion recording experiment. Figs 7A to 7L show the rectified and averaged EMG patterns, normalized to the maximal EMG recorded, compared with the muscle force calculation of the model. The comparison is severely hampered because the amplitude of surface EMG is affected by muscle length (Heckathorne & Childress, 1981), resulting in a large amplitude near full elevation angles, whereas
Figure 6: Muscle length (in cm) during humeral abduction unloaded. Numbers indicate the muscle lines of action (elements) of which the order is presented in Table 1.
Figure 7: Muscle force vs. EMG for twelve muscles.

---: abduction unloaded (ABU); ---: anteflexion unloaded (ANU); --.--.: abduction loaded (ABL); ........: anteflexion loaded (ANL).
Figure 7 (continued)
the force is decreasing to zero. Therefore, surface EMG can only be used qualitatively to verify
the model, particularly the on/off patterns during humeral elevation should correspond with the
force calculations (Van der Helm, 1991a; Mouton et al., 1991). Only EMG amplitudes at the same
muscle length can be compared. Since small external loads did not change the scapular position
(Van der Helm & Pronk, 1991a), loaded and unloaded conditions can be compared (ABU vs. ABL,
ANU vs. ANL). Values under 10% of maximal EMG are expected to be subject to noise and
cross-talk, and thus should be considered as silent. Then, m. deltoideus, p. posterior, m. pectoralis
major, p. clavicularis, and m. biceps have different EMG and force patterns, all other muscle
forces do correspond with the recorded EMG. Since surface EMG is hampered by methodological
and interpretation problems, it does not provide a reliable prediction of muscle forces. It is our
opinion that the predictions of the model are more realistic and provide a better understanding of
the mechanics of the shoulder mechanism than EMG-recordings.

III.2.2 Muscle forces

Figs 8A to 8O present the forces generated by 15 muscles during ABU, ABL, ANU and ANL,
calculated by minimizing the sum of squared muscle stresses. Five other muscles of the shoulder
mechanism (m. pectoralis minor, m. teres minor, m. teres major, m. latissimus dorsi and m.
triceps) are hardly active (less than 1N force) during these motions, and are not shown. In the
quasi-static analysis external forces and moments, caused by the gravitational forces, are in
equilibrium with muscle forces and exerted moments.

M. trapezius, scapular part (Fig. 8A)

M. trapezius, pars descendens (roughly elements 5 and 6) is active, especially in the first 60
degrees abduction. It has the largest moment arm around the sagittal axis through the SC-joint.
Main external moment during abduction is around this axis. The lower part of m. trapezius even
has a negative moment arm around this axis throughout the motion. Though, around 90 degrees
abduction all parts become active in order to counteract the protracting force of m. serratus
anterior. During anteflexion pars transversalis (element 4) is the least active part. Main external
moment during anteflexion is around the frontal axis. Pars transversalis is more or less parallel to
the frontal axis and can not contribute to the moment around this axis. Pars descendens is active
since it has a useful moment around the sagittal axis.
Figure 8: Muscle forces (N) for twenty muscles and muscle parts. Numbers indicate the muscle lines of action of which the order is presented in Table 2. first row: abduction unloaded (ABU); second row: anteflexion unloaded (ANU); third row: abduction loaded (ABL); fourth row: anteflexion loaded (ANL).
Figure 8: Muscle forces (N) for twenty muscles and muscle parts. Numbers indicate the muscle lines of action of which the order is presented in Table 2. first row: abduction unloaded (ABU); second row: anteflexion unloaded (ANU); third row: abduction loaded (ABL); fourth row: anteflexion loaded (ANL).
Figure 8: Muscle forces (N) for twenty muscles and muscle parts. Numbers indicate the muscle lines of action of which the order is presented in Table 2. first row: abduction unloaded (ABU); second row: anteflexion unloaded (ANU); third row: abduction loaded (ABL); fourth row: anteflexion loaded (ANL).
M. trapezius, clavicular part (Fig. 8B)
The largest activity of the clavicular part of m. trapezius is at 30 degrees during abduction and at 90 degrees during anteflexion. Activity is almost similar to element 6 of the scapular part. The fact that the scapular part is a bi-articular muscle and the clavicular part is mono-articular seems not to affect muscular activity.

M. levator scapulae (Fig. 8C)
M. levator scapulae is active up to 150 degrees abduction and only up to 30 degrees anteflexion. During anteflexion activity of the levator scapulae would add to the external moment around the frontal axis. Moments exerted by m. levator scapulae are small.

M. pectoralis minor
M. pectoralis minor is not active during humeral abduction and anteflexion.

M. rhomboideus (Fig. 8D)
During abduction especially the upper part (element 3) of m. rhomboideus (sometimes described as a separate muscle: M. rhomboideus minor) is active. Its activity is merely counterbalancing the activity of the upper part of m. serratus anterior, and together these muscles succeed to press the Trigonum Spinae (TS) to the thorax. During anteflexion m. rhomboideus is not active, because TS is pressed to the thorax by the external moment.

M. serratus anterior (Fig. 8E)
M. serratus anterior is the most important muscle for counterbalancing the external moment around the SC- and AC-joint, during abduction as well as anteflexion. The PCSA of m. serratus anterior is about half the size of m. deltoideus which is the main muscle around the GH-joint. Through the shape of the scapula as a flat triangular bone, the moment arm of the lower part of m. serratus anterior (elements 1 to 4), inserting at the Angulus Inferior, is one of the largest in the human body. Therefore, m. serratus anterior is able to produce even larger moments around the SC-joint than m. deltoideus around the GH-joint. The upper part (elements 5 and 6) is active during abduction in order to keep TS on the thorax. During anteflexion this activity is not necessary, and therefore the upper part is not active.
M. deltoideus, scapular part (Fig. 8F)
M. deltoideus has the largest PCSA of the muscles of the shoulder mechanism, and exerts by far the largest moments around the GH-joint. During abduction pars medialis (elements 5 and 6) is the most active part. Pars posterior (elements 1 and 2) is not at all active, whereas the part in between is becoming less active with higher abduction angles. Through the combined lateral rotation and tipping backwards of the scapula, together with axial rotation of the humerus, the moment arm of pars posterior becomes smaller. Above 90 degrees anteflexion this part gets a larger moment arm and becomes active. Due to the increasing activity of pars posterior, pars medialis has its largest activity during anteflexion at 60 degrees.

M. deltoideus, clavicular part (Fig. 8G)
The bi-articular part of m. deltoideus, more or less pars anterior, has the same pattern of activity as pars medialis of the scapular part. Peak levels of activity are at 90 degrees abduction and 60 degrees anteflexion.

M. coracobrachialis (Fig. 8H)
During abduction m. coracobrachialis becomes active when its insertion at the humerus rises above the origin at the processus coracoideus, at 90 degrees. During anteflexion m. coracobrachialis has a positive moment arm until the insertion rises above the origin, at 90 degrees. Altogether, m. coracobrachialis is a small muscle, so the exerted forces and moments are small as well.

M. infraspinatus (Fig. 8I)
Up to 60 degrees abduction m. infraspinatus has a small useful moment arm around the sagittal axis. At higher abduction angles the moment arm becomes negative and the muscle is inactive. During anteflexion this muscle has a useful moment arm and is able to counterbalance the external moment. In addition, moments exerted by m. pectoralis major and m. deltoideus, clavicular part, around the vertical axis are counterbalanced.

M. teres minor
M. teres minor is not active during humeral abduction and anteflexion.

M. teres major
M. teres major is not active during humeral abduction and anteflexion.
M. supraspinatus (Fig. 8J)
Lying cranially and dorsally to the GH-joint this muscle is active during both abduction and anteflexion. Because m. supraspinatus is a small muscle with a small moment arm, exerted forces and moments are small.

M. subscapularis (Fig. 8K)
Because m. subscapularis is enclosed between the scapula and thorax and is therefore hardly palpable, it is one of the most underestimated muscles of the shoulder mechanism. Due to the scapular motion towards the sagittal plane and the axial rotation of the humerus, resulting in an insertion of m. subscapularis cranial to the GH-joint, m. subscapularis obtains a useful moment arm to counterbalance the external moment during abduction. Because of the large PCSA of this muscle, it can be classified as the second important abductor muscle around the GH-joint. During anteflexion the muscle is barely active.

M. biceps, caput longum (Fig. 8L)
The tendon of the long head of m. biceps runs through the sulcus bicipitalis and is attached to the upper rim of the glenoid cavity. Through humeral axial rotation the tendon is always cranial to the GH-joint and the muscle is active throughout humeral abduction, especially when the arm is loaded. As from 90 degrees anteflexion the tendon shifts to a position medial to the rotation center of the GH-joint and the muscle is hardly active any more.

M. biceps, caput breve (Fig. 8M)
The short head of m. biceps has a combined origin with m. coracobrachialis, and the distinction between these muscles is hardly palpable. Activity of the short head of m. biceps is reflecting the activity of m. coracobrachialis. The muscle is active at the first 60 degrees anteflexion and starting from 60 degrees abduction.

M. triceps, long head
M. triceps is not active during humeral abduction and anteflexion.

M. latissimus dorsi
M. latissimus dorsi is not active during humeral abduction and anteflexion.
M. pectoralis major, thoracic part (Fig. 8N)
Starting at 90 degrees abduction the upper thoracic part of m. pectoralis major (element 5), originating close to the SC-joint, gets a larger moment arm. At higher abduction levels a larger part of the muscle becomes active. Up to 60 degrees anteflexion only the upper part (element 5) has a useful moment arm, and hence it is active. The lower part of this muscle is not active during humeral abduction and anteflexion.

M. pectoralis major, clavicular part (Fig. 8O)
The clavicular part of m. pectoralis major shows the same activity pattern as the upper thoracic part. Starting at 90 degrees abduction it becomes active. During humeral anteflexion it remains active throughout the motion, but at higher anteflexion angles the activity mainly counterbalances activity of the scapular part of m. deltoideus.

III.2.3 Moments
The bearing of the medial border of the scapula to the thorax converts the shoulder girdle into a closed-chain mechanism. This motion constraint results in compulsory rotations in the SC- and AC-joint. Though, for analysis of exerted forces and moments the shoulder mechanism can be regarded as an open chain of three links, clavicle, scapula and humerus, connected by three joints, SC-, AC- and GH-joint, to the rigid world, i.e. the thorax. For each muscle exerted moments have been calculated using the moment arm and exerted force which result from the model simulations, and summed for all elements constituting the mechanical effect of the muscle. Analogously the moment exerted by ligaments can be calculated. Moments exerted by the forces between thorax and scapula have been calculated as well, since the point of application, direction and magnitude of forces in the SURFACE elements are known.

Since input variables were corrected for thorax rotations, moments are calculated around the main axes of a global coordinate system fixed to the thorax. Consequently, gravitational forces are not exactly vertically directed, but can even have a (small) moment around the global Y-axis. Figs 9, 10 and 11 show the moments around the three axes of the SC-, AC- and GH-joints for unloaded abduction (ABU) and unloaded anteflexion (ANU). The most important difference between ABU and ANU is obviously that during ABU the main external moment, due to the weight of the arm, is around the Z-axis (sagittal axis), whereas during ANU the main external moment is around the X-axis (frontal axis).

The moment analysis presented in Figs 9, 10 and 11 is quite different from moment analysis
commonly presented in literature. Usually, moments are presented with respect to a defined rotation axis, e.g. flexion/extension axis or adduction/abduction axis, which is fixed to the local coordinate system of the proximal bone. In this study moments are presented with respect to axes of the global coordinate system. In this way the changing contribution of a muscle to counterbalance other moments can be attributed to a position change of its attachments. If the moments were presented with respect to the proximal bone, it would be uncertain for poly-articular muscles if a change in moment would be due to either a position change of the intermediate bone, or to a position change of the attachments of the muscles. In addition, because of the 3D motions of clavicle and scapula, the position of the rotation axes would be difficult to conceive.

**SC-joint, moment around the X-axis**

During abduction the weight of the arm exerts a more or less constant moment around this axis. The main moment is due to m. serratus anterior, counterbalanced by the moment exerted by the thorax at the angulus inferior. The moment of m. trapezius, scapular part, has a peak at 90 degrees abduction, because especially the lower part is most active at 90 degrees (Fig. 9A).

During anteflexion, the main external moment is around the X-axis. M. serratus anterior is throughout the anteflexion the most important muscle to counterbalance the external moment, with some support of the lower part of m. trapezius. Reaction forces at the scapulothoracic gliding plane are adding to the external moments (Fig. 9D).

**SC-joint, moment around the Y-axis**

Since the external moment around the Y-axis is virtually zero, moments around this axis are due to the activity of muscles, exerting moments around other axes, and muscles compensating for these moments. For the SC-joint, the moment of m. serratus anterior, and in the case of humeral abduction combined with the moment at AI, is counterbalanced by m. trapezius. The latter seems to be the main function of m. trapezius (Figs 9B and 9E).

**SC-joint, moment around the Z-axis**

During abduction, the moment around the Z-axis is the largest external moment. It is surprising that the main moment counterbalancing the external moment is exerted by the reaction force between thorax and AI. The scapula is laterally rotated and protracted until AI is lateral to the thorax. External moments at the humerus, transmitted by m. deltoideus to the scapula, seems to be easily counterbalanced, because the scapula is in a very stable position. M. serratus anterior is
Figure 9: Moments (in Nm) around the X-, Y- and Z-axis of the sternoclavicular joint for abduction unloaded (A, B and C, resp.) and anteflexion unloaded (D, E and F, resp.). Moments below 1 Nm have been omitted.

less active during humeral abduction than during humeral anteflexion (Fig. 9C).
During anteflexion, a similar mechanism is acting. Though, due to the much smaller external
moment, the moment through AI is smaller and m. serratus anterior exerts relatively a larger
moment (Fig. 9F).

AC-joint
No mono-articular muscle is crossing the AC-joint, so moments around this joint are side-effects
of the muscle activity needed around the SC- and GH-joint. Some moments, especially of m.
deltoides, clavicular part, and m. trapezius, clavicular part, are compensated by the extracapsular
conoid ligament, preventing the clavicula from rotating forwards around its length axis.

AC-joint, moment around the X-axis
During humeral abduction, m. serratus anterior and forces at AI are counterbalancing each other.
Moments are larger than at the SC-joint, because the AC-joint is more cranial than the SC-joint,
so the moment arms are larger. Some additional moments are due to m. trapezius, scapular part,
and the conoid ligament (Fig. 10A).
During anteflexion, m. serratus anterior is counterbalancing the external moments. Scapulothoracic
reaction forces cause much smaller moments than during abduction (Fig. 10D).

AC-joint, moment around the Y-axis
A number of muscles which are crossing the AC-joint, causes a substantial moment around the Y-
axis. During abduction the reaction force at AI and m. trapezius, scapular part, cause the largest
counteracting moments (Fig. 10B). The moment due to the reaction force at AI is reduced to
almost zero above 90 degrees since AI is lying underneath the AC-joint. During anteflexion a
similar pattern is discovered, though the reaction force at TS also adds to the moment balance
(Fig. 10E).

AC-joint, moment around the Z-axis
During abduction, a similar pattern as at the SC-joint is encountered: m. serratus anterior and the
reaction force at AI on the one side, and external moments on the other. The conoid ligament has
a contribution on the other side as well (Fig. 10C).
During anteflexion, the moment due to the conoid ligament catches the eye, caused by the activity
of m. deltoideus, clavicular part, and m. trapezius, clavicular part. Other important moments are
Figure 10: Moments (Nm) around the X-, Y- and Z-axis of the acromioclavicular joint for abduction unloaded (A, B and C, resp.) and anteflexion unloaded (D, E and F, resp.). Moments below 1 Nm have been omitted.

due to m. serratus anterior and the reaction force at AI on the one side, and the external moment and m. trapezius on the other. It is remarkable that at higher elevation angles m. trapezius is not contributing to counterbalance the external moment, but that this muscle is even adding to this moment (Fig. 10F).

**GH-joint, moment around the X-axis**
During abduction, moments around the X-axis are small compared to the SC- and AC-joint. Striking is the cross-over from positive to negative moments and vice versa, when the 90 degrees abduction position is passed.
During anteflexion, the main external moment is around the X-axis. It is surprising that the only two muscles counterbalancing this external moment are m. deltoideus, scapular part, and m. deltoideus, clavicular part. Contributions of m. pectoralis major, m. subscapularis, m. supraspinatus, m. biceps, etc, are negligible (Fig. 11D).

**GH-joint, moment around the Y-axis**
Since the direction of gravitational forces is mainly parallel to the Y-axis, the moment balance is composed of muscles compensating for each others additional moments. During abduction, on the one side m. deltoideus, scapular part, is active and on the other side, m. subscapularis is the most important muscle (Fig. 11B). During anteflexion, both parts of m. deltoideus are active in opposite directions, with some additional activity of m. pectoralis major on the side of the clavicular part, and activity of m. infraspinatus and m. supraspinatus on the side of the scapular part (Fig. 11E).

**GH-joint, moment around the Z-axis**
During abduction, the main external moment is around the Z-axis. Up to 120 degrees the scapular part of m. deltoideus causes the largest counterbalancing moment, with a substantial contribution of m. deltoideus, clavicular part, and m. subscapularis. Remarkably, m. supraspinatus, being a generally recognized abductor, is hardly contributing to the moment around the Z-axis. During abduction, the scapula is laterally rotating, while the humerus is axially rotating. The insertion of m. subscapularis is moving superior to the GH-joint, while the insertion of m. supraspinatus moves posterior to the joint. In addition, m. supraspinatus has a small physiological cross-sectional area compared with m. subscapularis (Veeger et al., 19913). This suggests that, next to m. deltoideus, m. subscapularis is the second largest abductor around the GH-joint (Fig. 11C).
During anteflexion, the moments around the Z-axis are much smaller. The external moment, due
Figure 11: Moments (Nm) around the X-, Y- and Z-axis of the glenohumeral joint for abduction unloaded (A, B and C, resp.) and anteflexion unloaded (D, E and F, resp.). Moments below 1 Nm have been omitted.

to the imperfect standardization of the upper extremity to the sagittal plane, is counterbalanced by m. deltoideus (Fig. 11F).

**III.2.4 Joint reaction forces**

Since magnitude as well as direction of all forces acting on the bones of the shoulder mechanism are calculated by the model, the direction and magnitude of the joint reaction forces can be calculated. This joint reaction force is exerted from one bone onto the other by their articular surfaces pressing onto each other, and by the joint capsule, strengthened by capsular ligaments. Theoretically, the role of extracapsular ligaments can be distinguished from capsular ligaments, since their attachments and hence their length, could be recorded (Veeger et al., 1991; Van der Helm et al., 1991). However in practice, because the stress/strain relationship and the rest length of the ligaments is not known, inclusion of force transmitting and motion constraining ligaments in the model would lead to high muscle forces and odd motion patterns. Thus, only the length of the costoclavicular and trapezoid ligament was calculated. The conoid ligament has been modelled as a rigid TRUSS element, capable of transmitting forces. The role of this ligament in constraining the motion and distributing forces and moments around the AC-joint will be discussed.

**Glenohumeral joint**

Both articular surfaces of the GH-joint have been modelled as spheres with identical radii (Van der Helm et al., 1991). Then, the GH-joint behaves as a spherical joint with a rotation center fixed with respect to the scapula. Because of the large range of motion of the GH-joint, the joint capsule is lax and it is assumed not to transmit any forces within the normal physiological range of motion. Consequently, the vector resulting from muscle forces and external forces at the humerus, must intersect the articular surface of the glenoid cavity, otherwise it can not be counterbalanced by any joint reaction force and the joint will dislocate (Van der Helm et al., 1989). Pointing the resultant force vector to the glenoid cavity was added as a constraint in the optimization procedure. Table 3 shows for ABU, ABL, ANU and ANL in which humeral elevation positions the constraint was working, indicating that additional muscle forces were needed to prevent the joint from dislocating. It was established in simulations that with increasing load especially during anteflexion the dislocating force is larger. Stabilizing the joint required antagonistic muscle forces. Especially, muscles of the rotator cuff were activated since these muscles have small moment arms, allowing high forces with relatively small (antagonistic) moments. The position of insertions of rotator cuff muscles as a half circle around the humeral head enables these muscles to point the joint reaction
Table 3: Plus-signs indicate in which humeral elevation positions additional muscle force was necessary to stabilize the glenohumeral joint.

<table>
<thead>
<tr>
<th>elevation angle</th>
<th>abduction unloaded</th>
<th>anteflexion unloaded</th>
<th>abduction loaded</th>
<th>anteflexion loaded</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
</tr>
<tr>
<td>30</td>
<td>+</td>
<td>-</td>
<td>+</td>
<td>-</td>
</tr>
<tr>
<td>60</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>+</td>
</tr>
<tr>
<td>90</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>+</td>
</tr>
<tr>
<td>120</td>
<td>-</td>
<td>-</td>
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<td>-</td>
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<tr>
<td>150</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>180</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

force vector in almost any direction. Therefore, it can be concluded that rotator cuff muscles are most likely the main stabilizing muscles of the GH-joint.

Fig. 12A shows the location (in pole coordinates $\phi$ and $\theta$, the angle with the frontal and transversal plane, respectively) where the resultant joint vector intersects the glenoid cavity. In the restposition (0 degrees humeral elevation) the humerus is merely 'hanging' on the inferior rim of the glenoid cavity, requiring small additional muscle force to be pressed against the glenoid cavity. At higher elevation angles, the muscle force vectors required to counterbalance the external moment are more perpendicular to the glenoid cavity. At lower elevation angles the intersection point is more cranially than at higher elevation angles. During abduction the intersection point is at the anterior side of the glenoid cavity, whereas during anteflexion it is at the posterior site.

The direction and magnitude of the resultant muscle force vector are shown in Figs 12B to 12D with respect to the global coordinate system. Evidently, the largest reaction forces are at 90 degrees humeral elevation.

**Acromioclavicular joint**

Since only two bony landmarks could be distinguished at the clavícula, the axial rotation of the clavícula could not be assessed in the motion recording study (Van der Helm & Pronk, 1991). Axial rotation was estimated by minimizing rotations in the AC-joint. It was hypothesized that the tight extracapsular ligaments prevented motions of the AC-joint, and forced the clavícula to rotate around its length axis in order to follow the lateral rotation of the scapula. In this study the axial
rotation of the clavicle only depended on the rigid conoid ligament. The close resemblance between the simulated axial rotation and the estimated axial rotation supports the hypothesis (Fig. 5A and 5B).

Fig. 13 shows the direction and magnitude of the reaction force vectors of the AC-joint. Forces at the AC-joint are mainly influenced by three structures: The clavicular parts of m. deltoideus and m. trapezius, and the conoid ligament. Both muscles, but particularly m. deltoideus, cause a moment around the length axis of the clavicle. This moment is counterbalanced by the conoid ligament, preventing the clavicle from rotating just opposite to the actual axial rotation. Simulations in which the conoid ligament was precluded from force transmission revealed that the clavicular parts of m. deltoideus as well as m. trapezius became inactive, because no force was

**Figure 12:**

A: Intersection of the joint reaction force vector and the articular surface of the glenoid (in pole coordinates $\phi$ and $\theta$, the angle with the frontal and transversal plane, respectively).

--- Ellips fitted to the rim of the glenoid cavity.

$x$: Intersection points for abduction unloaded.

$o$: Intersection points for anteflexion unloaded.

B-D: Direction and magnitude of resultant force vector w.r.t. the global coordinate system (X-, Y- and Z-direction, respectively). For explanation lines see Fig. 13.
counteracting their activity. In reality, absence of the conoid ligament will cause the acromioclavicular capsule to generate the counteracting axial moment. However, this will lead to reduced axial rotation of the clavicle and hence more to rotation in the AC-joint. In combination with the increased torsional stress in the acromioclavicular capsule, consequently the joint might be damaged.

Without the conoid ligament, joint reaction forces are mainly directed along the length axis of the clavicle. Inclusion of the conoid ligament causes the joint reaction force to be directed downwards with respect to a fixed scapula. Urist (1946) and Moseley (1972) found in a majority of cases some overriding of the clavicle on the acromion, which is in accordance to the observed force direction. The lateral curvature of the clavicle is effective to provide the conoid ligament a larger moment arm. The curvature could probably come into being through the traction force at the conoid attachment and the compression forces along the length axis of the clavicle. This could be an explanation for the observation that the curvature is more pronounced with higher activity, for example at the dominant side (Fick, 1911; Martin & Saller, 1959).

Since the trapezoid ligament was excluded from force transmitting in these simulations, one can only speculate on its function. Pointing from the processus coracoideus towards the AC-joint, the moments exerted by this ligament are nil. During the present simulated motions, i.e. humeral elevation, its function is not strictly needed to complete the analysis. However, especially during adduction motions (and forces) medially directed forces from scapula to clavicle will be much higher. Because the trapezoid ligament is mainly in medial-lateral direction, it can neutralize

![Graphs showing force changes](image)

**Figure 13:**
Joint reaction force of the acromioclavicular joint in X-, Y- and Z-direction of the global coordinate system.

- : abduction unloaded (ABU).
- : anteflexion unloaded (ANU).
- : abduction loaded (ABL).
- : anteflexion loaded (ANL).
shearing forces in the AC-joint, especially if the clavicula overrides the acromion.

**Sternoclavicular joint**

Fig. 14 shows the reaction force vectors of the SC-joint. The reaction force is mostly directed along the length axis of the clavica. Around 90 degrees abduction, the reaction force in the SC-joint along the X-axis (frontal axis) is negative, indicating that the clavicula is pulled out of the joint. In this position a large external moment is present. If the Angulus Inferior which is positioned lateral to the thorax (Fig. 4), is considered as a rotation point, the external moment results in a pulling force at the clavicula (Fig. 15). M. subclavius is a small muscle crossing the SC-joint and is situated parallel to the clavica. It has a small moment arm and is probably active in keeping the clavicular head in the sternal cavity. However, during the morphological measurements the muscle was destroyed while exarticulating the SC-joint (Veeger et al., 1991), and hence the muscle is not incorporated in the model.

The costoclavicular ligament is an extracapsular ligament of the SC-joint. During abduction and anteflexion its simulated length change is enormous: from 1.0 to 3.0 cm. Since the rest length of the ligament is unknown, it could not be assessed when the ligament would start exerting forces and influence the motion of the SC-joint. Hence, the function of the ligament is still not understood.

During humeral abduction and anteflexion the clavica is subject to compression forces. The combined forces of m. serratus anterior and m. trapezius cause a lateral rotation of the scapula.

![Figure 14: Joint reaction force of the sternoclavicular joint in X-, Y- and Z-direction of the global coordinate system.](image)

---

- : abduction unloaded (ABU).
- : anteflexion unloaded (ANU).
- : abstraction loaded (ABL).
- : anteflexion loaded (ANL).
Without the clavicle the acromion would move medially. The clavicle offers an ideal point of support for m. serratus anterior which can use the scapula as a lever to counterbalance the large moment caused by the weight and external load of the upper extremity. Because m. trapezius exerts only a small moment around the AC-joint compared to m. serratus anterior, it is mainly active in positioning the point of support. In addition, the small thoracoscapular muscles like m. rhomboideus and m. levator scapulae are stabilizing the scapula to the thorax.

III.2.5 Reaction forces at the scapulothoracic gliding plane

Two SURFACE elements represent the connection between the medial border of the scapula and the thorax. In the optimization procedure stress in the SURFACE elements was constrained to compression. Figs 16A and 16B show the magnitude of the compression forces for ABU, ANU, ABL and ANL. Angulus inferior is always pressed to the thorax by m. serratus anterior. At higher anteflexion angles external forces exert a forward rotation moment at the scapula, pressing TS to the thorax. At lower anteflexion angles and at all abduction angles additional force of m. serratus anterior and m. rhomboideus is necessary to pull the medial border of the scapula to the thorax. The shoulder girdle can be regarded as a chain connecting the upper extremity to the thorax. If this had been an open chain of bones, around each joint a force and moment balance would be necessary to control the motion and stability of the joint. In order to obtain a stable and controllable chain, mono-articular muscles are preferred to position and fixate the bones around

![Diagram](image)

**Figure 15:** At 90 degrees humeral abduction the gravitational forces pull the clavicle out of the sternoclavicular joint. Traction forces at the SC-joint are necessary. Angulus inferior is considered as a rotation point.
a joint, since poly-articular muscle are affecting the motions of at least two joints. However, only two (small) mono-articular muscles, i.e. m. trapezius, clavicular part, and m. subclavius, are crossing the SC-joint, whereas no mono-articular muscle is crossing the AC-joint (Table 1). Thus, both joints would be unstable links in the chain. One bearing point of the scapula on the thorax, and preferably two, reduces the number of DOF and leads to a better controllable mechanism. Therefore, the assumption that two bearing points exist between thorax and scapula seems to be justifiable, because such an assumption would provide the most stable situation. The scapulothoracic constraint in the optimization procedure is a minimum condition for a stable shoulder girdle, in reality additional muscle force is expected to be necessary for further stabilization and control.

VI Discussion
The shoulder mechanism is a complex mechanism, combining the contradictory demands of a large range of motion and exerting large forces in any direction at the hand, with its large moment arm with respect to the shoulder. Many muscles are involved in positioning and stabilizing the scapula and humerus. For an analysis of muscle function it is important to include all muscles which are active during a motion, since these muscles are mutually affecting each other’s function.

![Graphs](image)

**Figure 16:** Reaction forces of the thorax at Trigonum spinae (A) and Angulus Inferior (B).

- ---: abduction unloaded (ABU).
- -----------: anteflexion unloaded (ANU).
- --------: abduction loaded (ABL).
- ........: anteflexion loaded (ANL).
addition, the bearing of the scapula at the thorax is very important for the motion behavior of the scapula and the muscles attached to it. The scapulothoracic gliding plane reduces the number of DOF of the shoulder girdle. Therefore, the moments exerted around three axes of the SC- and AC-joint are not independent of each other. If a muscle exerts a moment around one of these axes, the scapulothoracic reaction force will be affected and consequently moments around other axes also. Any realistic model of the shoulder mechanism should contain a representation of the scapulothoracic gliding plane.

Validation of such a complex model as the present model of the shoulder mechanism is very difficult (Van der Helm, 1991b). The use of the amplitude of surface EMG as a measure of muscle force is limited, since to a large extent surface EMG depends on muscle length. Therefore, it was concluded that only on/off patterns of force and EMG can be compared for verification of the model. However, in the motions analysed in this study hardly any on/off patterns are present. Comparison of EMG and force at different load levels at roughly the same muscle length (e.g. ABU vs. ABB) revealed that with additional load the same muscles become more active.

A strong and reproducible relation exists between the motions of the scapula and the humerus, the so-called scapulohumeral rhythm (Inman et al., 1944; Pronk, 1991; Van der Helm & Pronk, 1991a). Due to this rhythm one can not state that the motions of the shoulder girdle and the humerus are independent, as would be suggested by the number of independent mechanical DOF of the shoulder mechanism, but are coupled by a certain muscle control strategy. Therefore, it is necessary to use recordings of the related position of the scapula and humerus as input variables. The exact nature of the relationship remains unknown. Clearly, the scapular position depends on the position of the humerus, presumably depends on the external load on the humerus and most likely depends on the velocity of the humeral motions.

In order to calculate muscle forces an optimization criterion is used which is assumed to correspond with the muscle control strategy used by the human body to distribute the effort over the available muscles. It is most likely that the same kind of criterion is used to search for an optimal position of the scapula. In order to minimize all kinds of energy consumption criteria, one important factor will be to shift the scapula to a position in which muscle moment arms are optimized. For example, in the present study it is suggested that during anteflexion the angulus inferior is positioned lateral to the thorax to provide an optimal moment arm for m. serratus anterior. But during abduction the main moment around the SC-joint is due to the force between the thorax and AI, thereby minimizing the necessary muscle effort. Other important factors in muscle control strategy are probably muscle fatigue or the ability to control the motion and to stabilize the scapula.
It is very likely that the muscle control strategy will be task dependent.
In this study minimization of the sum of squared muscle stresses is used as optimization criterion.
To what extent this criterion reflects the actual muscle control strategy is yet unknown. A large
number of optimization criteria has been proposed in literature, but neither of them could be
validated until now. Van der Helm (1991a) showed that for a number of non-linear criteria the
resulting muscle forces were comparable. Though muscle force predictions largely depend on the
model assumptions, it is our opinion that muscle function is reasonably assessed, since the most
important factors are included in the model: A good approximation of muscle moment arms and
incorporation of all mechanically important morphological structures, especially the scapulothoracic
gliding plane.
In literature muscle function is assessed using a kinematic approach (Fick, 1911; Benninghof &
Goerttler, 1964). In this study dynamics are added and muscle function is analysed not only in the
anatomical position. Hence, new light is shed on some muscle functions. Establishing moment
balances around joints revealed which muscles exerted the largest moments counterbalancing
external moments. For example, it is shown that, next to m. deltoideus, m. subscapularis is likely
to be the second important abductor around the GH-joint instead of m. supraspinatus. If during a
motion the direction of external force and moment vectors differs from the direction of translational
and rotational position changes, the kinematic approach can not provide a good prediction of
muscle function. Especially poly-articular muscles of which the function depends on the motion
in two or more joints are affected. In addition, muscle groups can cooperate in order to obtain a
couple on the scapula, or, like m. trapezius, provide a clavicular motion profitably for m. serratus
anterior in order to obtain a point of support. Other muscles have a more stabilizing function, like
m. rhomboideus in co-operation with the upper part of m. serratus anterior which keep the scapula
connected to the thorax. Rotator cuff muscles sometimes exert antagonistic forces in order to
prevent the humeral head from dislocating.
In analyses of the lower extremity only a few motions are studied of which walking is the motion
most studied. The number of activities which can be analysed for the upper extremity is almost
infinite. In this study only a small subset of possible humeral motions has been analysed: Elevation
of the humerus in the frontal and sagittal plane. Five muscles did not show any activity during
these motions, but will likely be active in other interesting motions like adduction, retroflexion and
endo/exorotation. Furthermore, in daily activities people work most often with the hands in front
of their body while exerting forces in all kind of directions. In this study external forces were
restricted to gravitational forces. If the elbow is flexed, the same gravitational force results in a
different moment while the humerus is in the same position. A different combination of external force and moment at the same humeral position is likely to be accompanied by a change of muscle activity, and probably by a change of scapular position. What to say about all other possible force directions?

If another position of the upper extremity or another force direction is analysed, the analysis should start with recording the resulting scapular position. The scapular position depends on the muscle control strategy which is yet unknown. A similar procedure should be followed if for instance the consequences of paralysis of some muscles are studied. It is very likely that the scapular position will change. In this study only static positions of the shoulder mechanism have been analysed. As soon as it will be possible to measure the dynamics, a whole new dimension will be added to the research of the shoulder mechanism.

Concluding remarks
1. A dynamic musculoskeletal model incorporating all relevant morphological structures has been developed. The model provides good insight in the mechanics of the shoulder mechanism and enables an analysis of the function of morphological structures. Static humeral abduction and anteflexion, unloaded as well as loaded have been analysed.

2. Since motions of the shoulder girdle and humerus are strongly related, simultaneous motion of these bones should be used as input variables in an inverse dynamic analysis.

3. Since the EMG-force relationship is unknown, amplitude of surface EMG could not be used to validate muscle force predictions.

4. The scapulothoracic gliding plane is very important for the motions and the stabilization of the shoulder girdle.

5. The clavica offers a point of support for m. serratus anterior which can use the scapula as a lever to exert large moments around the SC- and AC-joints. M. trapzius is mainly active in positioning the point of support.

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Chapter 7

Three-dimensional recording and description of motions of the shoulder mechanism.

F.C.T. van der Helm, G.M. Pronk,


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Abstract

A measurement technique is presented for recording positions of the bones of the shoulder mechanism, i.e. thorax, clavicle, scapula and humerus, in 3D space, based on palpating and recording positions of bony landmarks. The palpation technique implies that only static positions can be measured. Accuracy of retrieving bony landmarks is checked on-line using rigid body assumptions. Resulting measurement error is calculated afterwards and is comparable with cinegraphic methods. Axial rotation of the clavicle is estimated by minimizing rotations in the acromioclavicular joint. A number of motion definitions is compared by means of interindividual variation and subjective interpretability. Two useful definitions are proposed for describing motions of the shoulder mechanism. Four conditions have been recorded: abduction and anteflexion of the humerus with and without additional weight in the hand. Abduction and anteflexion result in large differences in scapular and clavicular motions. The effect of additional weight in the hand on the position of the shoulder girdle is negligible.
I INTRODUCTION
The shoulder mechanism is a constellation of bones, i.e. thorax, clavicula, scapula and humerus, which connect the trunk and upper extremity. Bones of the shoulder girdle, i.e scapula and clavicula, provide a moveable but stable base for the large range of motions of the humerus. Motions of the humerus are the result of simultaneous motions at the sternoclavicular (SC-)joint between the thorax and clavicula, at the acromioclavicular (AC-)joint between clavicula and scapula and at the glenohumeral (GH-)joint between scapula and humerus. In addition, to reach the utmost humeral elevation angle thorax rotations are necessary, either lateral rotations in one-sided elevation or backward rotations in two-sided elevation (Stenvers & Overbeek, 1981; Pronk, 1991). Scapular motions are constrained on the one hand by the clavicula, restricting the acromion to move on a sphere around the SC-joint with the clavicula as radius, and on the other hand by the scapulothoracic gliding plane, where the medial border of the scapula is pressed against the thorax by the activity of m. rhomboideus and m. serratus anterior (Van der Helm, 1991b). Since the connection between scapula and thorax restricts two degrees of freedom, the shoulder girdle can be regarded as a closed-chain mechanism. Considering the complex motion constraints of the scapula only a full three-dimensional (3D) motion recording and description will yield a good understanding of its motion behavior. Thus far, to our knowledge, no satisfying method for 3D recording of motions of the shoulder mechanism has been published. Consequently, no one ever elaborated the special problems of 3D motion description.

1.1 Recording methods
The most apparent problem in motion recording of the shoulder mechanism is that the scapula is moving underneath the skin. Hence, commonly used recording methods as film and video recording of markers glued to the skin are not applicable. In literature, a number of recording methods for scapular motions can be distinguished to quantify the in vivo motions of the scapula and clavicula (Pronk, 1991):

1) Roentgenographic studies in which the positions of the scapula and clavicula have been projected on the roentgen plate at several angles of humeral abduction or anteflexion (Steinhaussen, 1899; Inman et al., 1944; Meijers, 1961; Saha, 1961; Freedman & Munro, 1966; Poppen and Walker, 1976; Dvir & Berme, 1978; Wallace, 1982; Wallace & Johnson, 1982; Peterson et al., 1985).
2) Cinematographic studies in which the shadow contour of the lateral border have been used to reconstruct the position of the scapula (Hvorslev, 1927).
3) Photographic studies in which at several angles of humeral elevation the positions of scapula and clavicle have been recorded by marking a number of specific bony landmarks on both bones (Bagg & Forrest, 1988).

4) Studies in which goniometers have been used to record specific angles between clavicle and scapula (Conway, 1961; Doody et al., 1970a; 1970b).

5) Studies in which pins have been inserted into clavicle and scapula in order to measure externally the internal motion of the bones (Inman et al., 1944; Inman & Saunders, 1946; Kennedy & Cameron, 1954; Fontijne, 1965; Rockwood, 1975).

From the nature of most studies motion description has been restricted to a few angles between the bones or to a two-dimensional projection like in roentgenographic studies using one camera. 2D motion description of a clearly 3D motion is hazardous. If a bone is angular with the projection plane, projection errors are inevitable. As shown in Fig. 1A, rotation of the clavicle around the vertical axis results in an imaginary increase in elevation angle if projected on the frontal plane. Hence, the recorded elevation angle is a combination of real elevation and projection errors (Inman et al., 1944), see Fig. 1B. It is concluded that 2D motion recordings are not suitable for description of joint rotations. They will yield a very limited comparison of subjects and will never reveal the real 3D motions and motion constraints of the shoulder mechanism.

![Figure 1A: Effect of dorsal rotation of the clavicle around the vertical axis on the projected elevation angle.](image1a.png)

![Figure 1B: Results of roentgen measurements of Inman et al. (1944). The elevation angle is the combined result of real elevation and projection errors.](image1b.png)
Studies of the 3D motion of the shoulder mechanism are scarce. Wallace (1982) and Wallace & Johnson (1982) used two roentgen cameras to record the motion of one subject. The angles between bony ridges and anatomical planes were used to describe the motions. Unfortunately, these angles do not permit a 3D reconstruction of the scapular position. At least three non-collinear points of each bone are required to obtain the 3D position. An attempt to use bony landmarks as markers failed due to the impossibility to retrieve the same landmarks with accuracy in two perpendicular projections (Casolo et al., 1987). Peterson et al. (1985) used artificial markers for in vivo recordings of the shoulder bones. However, they did not present a detailed description of their methodology and results.

1.2 Motion description

Besides problems in recording 3D motions of the shoulder mechanism it is difficult to present an unambiguous and interpretable description of the motion. Such a description is necessary to compare shoulder motions of healthy subjects to those of patients in a way that differences can be used to gain insight in the pathology.

In medical practice, reflected in anatomical textbooks, motions are described with respect to the anatomical position as a reference position. Rotations are defined around the main anatomical axes which constitute a kind of global coordinate system. For instance, humeral anteflexion, exorotation and abduction are defined as starting from the position in which the arm is hanging down at the side of the body, around the frontal, vertical and sagittal axis, respectively. In this way, the motion in the transversal plane from 90 degrees anteflexion to 90 degrees abduction is not defined. In addition, for the bones of the shoulder girdle no anatomical position has been defined at all. As shown in Fig. 2, motion of these bones are mainly described by translations of certain bony landmarks and not by rotations around well-defined axes. For example, lateral rotation of the scapula is the motion in which angulus inferior moves to lateral. Combination of two motions is not quantifiable. Using medical definitions as a starting point for motion description of the shoulder will lead to ambiguous results. However, from a clinical, more qualitative, viewpoint these motions may be reasonably interpretable.

Using technical definitions, the movement of a bone from one position to another can be described unambiguously by a 3x3 rotation matrix and a translation vector which are usually defined with respect to the coordinate system of the proximal bone. Usually, in vivo only rotations are described and translations in the joints are neglected. Interpretation of the rotation matrix is difficult. Among the nine numbers of the rotation matrix only three independent variables are hidden. Two ways can
be discerned to separate the independent variables out of the rotation matrix: Euler angles and the
discrete helical axis or screw axis.

Using Euler angles, rotation matrix $R$ is decomposed in three elementary rotation matrices around
the axes of a predefined coordinate system, e.g.:

$$ R = R_x(\alpha) \cdot R_y(\beta) \cdot R_z(\gamma) $$

\[
\begin{bmatrix}
1 & 0 & 0 \\
0 & \cos(\alpha) & -\sin(\alpha) \\
0 & \sin(\alpha) & \cos(\alpha)
\end{bmatrix} \cdot
\begin{bmatrix}
\cos(\beta) & 0 & \sin(\beta) \\
0 & 1 & 0 \\
-\sin(\beta) & 0 & \cos(\beta)
\end{bmatrix} \cdot
\begin{bmatrix}
\cos(\gamma) & -\sin(\gamma) & 0 \\
\sin(\gamma) & \cos(\gamma) & 0 \\
0 & 0 & 1
\end{bmatrix},
\]

from which the angles $\alpha$, $\beta$ and $\gamma$ can be calculated. These angles can be interpreted as successive
rotations of the bone around the global X-axis, then around the rotated Y'-axis and finally around
the rotated Z''-axis of the coordinate system (Paul, 1983). Single (') and double (") quotes refer
to transformation of the rotation axis by one or two prior rotations, respectively. The order of the
elementary matrices is important because changing the order will result in other values for the
angles. This implies that if the upper arm is rotated 180 degrees around the frontal axis
(anteflexion) and then 180 degrees around the sagittal axis (adduction), another position is reached
than if the arm is moved 180 degrees around the sagittal axis (abduction) and 180 degrees around

![Diagram of scapular motions](image)

**Figure 2:** Medical definitions of scapular motions. Motions are mainly defined by translations of bony
landmarks, not around well-defined axes.
the frontal axis (reversed anteflexion). This is a trivial solution for the Codman paradox (Codman, 1934). Besides the order problem, interpretation of the rotation angles is also hampered by the fact that the directions of the rotation axes, and hence the rotation angles, are affected by each other. If $\beta$ is changed in Eq. (1), the direction of the rotated $Z''$-axis is changed, thus the effect of $\gamma$ and the resultant position of the bone. A certain angle $\gamma$ can only be valued considering $\alpha$ and $\beta$. This is called the interaction problem.

Using a discrete helical axis or screw axis, rotation and translation of the bone can be imagined as a single rotation around a certain axis (the helical axis) and a translation along the axis. Independent variables describing the bony rotation are the direction of the axis (only two variables, since the direction vector is defined to be an unit vector) and the rotation angle around the axis. The order problem, as encountered using Euler angles, is solved, but the interaction problem still remains: the rotation angle can only be valued considering the orientation of the axis and vice versa. Yet another problem appears if the difference in direction of two screw axes describing subsequent rotations should be interpreted. The discrete helical axis approach only seems useful if the rotations are very small and the discrete helical axis approaches the instant helical axis of which the excursion during the motion can be visualized. Even then, useful interpretation of this excursion is only possible if the helical axes are parallel or intersect at one point: the pivot of the motion (Lange, 1987). Because in this study position changes are large, the discrete helical axis is not approximating the instantaneous helical axis. In addition, since the recordings are not accurate enough to measure joint translations, the position vector of the helical axis and the translation along the axis could not be calculated. Therefore, the helical axis method is not elaborated.

Thus far, no unambiguous technical description of the bony motions of the shoulder mechanism has been published. For the shoulder mechanism a particular problem in motion definition exists because the anatomical position of scapula and clavicula is not defined. The rest position of the shoulder girdle which is defined as the position when the arm is hanging aside the trunk, differs from person to person. If during humeral elevation another position of e.g. the scapula is recorded, rotations can be calculated with respect to the rest position, different for each individual, or with respect to a predefined reference position, equal for everybody. Advantage of the latter method is that if two persons have the same scapular position, the rotations of the bone will be identical. However, starting from a different rest position the scapular motion can be identical for two persons. Using the first method, this will result in the same rotations since the different starting position is taken into account. Is it more important how much the scapula has been rotated at a
certain humeral elevation angle (relative motion), or is the position of the scapula more important (absolute motion)?

Purpose of this study is to record the 3D positions of the bones of the shoulder mechanism, and subsequently to find a suitable way to describe the motions of the bone which is unambiguously and reasonably interpretable. Positions of the shoulder mechanism have been recorded using a palpation technique (Pronk, 1987). In this technique bony landmarks have been palpated and next 3D coordinates have been recorded using the palpator, a specially developed measurement instrument for digitizing spatial coordinates (Pronk & Van der Helm, 1991). Hence, the position of the bones could be reconstructed. Recordings have been made during unloaded and loaded humeral abduction and anteflexion. In the data processing special attention is paid to the pros and cons of several motion definitions using Euler angles. Finally, two definitions are proposed for describing motions of the shoulder mechanism.

II METHOD

II.1 Subjects

The position of the shoulder mechanism bones has been recorded in ten healthy, male subjects without any prior shoulder complaints. In Table 1 some general and morphological characteristics of the subjects are presented.

II.2 Bony landmarks

For each bony structure at least three non-collinear points are needed for a 3D reconstruction of its position and orientation. It is important that for each subsequent position exactly the same bony landmarks are recorded. At the clavicle only two useful bony landmarks could be discerned. Therefore, additional assumptions are required to reconstruct the 3D orientation (see Section III: Data processing). A manchet has been fixed to the medial and lateral epicondyle of the humerus to provide for three discernable markers. Table 2 shows the bony landmarks which are selected for use in the measurements.

Incisura Jugularis (IJ) has been defined as origin of the global coordinate system. It is strictly taken not a bony landmark of the clavicle. However, addition of a bony landmark at the SC-joint did not change the clavicular rotations very much, so in order to reduce the number of bony landmarks IJ is considered as the most medial point of the clavicle.
Table 1: Subject characteristics.

<table>
<thead>
<tr>
<th></th>
<th>M1</th>
<th>M2</th>
<th>M3</th>
<th>M4</th>
<th>M5</th>
<th>M6</th>
<th>M7</th>
<th>M8</th>
<th>M9</th>
<th>M10</th>
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<tbody>
<tr>
<td>length (cm)</td>
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<td>180</td>
<td>184</td>
<td>180</td>
<td>176</td>
<td>176</td>
<td>183</td>
<td>186</td>
<td>183</td>
<td>187</td>
</tr>
<tr>
<td>weight (kg)</td>
<td>66</td>
<td>73</td>
<td>73</td>
<td>65</td>
<td>77</td>
<td>74</td>
<td>74</td>
<td>68</td>
<td>70</td>
<td>73</td>
</tr>
<tr>
<td>age (years)</td>
<td>24</td>
<td>23</td>
<td>22</td>
<td>19</td>
<td>24</td>
<td>23</td>
<td>26</td>
<td>22</td>
<td>23</td>
<td>25</td>
</tr>
<tr>
<td>dominancy (right/left)</td>
<td>R</td>
<td>L</td>
<td>R</td>
<td>R</td>
<td>R</td>
<td>R</td>
<td>R</td>
<td>R</td>
<td>R</td>
<td>R</td>
</tr>
<tr>
<td>upperarm length (cm)</td>
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<td>31</td>
<td>33</td>
<td>30</td>
<td>33</td>
<td>32</td>
<td>32</td>
<td>34</td>
<td>32</td>
<td>34</td>
</tr>
<tr>
<td>arm length (cm)</td>
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<td>68</td>
<td>67</td>
<td>67</td>
<td>63</td>
<td>67</td>
<td>69</td>
<td>68</td>
<td>68</td>
</tr>
<tr>
<td>thorax width (cm)</td>
<td>32</td>
<td>31</td>
<td>32</td>
<td>30</td>
<td>32</td>
<td>33</td>
<td>31</td>
<td>30</td>
<td>29</td>
<td>29</td>
</tr>
<tr>
<td>thorax depth (cm)</td>
<td>18</td>
<td>20</td>
<td>21</td>
<td>20</td>
<td>19</td>
<td>20</td>
<td>21</td>
<td>18</td>
<td>20</td>
<td>21</td>
</tr>
</tbody>
</table>

**II.3 Palpator**

The palpator is an open chain mechanism consisting of four links and four hinge joints. Its base is fixed to the rigid world. Rotations of the hinge joints are recorded using high-precision potentiometers (< 0.1% non-linearity). Voltage of the potentiometers is AD-converted (12 bits AD-converter, Data Translation DT 2801) and used to calculate the position of the endpoint of the final link. The accuracy of the palpator was estimated to be 0.96 mm standard deviation per coordinate and 1.43 mm per reconstructed point (Pronk & Van der Helm, 1991).

The position of a bony landmark can be recorded by pressing the endpoint of the palpator (radius 1.5 mm) to the palpated landmark. A foot-button is used as trigger and the 3D coordinates of the landmark are calculated on-line.

**II.4 Protocol**

Recordings have been made at seven positions (approximately 0, 30 60, 90, 120, 150 and 180 degrees) during two-sided unloaded humeral abduction (ABU) and anteflexion (ANU), while the subject was standing and actively raising his arms. The order of positions has been randomized to exclude fatigue effects. In addition, recordings have been made with a load of 0.75 kg in the hand during abduction (ABL) and anteflexion (ANL), totalling 28 positions per subject to be measured. For each position 11 landmarks as mentioned above were recorded. Total recording time of one position was about 2 - 2.5 minutes.

At the start of the measurements the subject’s rest position which is a standing position with the
Three-dimensional recording and description of motions

Table 2: Bony landmarks as selected for determining the position of bones of the shoulder mechanism.

<table>
<thead>
<tr>
<th>Location</th>
<th>Landmarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax:</td>
<td>Incisura Jugularis (IJ)</td>
</tr>
<tr>
<td></td>
<td>transition from sternum to Processus Xiphoideus (PX)</td>
</tr>
<tr>
<td></td>
<td>7th cervical vertebra (C7)</td>
</tr>
<tr>
<td></td>
<td>8th thoracic vertebra (T8)</td>
</tr>
<tr>
<td>Clavícula:</td>
<td>Incisura Jugularis (IJ)</td>
</tr>
<tr>
<td></td>
<td>most dorsal point on the acromioclavicular joint (AC)</td>
</tr>
<tr>
<td>Scapula:</td>
<td>most dorsal point on the acromioclavicular joint (AC)</td>
</tr>
<tr>
<td></td>
<td>Angulus Acromialis (AA)</td>
</tr>
<tr>
<td></td>
<td>Trigonum Spinae (TS)</td>
</tr>
<tr>
<td></td>
<td>Angulus Inferior (AI)</td>
</tr>
<tr>
<td>Humerus:</td>
<td>Manchet fixed to the medial and lateral epicondyle</td>
</tr>
</tbody>
</table>

Arms hanging aside the trunk (0 degrees elevation), was measured five times. These recordings were used to calculate the mean distance between bony landmarks on the same bone, i.e. IJ-AC, AC-AA, AC-TS, AA-TS, TS-AI and AA-AI. The distance between these bony landmarks does not change during humeral elevation. Landmarks were measured in the order: IJ - AC - AA - TS - AI. The distance between a bony landmark and previously recorded bony landmarks was used to check on-line whether the correct bony landmark was recorded. Beforehand, 6 mm standard deviation was considered to be acceptable as measurement error of each distance between bony landmarks. A standard deviation of 6 mm results in a 95% reliability interval with a range of 23.44 mm, assuming a normal distribution. Hence, a recorded point was accepted if the distance with previously recorded bony landmarks was within 11.72 mm of the mean distance. Recording of a bony landmark was repeated until these constraints were satisfied. If the constraints could not be satisfied, indicating that one of the former points was fault, recording of the entire shoulder position was started all over.

During the recording of the bony landmarks in one position, no movements of the subject were allowed. Therefore, the subject’s position was well standardized. Former experiments (Pronk, 1991) had indicated that sitting subjects leant back and forth, so in this experiment subjects were standing, forcing them to keep their own balance. Swinging motions were restricted by fixating the head through a helmet and the hips through bars against both trochanter major (Oranje, 1989), see Fig. 3. Precautions were made that the subject stood in a relaxed position. Starting from this position, the subject could raise his arms without being impeded by the fixation frame. Thorax rotation was permitted since this is considered to be part of the motion during humeral elevation.
Recordings were made after expiration in order to prevent thorax deformations.

Elevation angle was set by asking the subject to hold his hand against a marker. Afterwards the exact elevation angle was calculated using the landmarks on the humeral manchet. The hand palm was directed medially during anteflexion and ventrally during abduction.

III DATAPROCESSING

Orientation of a bone with respect to the global coordinate system is defined by a so-called orientation matrix which is similar in form as a rotation matrix and defines the rotation from the global to the local coordinate system of the bone. Change in orientation of the bone is described by a 3x3 rotation matrix which can be defined with respect to the local or global coordinate system. Hence, decomposition of this rotation matrix into Euler angles will define rotations around axes of the local or global coordinate system, respectively. In addition, the orientation of the bone with respect to the proximal bone can be calculated. Change of this orientation is described by a rotation matrix indicating rotations in the joint connecting both bones.

Figure 3:
Standardization frame. Foot position, hips and head are fixed. Humeral position is standardized by markers which should be touched.
III.1 Local coordinate systems

Recorded data were transformed to a global coordinate system with its origin at the incisura jugularis (I), X-axis pointing laterally, Y-axis pointing cranially and the Z-axis pointing dorsally. Only right shoulders were measured.

The first step in data processing is defining the orientation of each bone by its local coordinate system, resp. T, C, S and H for the thorax, clavicle, scapula and humerus. These local coordinate systems are defined w.r.t. the global coordinate system as follows:

Thorax: \( y_t \)-axis: \((\mathbf{I} - \mathbf{P}_X) \parallel \mathbf{I} - \mathbf{P}_X \) ;
\( x_t \)-axis: perpendicular to the plane \( \mathbf{I} - \mathbf{P}_X - \mathbf{C}_Z \), pointing right;
\( z_t \)-axis: as defined by \( x_t \)- and \( y_t \)-axis.

Clavicle: \( x_c \)-axis: \((\mathbf{A}_C - \mathbf{I}) \parallel \mathbf{A}_C - \mathbf{I} \) ;
\( z_c \)-axis: perpendicular to the \( x_c \)-axis and the global Y-axis, pointing dorsally;
\( y_c \)-axis: as defined by \( x_c \)- and \( z_c \)-axis.

Scapula: \( x_s \)-axis: \((\mathbf{T}_S - \mathbf{A}_C) \parallel \mathbf{T}_S - \mathbf{A}_C \) ;
\( z_s \)-axis: perpendicular to the plane \( \mathbf{A}_C - \mathbf{T}_S - \mathbf{A}_I \), pointing dorsally;
\( y_s \)-axis: as defined by the \( x_s \)- and \( z_s \)-axis.

Humerus: \( y_h \)-axis: longitudinal axis of the manchet, pointing caudally in the rest position;
\( x_h \)-axis: pointing from the medial to the lateral epicondyle;
\( z_h \)-axis: as defined by the \( x_h \)- and \( y_h \)-axis.

where \( \mathbf{I} \) is the vector containing the position coordinates of bony landmarks \( \mathbf{I} \), etc. The local coordinate systems of the clavicle, scapula and humerus are shown in Fig. 4. Each local coordinate system is described by a 3x3 orientation matrix \([x \ y \ z] \) describing the orientation of a bone with respect to the global coordinate system. In the next paragraphs, index \( i \) refers to the elevation angle of the humerus, e.g. \( T_i \) (\( i = 0, 1, \ldots 6 \)) refers to the orientation of the thorax at 0, 30, 60, 90, 120, 150 and 180 degrees humeral elevation, respectively.

III.2 Global vs. local rotation matrices

The rotation of a bone from one orientation to another is indicated by a 3x3 rotation matrix \( \mathbf{R} \), e.g.:

\[
\mathbf{R} \cdot T_0 = T_1 .
\]
If rotation matrix $R$ is located before the orientation matrix, it is indicating rotations around axes of the global coordinate system. On the contrary, if rotation matrix $R$ is located behind the orientation matrix, it is indicating rotations around axes of the local coordinate system as described by the orientation matrix (Pronk, 1984), e.g.

$$T_0 \cdot R = T_1.$$  \hspace{1cm} (3)

This is a general property of describing subsequent rotations using rotation matrices. If in Eq. (2) the rotation matrix $R$ is split into elementary rotation matrices, e.g.:

$$R \cdot T_0 = R_x(\alpha) \cdot R_y(\beta) \cdot R_z(\gamma) \cdot T_0,$$  \hspace{1cm} (4)

the order of elementary matrices x-y-z can be interpreted in two ways: rotations around local axes which move due to prior rotations, when reading from left to right, or as rotations around global non-moving axes, when reading from right to left. In other words, Eq. (4) can be viewed as a rotation $\gamma$ around the global Z-axis, followed by a rotation $\beta$ around the global Y-axis and finally a rotation $\alpha$ around the global X-axis, and as a rotation $\alpha$ around the global (or local) X-axis, followed by a rotation $\beta$ around the local y'-axis defined by $R_x(\alpha)$, and finally by a rotation $\gamma$ around the local z''-axis defined by the prior rotations $R_x(\alpha) \cdot R_y(\beta)$. However, since $\alpha$, $\beta$ and $\gamma$

\begin{figure}[h]
\centering
\includegraphics[width=0.8\textwidth]{figure4.png}
\caption{Local coordinate systems of clavicle, scapula and humerus (dorsal view).}
\end{figure}
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are exactly the same, global rotations do not solve the order and interaction problem, as stated in the introduction. In this study rotations are interpreted as local rotations, and will be read from left to right.

III.3 Rotations of the bones

In the following sections rotation matrices will be notated by e.g. $R_{01}^T$, indicating a rotation of the thorax from position 0 (0 degrees humeral elevation) to position 1 (30 degrees humeral elevation). For similar rotations of the clavicle, scapula and humerus the rotation matrices will be denoted by $R_{01}^C$, $R_{01}^S$ and $R_{01}^H$, respectively. Then, the rotation of a bone from one position to another can be defined in two ways:

Def. 1a: Rotations with respect to the axes of the global coordinate system:

\[
\begin{align*}
R_{0i}^T \cdot T_0 &= T_i \\
R_{0i}^C \cdot C_0 &= C_i \\
R_{0i}^S \cdot S_0 &= S_i \\
R_{0i}^H \cdot H_0 &= H_i
\end{align*}
\]

$\Rightarrow$

\[
\begin{align*}
R_{0i} &= T_i \cdot T_0^T \\
R_{0i} &= C_i \cdot C_0^T \\
R_{0i} &= S_i \cdot S_0^T \\
R_{0i} &= H_i \cdot H_0^T
\end{align*}
\]

where the superscript $^T$ means transposed and $i$ ranges from 1 to 6, indicating the bony orientation from 30 to 180 degrees humeral elevation.

Def. 2a: Rotations with respect to the axes of the local coordinate system of the bone:

\[
\begin{align*}
T_0 \cdot R_{0i}^T &= T_i \\
C_0 \cdot R_{0i}^C &= C_i \\
S_0 \cdot R_{0i}^S &= S_i \\
H_0 \cdot R_{0i}^H &= H_i
\end{align*}
\]

$\Rightarrow$

\[
\begin{align*}
R_{0i} &= T_0^T \cdot T_i \\
R_{0i} &= C_0^T \cdot C_i \\
R_{0i} &= S_0^T \cdot S_i \\
R_{0i} &= H_0^T \cdot H_i
\end{align*}
\]

Instead of calculating the rotation matrices with respect to the rest position (0 degrees humeral elevation), calculations can also be made with respect to a virtual reference position, identical for each person. A virtual reference position is necessary, since for the thorax, clavicle and scapula no well-defined anatomical position exists. The local coordinate systems $T_a$, $C_a$, $S_a$ and $H_a$ in the virtual reference position were chosen to be positioned along the axes of the global coordinate system:
\[
T_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} ; \quad C_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} ; \quad S_a = \begin{bmatrix} -1 & 0 & 0 \\ 0 & -1 & 0 \\ 0 & 0 & 1 \end{bmatrix} ; \quad H_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & -1 & 0 \\ 0 & 0 & -1 \end{bmatrix} .
\]

(5)

In other words, for the thorax the sternum is along the vertical axis, the clavicle is along the frontal axis, the scapular spine is along the frontal axis and the scapular plane is parallel to the frontal plane, the humerus is along the vertical axis with the medial and lateral epicondyle in the frontal plane. For the clavicle and scapula the virtual reference position is physically impossible.

Then, analogously to Defs 1a and 2a the following rotations with respect to the anatomical position can be defined:

Def. 1b: Rotations with respect to the axes of the global coordinate system:

\[
R_{t_{ai}} \cdot T_a = T_i \quad \Rightarrow \quad R_{t_{ai}} = T_i \cdot T_a^T ; \\
R_{c_{ai}} \cdot C_a = C_i \quad \Rightarrow \quad R_{c_{ai}} = C_i \cdot C_a^T ; \\
R_{s_{ai}} \cdot S_a = S_i \quad \Rightarrow \quad R_{s_{ai}} = S_i \cdot S_a^T ; \\
R_{h_{ai}} \cdot H_a = H_i \quad \Rightarrow \quad R_{h_{ai}} = H_i \cdot H_a^T .
\]

Def. 2b: Rotations with respect to the axes of the local coordinate system of the bone:

\[
T_a \cdot R_{t_{ai}} = T_i \quad \Rightarrow \quad R_{t_{ai}} = T_a^T \cdot T_i ; \\
C_a \cdot R_{c_{ai}} = C_i \quad \Rightarrow \quad R_{c_{ai}} = C_a^T \cdot C_i ; \\
S_a \cdot R_{s_{ai}} = S_i \quad \Rightarrow \quad R_{s_{ai}} = S_a^T \cdot S_i ; \\
H_a \cdot R_{h_{ai}} = H_i \quad \Rightarrow \quad R_{h_{ai}} = H_a^T \cdot H_i .
\]

However, since $T_a$, $C_a$, $S_a$ and $H_a$ are unity matrices, some with negative values, the rotation matrices of Defs 1b and 2b are similar. Hence, the rotations are identical, only the sign is different. In the following there will only be referred to Def. 1b.

### III.4 Rotations in the joints

Instead of describing rotations of the bone, sometimes rotations of one bone with respect to another can provide a better understanding of the mechanism, i.e. which rotations occur in the articulation between two bones. Using orientation matrices as defined before, the orientation of one bone with respect to another can easily be defined. Hence, change of this orientation is due to rotations in...
the joint. Then again, these rotations in the joint can be defined in two ways: with respect to the global coordinate system and with respect to the local coordinate system of the proximal bone, as is common practice e.g. in the analysis of the lower extremity. Though, for the SC-, the AC- and GH-joint no flexion/extension or abduction/adduction axis with respect to the proximal bone are defined. In addition, the position of these axes would change with the hidden motions of the shoulder girdle underneath the skin. Therefore, only joint rotations with respect to the global coordinate system will be defined:

Def. 3a: Joint rotations with respect to the global coordinate system.

Clearly for the thorax no proximal joint is present, so its rotations are described with respect to the global coordinate system (see Def. 1a):

\[ R_{T\theta i} \cdot T_0 = T_i \quad \Rightarrow \quad R_{T\theta i} = T_i \cdot T_0^T. \]

Analogously to Def. 1a rotations of the clavicle can be described with respect to the global coordinate system as:

\[ R \cdot C_0 = C_i \quad \Rightarrow \quad R = C_i \cdot C_0^T. \]

Rotation matrix \( R \) is composed of rotations of the thorax and of rotations in the SC-joint, which are to be calculated:

\[ R_{T\theta i} \cdot R_{C\theta i} \cdot C_0 = C_i \quad \Rightarrow \quad R_{C\theta i} = R_{T\theta i}^T \cdot C_i \cdot C_0^T. \]

And analogously to this line of reasoning for rotations in the AC-joint:

\[ R_{T\theta i} \cdot R_{C\theta i} \cdot R_{S\theta i} \cdot S_0 = S_i \]
\[ \Rightarrow \quad R_{S\theta i} = R_{C\theta i}^T \cdot R_{T\theta i}^T \cdot S_i \cdot S_0^T. \]

and the GH-joint:

\[ R_{T\theta i} \cdot R_{C\theta i} \cdot R_{S\theta i} \cdot R_{H\theta i} \cdot H_0 = H_i \]
\[ \Rightarrow \quad R_{H\theta i} = R_{S\theta i}^T \cdot R_{C\theta i}^T \cdot R_{T\theta i}^T \cdot H_i \cdot H_0^T. \]

Interpretation of the last line yields that rotations of the humerus with respect to its initial position are the result of rotations of the thorax and rotations in the SC-, AC- and GH-joint.

Def. 3b: Similarly, joint rotations can be described with respect to the anatomical position of the bones. In Def. 3a the orientation matrices of the rest position (\( T_0, C_0, S_0 \) and \( H_0 \)) are replaced by the anatomical position (\( T_a, C_a, S_a \) and \( H_a \)).
III.5 Decomposition of rotation matrices into Euler angles

As explained in Section I.2, a rotation matrix \( R \) can be split into three elementary rotation matrices, e.g.:

\[
R = R_x(\alpha) \cdot R_y(\beta) \cdot R_z(\gamma),
\]

in which \( \alpha, \beta, \) and \( \gamma \) are Euler angles. Formally, there are two solutions for this decomposition. In this study always the smallest and most obvious solution is chosen. The order of the elementary matrices is important for the outcome which means that a large number of results are possible. Hereby a choice is presented which is assumed to be reasonably interpretable. Rotations are described by their rotation axis which is the only way for unambiguously describing 3D rotations. However, it should be noted that only the first rotation is around the axis as defined, the next rotations are around local axes, transformed by prior rotations, denoted by single quote ('') and double quote (") for the first and second local rotation, respectively.

The following rotations and their interpretation are defined:

Def. 1a: \( R_{t0} \):

- \( R_x \): backward/forward rotation around the frontal (X-)axis.
- \( R_y \): torsion around the vertical (Y'-)axis.
- \( R_z \): lateral flexion around the sagittal (Z''-)axis.

\( R_{c0} \):

- \( R_y \): pro/retraction (ventral/dorsal rotation) around the vertical (Y-)axis.
- \( R_z \): elevation/depression (cranial/caudal rotation) around the sagittal (Z'-)axis.
- \( R_x \): rotation around the frontal (X")-axis.

\( R_{s0} \):

- \( R_y \): pro/retraction around the vertical (Y-)axis.
- \( R_z \): lateral/medial rotation around the sagittal (Z'-)axis.
- \( R_x \): tipping forward/backward around the frontal (X")-axis.

\( R_{h0} \):

- \( R_y \): pole angle of the humerus with the frontal plane, rotation around the vertical Y-axis.
- \( R_z \): elevation angle with respect to the vertical axis, rotation around the Z'-axis.
- \( R_y \): axial rotation around the Y"-axis.

Def. 1b: Identical to Def. 1a.

Def. 2a: \( R_{t0} \):

- \( R_x \): backward/forward rotation around the local \( x \)-axis, pointing laterally.
- \( R_y \): torsion around the \( y_1' \)-axis along the sternum.
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\[ \mathbf{R}_z: \text{ lateral flexion around the local 'sagittal' } z_t''\text{-axis.} \]

\[ \mathbf{R}_{c0i}: \quad \mathbf{R}_y: \text{ pro/retraction (ventral/dorsal rotation) around the local 'vertical' } y_c\text{-axis.} \]
\[ \mathbf{R}_z: \text{ elevation/depression (cranial/caudal rotation) around the local 'sagittal' } z_c''\text{-axis.} \]

\[ \mathbf{R}_x: \text{ axial rotation around the length (} x_c''\text{)-axis.} \]

\[ \mathbf{R}_{s0i}: \quad \mathbf{R}_y: \text{ pro/retraction around the local } y_s\text{-axis in the scapular plane.} \]
\[ \mathbf{R}_z: \text{ lateral/medial rotation around the } z_s''\text{-axis perpendicular to the scapular plane.} \]

\[ \mathbf{R}_x: \text{ tipping forward/backward around the } x_s''\text{-axis along the scapular spine.} \]

\[ \mathbf{R}_{h0i}: \quad \mathbf{R}_y: \text{ pole angle of the humerus with the frontal plane, rotation around the } y_h''\text{-axis.} \]
\[ \mathbf{R}_z: \text{ elevation angle with respect to the vertical axis, rotation around the } z_h''\text{-axis.} \]
\[ \mathbf{R}_y: \text{ axial rotation around the } y_h''\text{-axis.} \]

Def. 3a: Identical as in Def. 1a, with the precaution that defined global axes are rotating with motions of the proximal bone. Thus, joint motions can be interpreted as rotations around global axes in the case that the proximal bone would not be moving.

Def. 3b: Identical as in Def. 1a, but rotations start from the virtual reference position. When actual joint rotations are analyzed, the rest position should be considered.

### III.6 Axial rotation of the clavicle

One apparent problem in calculating the rotations in the SC- and AC-joint is that only two points (IJ and AC) have been measured at the clavicle. Rotation around the line connecting these two points (axial rotation) can not be assessed. Therefore, axial rotation of the clavicle is estimated by minimizing the rotations in the AC-joint.

True axial rotation occurs around the local axes of the clavicle:

\[
\begin{align*}
C_0 \cdot \mathbf{R}_{c0i} & = C_i, \\
C_0 \cdot \mathbf{R}_y(\beta) \cdot \mathbf{R}_x(\gamma) \cdot \mathbf{R}_x(\alpha) & = C_i,
\end{align*}
\]  

(7)
in which $\beta$ and $\gamma$ can be calculated but $\alpha$ (axial rotation) only depends on the impaired definition of the local coordinate system of the clavicular, and therefore can not be calculated. However, using this notation the virtual position of the scapula $S'_1$ can be calculated as a result of the rotations of the clavicular and without rotations in the AC-joint:

$$C_0 . R_3(\beta) . R_2(\gamma) . R_1(\alpha) . C_0^T . S_0 = S'_1 .$$  \hspace{1cm} (8)

If no rotation would occur in the AC-joint, $S'_1$ is similar to the measured position of the scapula $S_1$, and the following equation would be true:

$$S'_1 . S_1^T = I .$$  \hspace{1cm} (9)

After substituting Eq. (9) in Eq. (8) the following equation can be derived:

$$C_0 . R_3(\beta) . R_2(\gamma) . R_1(\alpha) . C_0^T . S_0 . S_1^T - I = E .$$  \hspace{1cm} (10)

where $E$ is the error matrix due to the difference between $S'_1$ and $S_1$ and thus due to rotations in the AC-joint. Minimizing joint rotations is similar to minimizing the sum of the diagonal elements of $E$. Then, axial rotation $\alpha$ can be estimated by minimizing the sum of the diagonal elements of $E$ using a least squares criterion in a non-linear Gauss-Newton optimization procedure (Ljung, 1987).

**IV RESULTS**

**IV.1 Stick diagram**

In order to get an impression of the motion of the shoulder girdle during unloaded humeral abduction (ABU), Figs 5A, 5B and 5C show stick diagrams for seven positions measured, projected on the frontal, sagittal and transversal plane, respectively. The lateral end of the clavicular (AC) moves dorsally throughout the motion. It moves cranially up to 120 degrees abduction and then moves caudally again. The medial border of the scapula firstly shifts medially ('the scapular plane is oriented towards the plane of humeral elevation'). Then, Angulus Inferior (AI) slides over the thorax laterally and ventrally, ending aside the thorax even more ventral than AC: The scapula is 'laterally' rotating. Trigonum Spinae (TS) starts moving downwards and laterally immediately from 30 degrees abduction.
IV.2 Accuracy
During the measurements it is essential that in each position the same bony landmarks are recorded. Therefore, variation of the distance between two bony landmarks on the same bone is an indication of the measurement error.

Accuracy of the measurements was on-line controlled by constraining the distance to be in a range of 23.44 mm around the mean distance which was determined at the beginning of the recording session. This should result in a standard deviation of less than 6.00 mm for each distance, assuming a normal distribution (which is not the case since tails of the distribution are cut off). Per subject 28 positions were measured from which the standard deviation could be calculated afterwards. The mean standard deviation for each distance between two bony landmarks on the same bone (section II.1) is mostly less than 6 mm (Table 3).

The distance $d_{12}$ between two bony landmarks $P1 (x_1, y_1, z_1)$ and $P2 (x_2, y_2, z_2)$ is a non-linear

---

Figure 5:
Stick diagram of clavicular and scapular motions (one subject) at 0 and 180 degrees unloaded humeral abduction. Circles indicate intermediate positions.
IJ: Incisura Jugularis; AC: Acromioclavicular joint;
TS: Trigonum spinae; AI: Angulus inferior.
A: projection frontal plane;
B: projection sagittal plane;
C: projection transversal plane.
Table 3: Standard deviation in mm (N=28) of the distance between bony landmarks using rigid body assumptions. The second part of the table shows the standard deviation (in mm) of the bony landmarks calculated by distributing the standard deviation of the distance between bony landmarks.

<table>
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<tr>
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<th>M1</th>
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<th>M3</th>
<th>M4</th>
<th>M5</th>
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<th>M7</th>
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<th>M9</th>
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</table>

The function of the coordinates of each landmark:

\[ d_{12} = \sqrt{(x_2-x_1)^2 + (y_2-y_1)^2 + (z_2-z_1)^2} \]  \( (11) \)

The variance of \( d_{12} \) is a function of the variance of the coordinates of each landmark. According to Jenkins & Watts (1969) and Pronk & Padt (1986) this function can be approximated using a
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first-order Taylor series expansion:

\[
\text{var}(d_{12}) = \left[ \frac{\delta d_{12}}{\delta x_1} \right]^2 \left[ \frac{\delta d_{12}}{\delta y_1} \right]^2 \left[ \frac{\delta d_{12}}{\delta z_1} \right]^2 \left[ \frac{\delta d_{12}}{\delta x_2} \right]^2 \left[ \frac{\delta d_{12}}{\delta y_2} \right]^2 \left[ \frac{\delta d_{12}}{\delta z_2} \right]^2 \cdot \begin{bmatrix}
\text{var}(x_1) \\
\text{var}(y_1) \\
\text{var}(z_1) \\
\text{var}(x_2) \\
\text{var}(y_2) \\
\text{var}(z_2)
\end{bmatrix}, \tag{12}
\]

in which it is assumed that all errors in the coordinates are independent and hence the covariances are zero. Since no information is available about the variance of each coordinate, an isotropic distribution of the variance of the coordinates is assumed. However, the variance will be different for each bony landmark. Then the following equations are achieved:

\[
\text{var}(x_1) = \text{var}(y_1) = \text{var}(z_1) = \text{var}(P1); \tag{13}
\]

\[
\text{var}(x_2) = \text{var}(y_2) = \text{var}(z_2) = \text{var}(P2); \tag{14}
\]

\[
\text{var}(d_{12}) = \left( \frac{\delta d_{12}}{\delta x_1} \right)^2 + \left( \frac{\delta d_{12}}{\delta y_1} \right)^2 + \left( \frac{\delta d_{12}}{\delta z_1} \right)^2 \cdot \text{var}(P1) + \left( \frac{\delta d_{12}}{\delta x_2} \right)^2 + \left( \frac{\delta d_{12}}{\delta y_2} \right)^2 + \left( \frac{\delta d_{12}}{\delta z_2} \right)^2 \cdot \text{var}(P2) \tag{15}
\]

\[
- \text{var}(P1) + \text{var}(P2).
\]

Application of the last equation to the distribution of the variance of the distance between bony landmarks to the variance of the position of bony landmarks yields (Oranje, 1989):

\[
\begin{bmatrix}
\text{var}(I-JC) \\
\text{var}(AC-AA) \\
\text{var}(AC-TS) \\
\text{var}(AC-AL) \\
\text{var}(AA-TS) \\
\text{var}(AA-AL) \\
\text{var}(TS-AL) \\
\text{var}(IJ-PX) \\
\text{var}(IJ-C7) \\
\text{var}(PX-C7)
\end{bmatrix}
\begin{bmatrix}
1 & 1 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
0 & 1 & 1 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
0 & 1 & 0 & 1 & 0 & 0 & 0 & 0 & 0 & 0 \\
0 & 1 & 0 & 0 & 1 & 0 & 0 & 0 & 0 & 0 \\
0 & 0 & 1 & 1 & 0 & 0 & 0 & 0 & 0 & 0 \\
0 & 0 & 1 & 0 & 1 & 0 & 0 & 0 & 0 & 0 \\
0 & 0 & 0 & 1 & 1 & 0 & 0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 & 1 & 0 & 0 & 0 & 0 & 0 \\
0 & 1 & 0 & 0 & 0 & 1 & 0 & 0 & 0 & 0 \\
0 & 1 & 0 & 0 & 0 & 0 & 1 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
\text{var}(I-J) \\
\text{var}(AC) \\
\text{var}(AA) \\
\text{var}(TS) \\
\text{var}(AI) \\
\text{var}(PX) \\
\text{var}(C7)
\end{bmatrix}, \tag{16}
\]
or
\[ \mathbf{Y} = \mathbf{U} \cdot \mathbf{\Theta}. \]

(17)

This overdetermined set of equations can be solved for \( \mathbf{\Theta} \) using a least squares criterion:

\[ \mathbf{\Theta} = (\mathbf{U}^T \mathbf{U})^{-1} \mathbf{U}^T \mathbf{Y} \]

(18)

Only positive values for variances in \( \mathbf{\Theta} \) are accepted. In Table 3 the mean standard deviation per coordinate of each bony landmark is presented. Values are between 2.91 and 6.51 mm. Negative values, as found for subject m8, are due to inaccuracies in the determination of the variance of the distances.

The sensitivity of the variables for the estimated measurement errors is analyzed. Analogously to Eq. (12) also the variance of the calculated variables, i.e. Euler angles, can be determined as a function of the variance of the bony landmarks. Derivatives (e.g. \( \delta \alpha/\delta \mathbf{A}_x \)) have been calculated numerically and the variance per coordinate (Var(\( \mathbf{A} \))) has been used as presented in Table 3. In general, the mean standard deviation per Euler angle is less than two degrees. Lange et al. (1990) showed that the variance of Euler angles is relatively insensitive to measurement errors compared with helical axis parameters, it is independent from the rotation-step magnitude and it will decrease when the distance between markers increases.

IV.3 Euler angles

In Section III six different ways for describing bony motions are presented. The distinguishing factors are virtual reference (defined) position vs. rest position (0 degrees humeral elevation), local vs. global rotations and rotations of the bones vs. rotations in the joints:

1a: Rest position, global coordinate system, bones.
1b: Virtual reference position, global coordinate system, bones
2a: Rest position, local coordinate system, bones.
2b: Virtual reference position, local coordinate system, bones.
3a: Rest position, global coordinate system, joints.
3b: Virtual reference position, global coordinate system, joints.

Whenever motions refer to the virtual reference position, the difference between local and global coordinate system vanishes, since the local coordinate system is defined along the axes of the
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global coordinate system. So, the results of Def. 1b are interchangeable with Def. 2b.
Two criteria exist for preference of one of these descriptions: minimal variation between subjects
and interpretability. The first criterium is more or less objective and quantifiable, the second and
probably most important is subjective. Mean and standard deviation of ten subjects have been
calculated for the Euler angles describing bony orientations and rotations in the joints forDefs 1a,
2a, 3a, 1b and 3b. As an example Figs 6A to 6C show the standard deviation of the Euler angles
descending clavicular rotations and rotations in the SC-joint. Regarding bony rotations (Defs 1a,
2a and 1b) no important differences exists. Regarding joint rotations the standard deviation using
Def. 3b is slightly smaller than using Def. 3a. However, considering the insignificance of the
differences no preference of one of the proposed definitions exists.

Figs 7A to 7O show the scapular rotations and rotations in the AC-joint for ABU, ANU, ABL and
ANL using Defs 1a, 2a, 3a, 1b and 3b. If the rotations are described with respect to a local
coordinate system (Def. 2a, Figs 7D to 7F) the position of the coordinate axes should be kept in
mind. In Def. 2a the initial orientation can not be reconstructed and therefore this definition is not
preferred. In Def. 1b the initial orientation of the bone can be reconstructed using the Euler angles
at zero degrees humeral elevation (Fig. 7J to 7L). If one imagines the orientation of the bone, it
is very useful if the last rotation is around the length axis of the bone, since the direction of the
bone is determined using only the first two rotations. For the thorax and scapula no length axis

Figure 6:
Standard deviation of mean rotation angles of
the clavicular (N=10). Several motion
definitions as explained in the text are
compared.

---: Definition 1a.
---------: Definition 2a.
-....-....-: Definition 3a.
...........: Definition 1b.
o-o-o-o-o: Definition 3b.
exists, and the sternum and spine scapulae serve as imaginable bony ridge. Therefore, rotations with respect to a local coordinate system (Def. 2a, but also 1b) are preferred above rotations with respect to a global coordinate system of which the axes do not coincide with the defined length axis, like in Def. 1a (Fig. 7A to 7C). Summarizing, in our opinion Def. 1b is the best interpretable definition of bony rotations.

In order to get more insight in the functional anatomy of joints, joint rotations are calculated as well. Data obtained using the palpation technique are not accurate enough to determine joint translations which are likely to be small considering the capsule and shape of the joints (Van der Helm et al., 1991). Figs 7M to 7O show rotations of the SC-joint using Def. 3b. For other joints of the human body, e.g. the knee joint, this definition is usually used, describing the motion of the lower leg with respect to the upper leg. But for the knee joint actual rotations can start from the reference position where the global coordinate system coincides with the local coordinate system. In contrast, in the shoulder mechanism these coordinate systems do not coincide initially. In addition, no clear flexion/extension or adduction/abduction axis are defined. Therefore, since in the initial position (0 degrees abduction) joint rotations are not zero, the amount of additional joint rotations is obscured. Figs 7G to 7I (Def. 3a) show that actual joint rotations with respect to the rest position start at zero. For comparison of joint rotations between subjects this definition is preferred. It should be noted that the rotations presented describe the motion of the distal bone with respect to the proximal bone fixed in the global coordinate system.

IV.3.1 Rotations of thorax, clavicle, scapula and humerus

Using Def. 1b, rotations of thorax, clavicle, scapula and humerus are shown in Figs 8A to 8C, 9A to 9C, 10A to 10C and 11A to 11C, respectively. Mean values of ten subjects are presented for ABU, ANU, ABL and ANL. Differences between these conditions have been tested for significance using a non-parametric sign test with significance probability: p < 0.05 (two-tailed). Loaded and unloaded trials have been compiled for abduction and anteflexion comparison, and analogously the loaded and unloaded conditions have been compared. In this section only significant differences are discussed. It is remarkable that the difference between abduction and anteflexion is much larger than between loaded and unloaded humeral elevation. In fact, there is no significant difference between loaded and unloaded humeral elevation.

During anteflexion the thorax starts rotating backwards immediately, and continues throughout the motion (Fig. 8A). Up to 90 degrees abduction the thorax is more or less fixed, but then it rotates further backward than during anteflexion. During anteflexion subjects seem to stretch their hand
Figure 7: Mean rotation angles of the scapula (N=10), resulting from motion definitions as explained in the text. A - C: Def. 1a; D - F: Def. 2a; G - I: Def. 3a; J - L: Def. 1b; M - O: Def. 3b.

--- Abduction unloaded.
--- Anteflexion unloaded.
--- Abduction loaded.
--- Anteflexion loaded.
towards the marker, resulting in a torsion of the thorax around a vertical axis (Fig. 8B). Lateral flexion of the thorax is negligible, likely due to the instruction of two-sided humeral abduction and anteflexion (Fig. 8C).

Axial rotation of the clavicle is by far the largest rotation, ranging from zero (as defined in the initial position) up to 70 degrees (Fig. 9C). During abduction the axial rotation of the clavicle is somewhat larger than during anteflexion. Larger differences are noted for retraction and elevation. During abduction the clavicle is more retracted and more elevated, with a clear peak around 90 degrees abduction (Figs 9A and 9B, respectively). During the first 90 degrees of anteflexion the clavicle remains lower and is much less retracted. Though, even if the humerus is moved forward, the lateral end of the clavicle is moving backward.

During abduction the scapula is retracted more than during anteflexion (Fig. 10A). Johnston (1937) assumed that in the case of humeral elevation in the initial scapular plane which would have an angle of 30 degrees with the frontal plane, the scapula would not be pro- or retracted. A large number of authors have studied the scapulohumeral rhythm in the scapular plane (Freedman & Munro, 1966; Poppen & Walker, 1976; Wallace & Johnson, 1982; Kondo et al., 1984; Bagg & Forrest, 1988), relying on this assumption of Johnston. However, as can be seen in Fig. 10B, even during anteflexion the scapula retracts throughout the motion. Therefore, no humeral elevation in the scapular plane exists because the scapular plane is steadily moving. During abduction the scapula is laterally rotated more than during anteflexion (Fig. 10B). Because the clavicle is almost perpendicular to the scapular plane, axial rotation of the clavicle and lateral rotation of the scapula

Figure 8:
Mean rotation angles of the thorax as calculated using Def. 1b (N=10). A: Rotation x-axis (forward/backward rotation); B: Rotation y-axis (torsion around the vertical axis); C: Rotation z-axis (lateral flexion).
Figure 9:
Mean rotation angles of the clavicle as calculated using Def. 1b (N=10). A: Rotation y-axis (ventral/dorsal rotation); B: Rotation z-axis (cranial/caudal rotation); C: Rotation x-axis (axial rotation).


Figure 10:
Mean rotation angles of the scapula as calculated using Def. 1b (N=10). A: Rotation y-axis (pro/retraction); B: Rotation z-axis (medial/lateral rotation); C: Rotation x-axis (tipping forward/backward).


Figure 11:
Mean rotation angles of the humerus as calculated using Def. 1b (N=10). A: Rotation y-axis (pole angle); B: Rotation z-axis (elevation angle); C: Rotation y'-axis (axial rotation).

are roughly around the same axis. The magnitude of these rotations is almost equal. Through the lateral rotation of the scapula, AI is moved besides the thorax and the glenoid is directed more cranially which is useful for preventing the GH-joint from dislocation (Van der Helm, 1991b). When AI is moving to the lateral side of the thorax, AI is finally located more ventrally than AC (Fig. 5B). When AI follows the widening contour of the ellipsoidal-shaped thorax, the scapular plane is rotated around the scapular spine: Tipping backward around the \( x_s \)-axis along the scapular spine (AC - TS), see Fig. 10C.

The position of the humerus was standardized. This can be recognized since the pole angle for anteflexion is around 90 degrees and for abduction around zero degrees (Fig. 11A), whereas the elevation angle is linear (Fig. 11B). The pole angle and axial rotation are both around the local \( y \)-axis and are affecting each other to a large extent. Therefore, axial rotation is only imaginable combined with the complete rotation sequence (Fig. 11C). The difference in unloaded and loaded axial rotation is due to an adjustment to hold the weight in the hand.

**IV.3.2 Rotations in the SC-, AC- and GH-joint**

Using Def. 3a, rotations in the SC-, AC- and GH-joint are shown in Figs 12A to 12C, 13A to 13C and 14A to 14C, respectively. Mean values of ten subjects are presented for ABU, ANU, ABL and ANL. Rotations in the SC-joint are large, up to 60 degrees (Fig. 12C). It is remarkable that rotations in the AC-joint are almost nil as a result of the estimation of the axial rotation of the clavicula (Figs 13A to 13C). It is likely that the strong extracapsular conoid and trapezoid ligaments prevent rotations of the AC-joint. Rotations of the humerus with respect to the scapula are very interesting (Figs 14A to 14C). The difference in pole angle between abduction and anteflexion is less than 90 degrees, indicating that orientation of the scapula is affected by the elevation plane. The elevation angle in the GH-joint is around 100 degrees which means that additional elevation is due to the rotations of the scapula. Axial rotation in the GH-joint is very difficult to interpret.

**IV.4 Deformation of the thorax**

In this study it is assumed that the thorax is a rigid body without deformations. This assumption can partly be checked by the position of the landmarks II, PX, C7 and T8 relative to each other. Deformation of the thorax can be operationalized by the angle between the lines through II-PX and C7-T8, i.e. the angle between the spine and sternum. Standard deviation of this angle ranges between 1.6 and 2.7 degrees (10 subjects) which is within the measurement error. It is concluded
Figure 12: Mean rotation angles in the sternoclavicular joint as calculated using Def. 3a (N=10).

- ---: Abduction unloaded (ABU).
- ---: Anteflexion unloaded (ANU).
- ---: Abduction loaded (ABL).
- ---: Anteflexion loaded (ANL).

Figure 13: Mean rotation angles in the acromioclavicular joint as calculated using Def. 1b (N=10).

- ---: Abduction unloaded (ABU).
- ---: Anteflexion unloaded (ANU).
- ---: Abduction loaded (ABL).
- ---: Anteflexion loaded (ANL).

Figure 14: Mean rotation angles in the glenohumeral joint as calculated using Def. 1b (N=10).

- ---: Abduction unloaded (ABU).
- ---: Anteflexion unloaded (ANU).
- ---: Abduction loaded (ABL).
- ---: Anteflexion loaded (ANL).
that the thorax is not deformed during humeral abduction and anteflexion when the effect of in- and expiration is excluded (Section II.4).

V DISCUSSION

The shoulder mechanism is a chain of links consisting of thorax, clavicula, scapula and humerus. The position of the humerus is the result of simultaneous rotations in the SC-, AC- and GH-joint. Very often only rotations of the humerus with respect to the trunk have been described (Poppen & Walker, 1976; Engin & Peindl, 1987), neglecting the large rotations of the scapula. This is comparable to describing motions of the lower leg with respect to the hip whereas motions of the upper leg are disregarded: no joint rotations can be calculated and very little of muscle functions can be analyzed.

The humerus has an extremely large range of motion with respect to the global coordinate system: 180 degrees elevation actually means turning upside-down. Joint rotation of 180 degrees would imply that origin and insertion of the muscles are fully twisted and muscles would be entrapped in the articulation. Hence, this large range of motion requires motions of a supplementary bone in order to provide a well-directed articulation surface preventing dislocation of the joint. Motions of the scapula are necessary to realize a stable base for motions of the humerus and serve to transmit forces exerted at the humerus to the trunk. Thus, it can be expected that the scapular position depends on the position of the humerus, and probably on the load at the humerus.

Within the shoulder girdle, the scapula is the propelled bone. Only a few muscles are attached to the clavicula. The clavicula is acting like a strut for the acromion: It provides an excellent supporting point at which the scapula can be used as a lever for the thoracoscapular muscles, especially the m. serratus anterior (Van der Helm, 1991b). Motions of the clavicula are induced by the scapular motions. But in reverse, scapular motions are constrained by the clavicula and the thorax which determine the possibilities of the scapula to reach the optimal position for supporting the humerus.

The palpation technique used in this study is only applicable in static situations. Whether the results can be extrapolated to dynamic situations is simply not known, since dynamic recordings of the shoulder girdle have never been published. Thus, whenever in this study the word 'motion' is mentioned, just the change from one static position to another is meant.

In studies of the lower extremities often locomotion is used for function analysis. Analysis of the motion of the shoulder mechanism is impeded because no standard task for the upper extremity exists. In this study humeral abduction and anteflexion are analysed. Subjects were asked to
position their arm actively, because it is our opinion that the scapulohumeral relationship is mainly
due to the active properties of the muscles, i.e. force-length relation and optimal use of energy,
and not to passive properties.

The accuracy of the palpator is within the range of the best 3D motion recording methods: 0.96
mm standard deviation per coordinate (Hatze, 1988; Pronk & Van der Helm, 1991). Most of the
measurement error is due to the difficulties in palpating the bony landmarks, resulting in a standard
deviation between 1.5 and 5.2 mm per coordinate of on-line checked bony landmarks (Table 3).
However, the position of the bones is directly recorded. When markers glued to the skin are used
for motion recording, the displacement of the skin with respect to the bones is unknown. Weeren
et al. (1988) recorded in direct measurements a skin-to-bone motion ranging from 29 to 146 mm
(peak-peak) during horse locomotion. Capozzo et al. (1988) estimated a maximal marker variation
of 10 mm. Therefore, in comparison with accepted standards a fairly accurate recording of the
motion of the shoulder mechanism has been achieved.

IV.1 Comparison with literature

Thus far, in literature no data were reported for 3D motions of the shoulder mechanism. Therefore,
for comparison with former studies like the outstanding work of Inman et al. (1944) the 2D
scapulohumeral rhythm is used. The scapulohumeral rhythm has been defined by Inman and
coworkers as the relation between the angle of the scapular spine, projected on the frontal plane,
and the humeral elevation angle. Figs 15A and 15B show that the spinohumeral angles,

![Graph A](image1)

![Graph B](image2)

**Figure 15:** Scapulohumeral rhythm. Dotted lines: 95% confidence interval according to Inman et al. (1944);
Continuous lines: Subjects recorded in this study.
A: Abduction unloaded (ABU).
B: Anteflexion unloaded (ANU).
operationalized in this study as the line through AA and TS, projected on the frontal plane, and humeral elevation angle, are somewhat smaller with respect to the 95% confidence interval as found by Inman et al. (1944), for humeral abduction and anteflexion respectively. The linearity of this relation is remarkable since it contains the 3D motion of the scapular spine distorted by projection on the frontal plane.

In this study only two points at the clavicle were measured, and therefore, no axial rotation was recorded. Axial rotation of the clavicle was estimated by minimizing rotations in the AC-joint. Including the axial rotation enabled a complete 3D description of motions of the clavicle. Only in a few studies rotations of the clavicle around its length axis were measured using pins inserted through the skin into the bone (Inman et al., 1944; Inman & Saunders, 1946; Kennedy & Cameron, 1954; Fontijn, 1965; Rockwood, 1975). Inman & Saunders (1946) did not describe their measurement method, but according to photographs included we inferred that they inserted a pin pointing ventrally close to the SC-joint. The angle between the pin and the transversal plane has been presented as axial rotation. This procedure has been simulated and the results are presented in Fig. 16. Axial rotation estimated in this study is considerably larger than the axial rotation as measured by Inman & Saunders (1946). The cause of this difference is rather speculative. The optimization criterium used seems to represent a physically good interpretable phenomenon: Axial rotation of the clavicle results in minimal rotations in the AC-joint. Residual joint rotations are very small (Figs 13A to 13C). This is probably caused by the function of the very strong conoid and trapezoid ligaments: Limiting the motion range of the AC-joint by transducing the lateral rotating motion of the scapula to the axial rotation of the clavicle, thereby avoiding torsion in the capsule of the AC-joint. The difference with the study of Inman & Saunders (1946) could be explained by the fact that motions in the AC-joint are not restricted as much as is estimated in the optimization procedure, maybe because the ligaments are not tight immediately when the motion starts. Maybe Inman & Saunders have recorded sitting subjects, thereby limiting the backward rotation of the thorax. In addition, insertion of the pins into the clavicle could hamper the natural motion of the clavicle.

V.2 Interpretation

In this study, a complete 3D description of the motion of the shoulder mechanism for a normal population is presented. Motion of the mechanism is described in terms of rotations in order to exclude interindividual differences in bony dimensions. In addition, translations in the joints are neglected, because they are considered to be very small with respect to the large rotations in the
joints and they can not be distinguished from the measurement error. Rotations of a complex mechanism as the shoulder can be described unambiguously in technical terms using rotation matrices. For clinical use, these rotation matrices pose several problems: They are very difficult to interpret and therefore not very useful to distinguish normal and abnormal motions. A useful method of motion description should enable a comparison between subjects, for instance the difference between a patient's pattern and a normal population. Since the standard deviation of several motion definitions do no differ very much, a final choice is mainly based on subjective arguments, i.e. whether or not the resulting rotations are imaginable. Normally, joint rotations are defined with respect to the proximal bone. In the shoulder girdle the skin hides the position of the bones from the eye. In addition, due to the lack of definition of an anatomical position, no main rotation axes like a flexion/extension or abduction/adduction axis are defined. Therefore, joint rotations are described around axes of the global coordinate system with respect to the rest position of the bones (Def. 3a; Figs 12 to 14).

Clearly, the scapular orientation is the result of the rotations in the SC- and AC-joints. The shoulder girdle is a closed-chain mechanism in which the scapula is the propelled bone. The clavicular is merely forced to rotate accordingly. The orientation of the scapula with respect to the trunk is more important than the orientation with respect to the clavicle. In addition, the humerus is positioned with respect to the trunk and not to the scapula. Therefore, in contrast to the motion description of all other parts of the human body, the orientation of the bones with respect to the global coordinate system is presented (Def. 1b; Figs 8 to 11).

Generally, humans are unable to consciously control individual joint rotations in the shoulder mechanism. The seven degrees of freedom are not strictly independent: There is a strong relation

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**Figure 16:**
Simulation of axial rotation of the clavicle, compared to the measurements by Inman et al. (1944) using inserted pins.

- - - - - - - -: Abduction unloaded, this study.
- - - - - - - -: Anteflexion unloaded, this study.
- - - - - - - -: Abduction unloaded, Inman et al. (1944).
- - - - - - - -: Anteflexion unloaded, Inman et al. (1944).
between these degrees of freedom (Van der Helm, 1991a). In Figs 8 to 11 it is shown that scapular motions depend on the plane in which the humerus is elevated: Abduction or anteflexion. The effect of additional load in the hand on scapular motions is small, maybe due to the small loads applied. An interesting question is if higher loads would have resulted in larger differences in scapular motion, but the duration of the measurement sessions in this study did not permit higher loads. The nature of the relation between scapular and humeral motions is for the time being unknown. It can be speculated that motions of the scapula are optimized, e.g. to minimize energy, during stabilization of the humerus or controlling motions of the upper extremity.

As stated in Section 1, two problems arise in interpretation of 3D motions using Euler angles: the order problem and the interaction problem.

The order of Euler angles affects the results which means that in interpretation of the results the order should be considered. In a spherical joint like the GH-joint the definition of main rotations (e.g. anteflexion/retroflexion and abduction/adduction) is only applicable if the motion takes place in one of the main anatomical planes. A combination of rotations implies that the second and third rotation are not around one of the main anatomical axes and are strictly seen not well-defined main rotations. In this study the order problem is acknowledged, but for sake of better understanding the terminology of the main rotations has been maintained.

Position coordinates in a global coordinate system form a linear system which means that a change of Δy describes an identical position change independent of the values of x, y and z. The orientation of a bone with respect to a global coordinate system is described by three variables, e.g. Euler angles or a helical axis. However, these variables form a non-linear system which means that the effect of changing one variable depends on the value of the other variables: The 'working' point. Subsequent rotations around the y-, z- and x-axis do not mean that the last Euler angle describes a rotation around the original x-axis, but a rotation around a local x-axis rotated by the former rotations around the y- and z-axis. Thus, the value calculated for each rotation is affected by the values of the other rotations: The interaction problem. Since rotations take place in 3D space three strongly related variables are required to describe the rotation of a bone. All three variables need to be simultaneously comprehended for complete understanding of the motion which is very difficult for human beings. There is no method available describing 3D rotations with a linear system and hence avoiding the interaction problem.

The motion of one bone with respect to another is a combination of rotations and translations. The choice of a model structure of rotational and translational axes is decisive for the calculated values of rotations and translations. Pennock & Clark (1990) propose a combination of fixed and floating
Three-dimensional recording and description of motions

rotation axes for the knee joint, suggestively based on the geometry of the bones. However, the
technical solution for the motion description problem is not identical to the way the motion actually
takes place. In addition, if the rotations axes are not orthogonal, some orientations of one bone
with respect to another can not even be described (Pronk, 1984). In contrast to Pennock & Clark
(1990), it is our opinion that preference of a set of rotation axes and the sequence of rotations is
rather subjective, and at least the rotation axes should be orthogonal.
Direction of the helical axis and the magnitude of rotation around this axis are solely determined
by the rotation of the bone. The position vector of the axis and translation along the axis are
determined by the translation of the bone, but how the latter translation is distributed depends on
the direction of the helical axis. The instantaneous helical axis represents a rotation axis of the
joint. However, for joints with more than one DOF, the joint rotation axis is not unique but
depends on the actual motion pathway (Blankervoort et al., 1990). The relation with passive motion
constraints as articular surfaces and ligaments is not yet clear. Thus far, discrete helical axes seems
to be only useful for motion description, but they are hard to interpret.
The standard deviation, as presented in Fig. 6 is due to the summation of interindividual variations
and measurement errors. As calculated in Section IV.3, the measurement error which is caused by
difficulties in retrieving the bony landmarks, results in a maximal standard deviation for the Euler
angles of less than 2 degrees. The measurement method as presented in this study is accurate
enough to distinguish between individuals. Obviously, most of the variation is due to interindividual
differences. Large interindividual differences hamper the discrimination between patients and a
normal population: Even if the separate rotations for a patient are within the 95% confidence
interval, combination of these rotations could result in an extraordinary motion.

Concluding remarks
1. A palpation method is presented for recording the three-dimensional position of the bones of the
shoulder mechanism in static situations. The accuracy of the palpation method is within the
range of the best three-dimensional recording methods.
2. Two methods for motion description in the shoulder mechanism are preferred. Bony rotations
are described with respect to a virtual reference position as rotations in the global coordinate
system. Joint rotations are described with respect to the rest position as rotations in the global
coordinate system to which the proximal bone is fixed.
3. Small loads applied at the hand do not affect the motions of the shoulder mechanism bones. A
significant difference occurs between the thoracic, clavicular and scapular rotations during
abduction and anteflexion.

4. Interpretation of three-dimensional rotations is hampered by the interaction between the three variables necessary for motion description.

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Chapter 8

The loading of shoulder girdle muscles in consequence of a glenohumeral arthrodesis.

F.C.T. van der Helm, G.M. Pronk,


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Abstract

Patients with a brachial plexus lesion sometimes undergo a glenohumeral arthrodesis operation in which the humerus is fused to the scapula. The fusion angles between scapula and humerus determine the function restoration after the operation. In this study the effect of fusion angles on the hand position and maximal force exertion has been simulated using a musculoskeletal model of the shoulder girdle. Mean scapular and clavicular motions (10 subjects) are used as input variables. Mean fusion angles (18 patients with a glenohumeral arthrodesis) are used as reference position. Output variables are the maximal force which can be exerted with the hand and muscle forces of the thoracoscapular muscles. In order to achieve a mobility area in the midsagittal plane, the humerus should be endorotated 60 degrees with a little abduction and anteflexion. Maximal force can be exerted when the scapula is laterally rotated and protracted besides the thorax. In that position m. serratus anterior gets a good moment arm to counterbalance the external moment. It is recommended to use an external fixator for the glenohumeral arthrodesis in order to be able to set the fusion angles outside the operation theater.
I Introduction
Sometimes, due to a brachial plexus lesion muscles controlling the motions of the upper extremity are paralysed. Which muscles are affected depends on the location of the lesion. If the patient lacks control of one or more joints, he/she will be unable to position his/her hand. In the rehabilitation process it will be attempted to provide the patient some sort of functional control of the joint motions (Stassen, 1989), see Fig. 1. If after two years of rehabilitation and, when applicable, attempts to neurosurgical repair, muscles crossing the glenohumeral joint are still paralysed, it can be advisable to arthrodize the joint (Jaspers, 1990). In a glenohumeral arthrodesis the humerus is fused to the scapula, and the combination of both bones moves as one. Therefore, the humerus can be positioned again by actions of the thoracoscapular muscles which are in most cases unaffected by the brachial plexus lesion. In addition, the arthrodesis will prevent subluxation of the glenohumeral joint. Elbow control can be established by an elbow orthosis (Cool, 1989), allowing either an unconstrained hanging of the forearm or a fixed position of 90 degrees elbow flexion. Wrist control can be established by an arthrodesis or an orthosis. Ultimate goal is recovering of the ability to position the hand and some force exertion with the hand. In addition, the arm is under control again during walking and running which yields better cosmesis of wearing on the one hand

Figure 1: The control of the hand (Stassen, 1989).
and protects the arm against damage on the other. In other words, in the patient's perception the arm has become again a part of the body.

Pronk (1991) has made an analysis of the function restoration by interviewing 18 patients and by recording their motions of the upper extremity. If the patient has left some control of the hand muscles, he/she will be able to use his/her hand again in daily activities. Even if no hand functions are left, the patient will be able to press or lift with his/her hand, and press his/her elbow against the side of the trunk, holding objects in between. However, the most important advantage is that the upper arm is not dangling any more aside and in front of the body, often hitting the genitals with the hand.

The amount of function restoration clearly depends on the fusion angles between scapula and humerus and on the condition of the thoracoscopular muscles. Pronk (1991) demonstrated that in literature a vast variety exists in recommended fusion angles, ranging from 10 to 80 degrees abduction, 10 to 60 degrees forward flexion and 45 degrees endorotation to 45 degrees exorotation. Apparent problem in determining the fusion angles is that for definition of the three-dimensional position of one bone to another, three subsequent rotations around well-defined axes are necessary. Then, the order of the subsequent rotations as well as the location of the axes are important. The lack of unambiguous definitions in clinical practice could explain the variety in advised fusion angles. In addition, the fusion angles are defined with respect to the trunk instead of with respect to the scapula. During surgery the patient is mostly in horizontal position, causing the scapula to shift over the thorax which hampers an accurate positioning of the humerus.

The desired fusion angles depend on the individual demands of the patient concerning his/her daily activities, the remaining functions of the upper extremity and the patient's morphology. If some hand function is left, good positioning of the hand is very important. If some thoracoscopular muscles like the m. serratus anterior are paralysed, the operation is performed merely for cosmetic purposes and to provide the patient with some proprioceptive feedback of the arm motions in order to protect the arm. After the arthrodesis motions of the humerus are due to motions of the scapula. Orientation and position of the scapula are constrained by the thorax and clavícula.

As a result of the operation the glenohumeral joint is destroyed irreversibly. Combined with the lack of accurate predictions of the results this fact impedes the patient's decision whether or not to operate. Of late, an external fixator is used to perform the glenohumeral arthrodesis (Nieuwenhuis & Pronk, 1989). Using the external fixator the arthrodizing operation can be split up into two sessions. Firstly, pins are inserted into the scapula and humerus and the fixator is placed. Then, while the glenohumeral joint is still intact, the patient in cooperation with the
rehabilitation team is able to adjust the fusion angles until a satisfactory position is achieved. Secondly, during the denuding of the articular cartilage the fixator is removed while the fusion angles are preserved by the settings of the fixator. In addition, using the telescopic tubes of the fixator the resulting gap between the bones can be closed and a compression force can be applied. An additional advantage is that the patient is able to make small movements with his/her arm, which prevents muscle atrophy. It has been concluded that the external fixator is an important tool in conduction of the glenohumeral arthrodesis (Nieuwenhuis & Pronk, 1989). However, still a few problems remain. Wearing the external fixator, motions of the upper extremity are hampered by the tension of the skin around the inserted pins. Therefore, the fusion angles are chosen mainly based on the rest position. The patients lack experiences in real life motions of the hand and their consequences. In addition, due to the weight of the upper arm and the resulting moment on the scapula, the loading of the thoracoascapular muscles changes drastically. Very little is known about the newly acquired muscle functions and the resulting forces on e.g. the hand.

Such a function analysis can be performed using model simulations. Therefore, a biomechanical model has been developed describing the motions and muscle forces of the shoulder girdle, extended with a fused humerus. This model is a subset of a larger model describing the kinematic and dynamic behavior of the complete shoulder mechanism (Van der Helm, 1991b).

Aim of the present study is to show the effect of fusion angles on the function restoration of the upper extremity. The achieved mobility area of the hand and the maximal force that can be exerted in that position will be analysed. Input variables of the model will be the position of the scapula and external load, and the range of fusion angles, and output variables will be hand position and muscle forces (Fig. 2).

![Figure 2: Block diagram of the musculoskeletal model. Input variables are the position of the shoulder girdle bones, the fusion angles at the glenohumeral joint and external load at the hand. Output variables are hand position and muscle forces.](image-url)
Data for the model have been obtained in several studies. Mean fusion angles between scapula and humerus have been derived in an extensive motion recording study of 18 patients (Pronk, 1991). Morphological data have been recorded in a cadaver study (Veeger et al., 1991; Van der Helm et al., 1991; Van der Helm & Veenbaas, 1991). Measurements of the scapular motion of healthy subjects were described in Van der Helm & Pronk (1991a). Scapular motions of healthy subjects are used as input for the model, since motions of patients are definitely already affected by the arthrodesis. The effect of the weight of the humerus on the scapular motions after an arthrodesis is unknown. It is assumed that normal scapular motions are still possible as they were recorded during humeral abduction and anteflexion. Due to the impossibility to acquire morphological data of individual patients, predictions have been limited to a general case.

II Model structure
In modelling the shoulder mechanism a finite element method has been used which is implemented in the computer program SPACAR (Werff, 1977; Werff & Jonker, 1983; Jonker, 1988; Van der Helm, 1991a). This finite element method has specially been developed for multi-degree of freedom spatial mechanisms with flexible bodies. In this application morphological structures are represented by elements of which the kinematic and dynamic behavior is well-known. Then, the behavior of the complete mechanism can be calculated.

In the finite element method bones are represented by rigid BEAM elements: The clavicle by one BEAM between the sternoclavicular and acromioclavicular joint; the scapula by two BEAM elements, between the acromioclavicular joint and trignonum spineae (TS) and between TS and angulus inferior (AI), respectively; the humerus by one BEAM from the glenohumeral joint to the elbow joint; the forearm bones by one BEAM from the elbow joint to the hand. The sternoclavicular, acromioclavicular and glenohumeral joint are each represented as spherical joints by three orthogonal HINGE joints, the elbow joint is composed of two HINGE joints allowing flexion/extension and pro/supination. The wrist joint is omitted since most of the patients do not have active wrist motion, and if they do, it does not add much to the hand position. Ligaments are represented by flexible TRUSS elements. The medial border of the scapula is pressed against the thoracic wall by the combined action of the m. serratus anterior and m. rhomboideus. The motion constraints of the so-called scapulothoracic gliding plane are represented by two SURFACE elements, each limiting one point of the medial border to slide over the surface of the thorax, modelled as an ellipsoid. Muscles are represented by one or more force generating 'active' TRUSS or CURVED-TRUSS elements, of which the latter are wrapped around the surface of bony
contours (Van der Helm, 1991). Generally, more than one muscle element is necessary for representing the mechanical effect of muscles, considering the sometimes large attachment site and muscle architecture (Van der Helm & Veenbaas, 1991).

III Model simulations

A global coordinate system has been defined with its origin at Incisura Jugularis (IJ) and the axes along the main anatomical axes (X-axis pointing from medial to lateral, Y-axis pointing from caudal to cranial, Z-axis pointing from ventral to dorsal; right shoulder). The global coordinate system is fixed to the thorax.

The orientation of scapula and humerus are determined by their local coordinate systems (Fig. 3). The $x_s$-axis of the local coordinate system of the scapula is pointing from the tip of the acromion (AC) to trigonum spinae (TS), $y_s$ is in the scapular plane defined by the AC, TS and angulus inferior (AI), pointing caudally, and the $z_s$-axis is perpendicular to the scapular plane, pointing dorsally. The local $x_h$-axis of the local coordinate system of the humerus is defined along the length axis of the humerus, pointing cranially, $x_h$ is along the flexion/extension axis of the elbow, pointing laterally, and $z_h$ is perpendicular to the previously defined $x_h$- and $z_h$-axis, pointing along a 90 degrees flexed elbow. The orientation of the local coordinate system with respect to the global coordinate system is described by the $3 \times 3$ orientation matrix $S$ of the scapula and $H$ of the humerus, where $S = [x_s \ y_s \ z_s]$ and $H = [x_h \ y_h \ z_h]$. Then, fusion matrix $F$ is obtained by

$$ S \cdot F = H \quad \Rightarrow \quad F = S^T \cdot H. $$

Fusion positions of scapula and humerus have been recorded at 18 patients with a glenohumeral
Glenohumeral arthrodesis

arthrodesis. Mean fusion matrix $F_m$ is obtained by minimizing the deviation of the diagonal elements of all matrices $F$ using a least squares criterion by means of an iterative Gauss-Newton method (Ljung, 1987) which results in (Pronk, 1991):

$$
F_m = \begin{bmatrix}
-0.9500 & 0.0664 & -0.3051 \\
-0.1160 & -0.9822 & 0.1475 \\
-0.2899 & 0.1755 & 0.9408
\end{bmatrix}.
$$

(2)

Motions of the shoulder mechanism during humeral abduction have been recorded at 10 healthy subjects using a palpation technique (Van der Helm & Pronk, 1991). Using this technique bony landmarks have been palpated and its 3D coordinates have been recorded by means of the so-called palpator, a kind of spatial digitizer (Pronk & Van der Helm, 1991). Fig. 4 shows the bony landmarks which are recorded. Three bony landmarks at the same bone are sufficient to reconstruct the 3D position of the bone. Bony landmarks AC, TS and AI are used similarly as described above to reconstruct the local coordinate system of the scapula in the restposition (0 degrees humeral abduction), denoted by $S_0$. Then, the mean orientation matrix $S_{0m}$ of the local coordinate system can be calculated, which results in:

$$
S_{0m} = \begin{bmatrix}
-0.7381 & 0.2274 & 0.6352 \\
-0.2015 & -0.9728 & 0.1142 \\
0.6439 & -0.0437 & 0.7638
\end{bmatrix}.
$$

(3)

Figure 4: Bony landmarks used in the palpation experiment.
Hence, an average orientation $H_{0m}$ of a fused humerus with respect to the global coordinate system can be calculated:

$$H_{0m} - S_{om} F_m - \begin{bmatrix} 0.4906 & -0.1609 & 0.8564 \\ 0.2712 & 0.9622 & 0.0254 \\ -0.8281 & 0.2198 & 0.5157 \end{bmatrix}.$$  \hspace{1cm} (4)

For the sake of better understanding, $H_{0m}$ can be decomposed into three Euler angles. These Euler angles are chosen to be defined around the axes of the global coordinate system. Rotations described by Euler angles refer to the anatomical position of the humerus $H_a$, with both its length axis and the flexion/extension axis in the frontal plane:

$$H_a - \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix}.$$  \hspace{1cm} (5)

In this decomposition the order of rotations is important. A rotation order is chosen which can be applied in clinical practice: Endorotation $\beta$ around the global Y-axis, then abduction $\gamma$ around the global Z-axis and finally anteflexion $\alpha$ around the global X-axis. These rotations can be denoted in elementary matrices, $R_Y(\beta), R_Z(\gamma)$ and $R_X(\alpha)$, respectively. Rotations around axes of the global coordinate system are comparable with premultiplying with the rotation matrices (Pronk, 1984; Van der Helm & Pronk, 1991\textsuperscript{a}). Then:

$$R_X(\alpha) \cdot R_Z(\gamma) \cdot R_Y(\beta) \cdot H_a - H_{0m},$$  \hspace{1cm} (6)

or:

$$\begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\alpha) & -\sin(\alpha) \\ 0 & \sin(\alpha) & \cos(\alpha) \end{bmatrix} \begin{bmatrix} \cos(\gamma) & -\sin(\gamma) & 0 \\ \sin(\gamma) & \cos(\gamma) & 0 \end{bmatrix} \begin{bmatrix} \cos(\beta) & 0 & \sin(\beta) \\ 0 & 1 & 0 \\ -\sin(\beta) & 0 & \cos(\beta) \end{bmatrix} \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} - H_{0m}.$$  \hspace{1cm} (7)

Generally, two solutions exist solving this equation for $\alpha$, $\beta$ and $\gamma$. The solution yielding the smallest values has been chosen. Using the average orientation of the scapula (obtained from 10 healthy subjects) and the mean fusion matrix (obtained from 18 patients), the mean orientation of the fused humerus with respect to the thorax is as follows: endorotation 60.2 degrees, abduction 9.3 degrees and anteflexion 12.9 degrees. If the humeral position is known, the position of the hand can be calculated. Most of the time during object manipulation, the elbow is flexed 90
degrees using the elbow orthosis. Combined with the bony dimensions as obtained in the cadaver study, the resulting position of the hand is shown in Fig. 5.

Input variables are the motions of the scapula. Scapular motions are constrained by the clavicle, restricting the acromion to move on a sphere around the sternoclavicular joint, and by the scapulathoracic gliding plane, restricting the medial border of the scapula to move on an ellipsoid which represents the thorax. The sternoclavicular and acromioclavicular joint can be assumed as spherical joints with each 3 degrees of freedom (DOF). The connection between thorax and scapula reduces the number of DOF with two, leaving a total of 4 DOF for the shoulder girdle (Van der Helm, 1991a). Due to the complex motion constraints, the use of Euler angles as input for the description of the scapular orientation would lead to odd positions on the thorax. Therefore, some position coordinates of bony landmarks have been chosen as generalized coordinates which are comparable with the Lagrangian approach of modelling multibody systems. These generalized coordinates define the position of the shoulder girdle unambiguously. The number of generalized coordinates is equal to the number of DOF. In this simulation study the y- and z-coordinate of the AC and the x-coordinate of TS have been used as input variables. Values are obtained as mean values from 10 subjects in a motion recording study using a palpation technique (Van der Helm & Pronk, 1991b). Still one DOF remains: The axial rotation of the clavicle. This rotation has not been measured since only two bony landmarks could be accurately distinguished at the clavicle. However, if the conoid ligament is modelled as a rigid TRUSS element, the axial rotation is

Figure 5: Initial position of the hand, resulting from the mean scapular position and the mean fusion angles, elbow flexed 90 degrees (cranial view).
defined and happens to be in a normal range (Van der Helm, 1991b).

Scapular motions during humeral abduction have been used as input. These motions are assumed to be a reasonable subset of the scapular mobility. Humeral anteflexion has been tested as well but did not add much to the mobility area of the hand. Scapular positions have been measured statically at approximately 0, 30, 60, 90, 120, 150 and 180 degrees of humeral abduction, and will be referred to as position 1 to 7, respectively. Thus, position 4 of the scapula is the mean position recorded in healthy subjects at 90 degrees humeral abduction.

The whole range of scapular positions is simulated for every fusion position of the humerus. At position 1 (mean rest position of the scapula), the humerus is moved to a certain position defined by a combination of fusion angles. Next, the scapula is moved to position 2, 3, etc., forcing the humerus to move accordingly. Starting from the initial fusion position (60.2 degrees endorotation, 9.3 degrees abduction, 12.9 degrees anteflexion), fusion angles are changed for each rotation one at the time, while the other two rotations remain at their initial value. Firstly, the endorotation angle is set to 0, 30, 60.2 (the initial fusion position), and 90 degrees. Secondly, the abduction angle is set to 0, 30 and 60 degrees, and finally the anteflexion angle is set to 0 and 30 degrees. Muscle forces are estimated using the inverse dynamic approach. Since more muscle elements are present than DOF in the model, the model is indeterminate. A solution can be obtained using an optimizing procedure. In this study minimizing of the sum of squared muscle stresses

![Figure 6: Hand position resulting from motions of the scapula, mean fusion angles and elbow flexed 90 degrees, projected on the transversal plane. Inside the circles is indicated the maximal lifting force (in N) at the hand.](image-url)
(min \Sigma (\text{force/PCSA})^2) \text{ is used as optimizing criterion. The criterion is subject to equality constraints imposed by the motion equations, inequality constraints imposed by the scapulothoracic gliding plane (only compression forces) and ligaments (only traction forces). Furthermore, muscle forces are limited to range between zero and maximal force as derived from the PCSA (maximal force per unit area is 30 N/cm}^2\text{).}

For each hand position the maximal lifting force at the hand is estimated by increasing the external force at the hand until no solution could be found using the optimizing procedure, i.e. maximal muscle force was exceeded.

IV. Results

Fig. 6 shows a sequence of circles indicating the hand position as a result of the motion of the scapula from position one to seven and the initial fusion position, projected to the transversal plane. Inside the circles the maximal force which can be exerted is printed. Maximal force can be exerted at position 6 of the scapula. Close to the body maximal force is less than further ventral, though the moment arm with respect to e.g. the sternoclavicular joint is smaller, resulting in a smaller external moment.

Figs 7A and 7B show the effect of changing the endorotation angle of the arthrodesis. It can be seen that the initial endorotation angle of 60.2 degrees results in a mobility area in the midsagittal

![Figure 7: Effect of changing the endorotation fusion angles (in deg.) on the mobility area of the hand (A: Transversal plane; B: Sagittal plane), elbow flexed 90 degrees.](image-url)
plane of the body. More endorotation (e.g. 90 degrees) is prevented by the trunk colliding with the forearm and the hand. Less endorotation results in a mobility area aside of the midplane. Working height increases with decreasing endorotation angle. The initial endorotation angle of 60.2 degrees results in a working height ranging from about table height to halfway the chin. Figs 8A and 8B show the effect of changing the abduction angle. Decreasing the abduction angle is prevented by the trunk colliding to the elbow. An increase in abduction results in a shift of the mobility area to

Figure 8: Effect of changing the abduction fusion angles on the mobility area of the hand (A: Transversal plane; B: Sagittal plane), elbow flexed 90 degrees.

Figure 9: Effect of changing the anteflexion fusion angles on the mobility area of the hand (A: Transversal plane; B: Sagittal plane), elbow flexed 90 degrees.
Table 1A: Maximal lifting force (in N) for different scapular positions and fusion angles: Change of endorotation angle.

<table>
<thead>
<tr>
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<th>scapula position 3</th>
<th>scapula position 4</th>
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Table 1B: Maximal lifting force (in N) for different scapular positions and fusion angles: Change of abduction angle.

<table>
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Table 1C: Maximal lifting force (in N) for different scapular positions and fusion angles: Change of anteflexion angle.

<table>
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<tr>
<th>anteflexion angle</th>
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<td>58</td>
</tr>
</tbody>
</table>

the side, which is not favourable. Figs 9A and 9B show the effect of changing the anteflexion angle. The mobility area remains in the midsagittal plane, but it shifts to the front and upwards. Tables 1A, 1B and 1C show the maximal lifting force as can be exerted at the hand, for a range of endorotation, abduction and anteflexion angles, respectively. The largest forces are obtained at position 6 of the scapula, almost irrespectively of the fusion angles. Due to the large moment arm
and external force, the largest external moments around the frontal axis of the SC-joint are also found in position 6 of the scapula.

Forces of the five thoracoscapular muscles during maximal force exertion are shown in Figs 10A to 10F, for the average fusion angles. In the model, muscles are partitioned into a number of elements according to the architecture in order to adequately represent the mechanical effect of the complete muscle (Van der Helm & Veenbaas, 1991). This partitioning allows an analysis of the activity of independent parts of the muscle. Major contributions come from the lower part of m. serratus anterior (elements 1 to 4) and m. trapezius, pars ascendens (elements 1, 2 and 3, Fig. 10A) and descendens (element 5 and 6). It is striking that the m. trapezius, pars transversalis (element 4 and 5) is not very active. M. rhomboideus and m. levator scapulae act mainly in order to stabilize the medial border of the scapula on the trunk.

V Discussion

One of the motives for performing a glenohumeral arthrodesis in patients with a brachial plexus lesion is function restoration. Instead of an arm dangling aside of the body which often needs support to prevent joint subluxation and which has sometimes a still functioning hand which is useless due to the lack of a firm base, a glenohumeral arthrodesis can provide a patient with some control over the position of the upper arm. An orthosis can stabilize the elbow and wrist position.

![Figure 10: Muscle activity during scapular motion with the mean fusion angles.](image)

A: M. trapezius, scapular part; B: M. trapezius, clavicular part; C: M. levator scapulae; D: M. pectoralis minor; E: M. rhomboideus; F: M. serratus anterior.
Glenohumeral arthrodesis (Cool, 1989). However, the irreversible destruction of the joint and the uncertain result of the operation impede the patient’s decision whether or not to undergo this operation (Pronk, 1991). Therefore, a reliable prediction of the positive and negative effects of the operation is desired. In comparing the function of the shoulder mechanism after a glenohumeral arthrodesis with a normal shoulder, a few important differences have to be taken into account. Assuming the sterno- and acromioclavicular joints to act as spherical joints and considering the constraints of the scapulothoracic gliding plane, the shoulder girdle has four DOF. In a kinematic analysis the axial rotation of the clavícula is not affecting the position of the scapula. Hence, just three DOF remain for positioning and orienting the scapula. If a patient wants to position his/her hand at a certain spot, the remaining three DOF of the scapula are just sufficient for positioning the hand: no redundancy is left. If a subject with a normal functioning shoulder is positioning his/her arm, all three DOF of the scapula can be used to seek for the most favourable position of the scapula. The scapula moves to an optimal position dependent on the position and external load on the humerus, considering the moment arms of the muscles and the force-length (and eventually the force-velocity) relationship for each muscle. After a glenohumeral arthrodesis no DOF are left in the system to shift the scapula to an optimal position. Thus, only one choice is left to optimize the scapular position: The choice of the fusion angles, which are chosen once and for ever.

The size of the mobility area of the hand is not affected by changing the fusion angles. As shown in Figs 6, 7 and 8, only the position of the mobility area is changed. It is remarkable that the initial fusion angles, obtained by combining data from a cadaver, patients and normal subjects, result in a quite well positioned mobility area. However, it does not mean that all patients observed by Pronk (1991) had a satisfying fusion position. One patient with too little endorotation held his hand out like a beggar. Nine patients had a strongly medially rotated scapula due to a large abduction angle. Four of them had so much abduction that they could not bring their elbow against the body. Almost all patients had a winging scapula (scapula alata) caused by the anteflexion angle. Some of these inconveniences are inevitable. Nevertheless, the mobility area obtained by normal scapular motions, as shown in Figs 6, 7 and 8, is very important. It can be assumed that the thoracoscapular muscles act at their normal physiological length, and their location with respect to each other is optimal for stabilizing the scapula. If the patient wants to reach outside this mobility area, rare positions of the scapula or additional motions of the trunk are necessary.

Besides the mobility area the force exerted by the hand is important for the patient. In Tables 1A, 1B and 1C it is shown that maximal external force is not exerted when the hand is close to the body, as one would expect, but in scapular position 6. For the average fusion position the moment
Figure 11: Moment balance during scapular motion with the mean fusion angles.
1: m. trapezius, scapular part; 2: m. trapezius, clavicular part; 3: m. levator scapulae; 4: m. pectoralis minor; 5: m. rhomboideus; 6: m. serratus anterior;
balance around the joint axes of the sternoclavicular and acromioclavicular joints is shown in Figs 11A to 11F. The external moments around the frontal (X-)axis of the SC-joint in position 1 and 2 of the scapula are small, since the moment arm is small (the hand is close to the body) and the external force is small. The external moment is increasing for the subsequent scapular positions, moving the hand away from the body. For an explanation of this phenomenon it is analysed which forces deliver the external moment and to what extent. Especially, m. serratus anterior and m. trapezius are contributing to the external moment around this axis (Figs 11A and 11D). In the initial position m. rhomboideus is the limiting factor, since it is maximally active in pressing AI to the thorax. Due to the laterally rotating and protracting motion pattern of the scapula, resulting in a position of AI lateral to the thorax, the muscle fibers of m. serratus anterior are oriented more in a ventral-dorsal direction. Hence, a larger ventrally directed force at AI can be exerted so that m. rhomboideus is no longer the limiting factor. In addition, a larger contribution to the maximal moment around the frontal axis is established. M. trapezius, pars transversalis, is hardly contributing to the moment around the x-axis, because its fibers are oriented parallel to this axis, resulting in a very small moment arm.

A similar force analysis is performed for maximal pressing forces of the hand. The most important muscle for this force direction is m. pectoralis minor. Because the force patterns do not show much differences between subsequent scapular positions, due to the fact that the moment arm of m. pectoralis minor barely changes, it is not an item of consideration for the glenohumeral arthrodesis. It depends on the Activities of the Daily Living (ADL) which fusion ages are optimal for an individual patient. Using the external fixator the fusion ages can be adjusted to the patient’s demands. If the fusion angles are adjusted, the position of the hand should be considered when the scapula is in a laterally rotated and protracted position.

VI Concluding remarks
1) In order to obtain a mobility area of the hand in the midsagittal plane, the humerus should be endorotated about 60 degrees and a little abducted and anteflexed.
2) The position where the maximal force can be exerted is roughly 10 cm in front of the initial hand position. In that position the scapula is laterally rotated besides the thorax and m. serratus anterior gets a large moment arm around the frontal axis.
3) The optimal hand position can not be set during the arthrodesing operation, because due to the patient’s horizontal position the scapular position is changed.
4) It is recommended to use an external fixation technique for the glenohumeral arthrodesis. Using
the external fixator optimal fusion angles can be set outside the operation theater.

5) Since the scapula has only 3 degrees of freedom, positioning the hand after a glenohumeral arthrodesis implies that no redundancy is left for the scapula to seek for an optimal position.
Chapter 9

The use of a musculoskeletal model of the shoulder mechanism in wheelchair propulsion.

F.C.T. van der Helm\textsuperscript{1}, H.E.J. Veeger\textsuperscript{2}.

\textsuperscript{1} Man-machine Systems Group, Lab. for Measurement and Control, Dept. of Mechanical Engineering and Marine Technology, Delft University of Technology, Delft, The Netherlands.

\textsuperscript{2} Fac. of Human Movement Sciences, Free University Amsterdam, The Netherlands.

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Abstract

During wheelchair propulsion the largest net joint moments and net joint powers are generated at the shoulder. In order to determine the muscles which contribute to the joint moments, at present a musculoskeletal model of the shoulder mechanism is necessary. Since for the time being the three-dimensional position of the shoulder girdle can only be recorded in static situations with help of a palpation technique, in this study the use of a musculoskeletal model was verified in a static experiment on a wheelchair ergometer. Positions of the trunk, shoulder girdle and upper extremity were recorded simultaneously with forces on the rim and surface EMG of ten muscles. Four healthy male subjects participated in the experiment. Five hand positions on the rim and five different load levels per hand position were measured for each subject. A musculoskeletal model of the shoulder mechanism was used to calculate muscle forces in an inverse dynamic simulation. EMG and muscle force comparison showed that model calculations are reliable. In addition, the moment balance around each joint axis of the shoulder mechanism is discussed. Results of the experiment will be used for future expansion of the experimental set-up to dynamic situations. It is concluded that the musculoskeletal model of the shoulder mechanism can be very useful in ergonomic studies to determine the mechanical load on morphological structures.
I Introduction

In the field of human movement studies and ergonomics very little is known about the mechanics of the shoulder mechanism. The shoulder mechanism consists of the thorax, clavicle, scapula and humerus. Since the motions of these bones are closely related, the shoulder mechanism ought to be studied as a whole. Usually the shoulder joint is modelled as a spherical joint between the upper arm and the trunk, neglecting the bones of the shoulder girdle, i.e. the scapula and clavicle (Chaffin & Andersson, 1984; Dul, 1988). Consequently, conclusions drawn from this modelling approach are limited. Assessment of admissible load on the shoulder mechanism is not precise and no relation can be established between complaints and load applied.

Recently, a detailed musculoskeletal model of the shoulder mechanism has been developed (Van der Helm, 1991a). In this model bones of the shoulder girdle as well as the thoracoscapular muscles are incorporated. The model has been used for a biomechanical analysis of muscle functions during abduction and anteflexion (Van der Helm, 1991b) and to predict the outcome of operations, e.g. the effect of fusion angles in a glenohumeral arthrodesis on function restoration (Van der Helm & Pronk, 1991b). In the present study it is investigated whether the shoulder mechanism model can be applied in complex ergonomic situations in which a wide variety of force directions and shoulder girdle positions occurs.

One of the very interesting areas of ergonomics of the shoulder mechanism is the wheelchair-user interface. Cerquiglini et al. (1981) discovered that most power for wheelchair propulsion was generated around the shoulder joint. An important factor in the design of wheelchair propulsion mechanisms should be the human ability to perform this non-natural type of arm work. Woude (1989) proposed a combined physiological and biomechanical approach to evaluate the wheelchair-user interface. He used external power output and gross mechanical efficiency as variables to estimate the effect of propulsion mechanism, rim diameter, seat position and cycle frequency. Recently, a segment based biomechanical model was presented (Veeger et al., 1991b) for analysis of arm positions and net joint moments in wheelchair propulsion.

Gross mechanical efficiency (the quotient between power output used for propulsion and total power output) is lower for wheelchair propulsion than for arm cranking (Veeger et al., 1991b). Woude (1989) proposed the following possible causes for this low mechanical efficiency: Discontinuity of force application, relatively large recovery phase, synchronous force application, trunk movements, the need for high muscular forces and the muscular effort needed for stabilization of the shoulder girdle.

Thus far, only net joint torques around the idealized shoulder joint between the trunk and upper
arm have been calculated which are insufficient to gain more insight into the complex function of the shoulder mechanism (Cerquiglini et al., 1981; Veeger et al., 1991b). During humeral motions the scapula slides and rotates over the so-called scapulothoracic gliding plane. It provides a moveable but stable base for motions of the upper extremity. External forces are transmitted by the shoulder girdle bones and thoracoscapular muscles to the trunk. Factors determining power development need to be analyzed on the level of individual muscles. In order to generate power in the shoulder mechanism much effort is needed to stabilize the shoulder girdle. Muscles can only generate positive work when they shorten while exerting force. During some movements a contradictory situation can appear when e.g. a joint is extending while a flexing moment is generated (Ingen Schenau, 1989; Veeger et al., 1991b), Fig. 1. Mono-articular muscles would lengthen while contracting in order to generate the required moment. Hence, these muscles would dissipate energy which is generated around more proximal joints. Ingen Schenau et al. (1990) showed that bi-articular muscles are capable to exert a moment around one joint while shortening due to motion of the other joint. In the shoulder mechanism many poly-articular muscles are present whereas the external force can be directed almost anywhere. However, which muscle is most important in power generation is yet unknown.

A musculoskeletal model of the shoulder mechanism (Van der Helm, 1991a) will be used to generate answers to these questions. Input variables for the model are position, velocity and acceleration of the bones and external load, output variables are muscle forces calculated by using an optimization criterion. Model predictions must be verified in a complex situation as wheelchair

![Diagram](https://example.com/diagram.png)

Figure 1: Joint B is extending ($\dot{\phi}_b$) while a flexing moment ($M_b$) is needed for the desired external force direction. Bi-articular muscle II can provide the flexing moment while it is shortening due to the motion at joint A ($\dot{\phi}_a$). Hence, muscle II will contribute to the external power generated.
propulsion. However, main problem for dynamic simulations of the shoulder mechanism is the recording of input variables, i.e. the motions of the clavicle and scapula. Therefore, aim of the present study is verification of the model in a static experiment in which a complete set of position recordings, including the position of clavicle and scapula, could be obtained using a palpation technique. Muscle forces predicted by simulation are compared with EMG recordings. In addition, the mechanics of the shoulder mechanism during quasi-static wheelchair propulsion can be analysed. It will be discussed to what extent the results of this study, i.e. a verified model of the shoulder mechanism and recordings of the scapular and clavicular positions, can be extrapolated to a dynamic situation which is the only realistic situation for wheelchair propulsion.

II Method

II.1 Subjects

Four healthy male subjects without prior shoulder complaints participated in the experiment. All subjects gave written informed consent. Relevant data, including some anthropometric measurements, are presented in Table 1. Similar antropometric data are presented for a male cadaver of which the morphology has been used for the musculoskeletal model. This cadaver was considered to be a more or less median cadaver in an extensive cadaver study consisting of seven cadavers (Veeger et al., 1991\textsuperscript{a}; Van der Helm et al., 1991).

II.2 Protocol

Subjects were seated on a stationary wheelchair ergometer (Niesing et al., 1990). The ergometer allowed for on-line measurement of torque and velocity on both left and right wheel up to a sample frequency of 100 Hz. Seat position was adjusted in order to obtain a specified elbow angle (100 degrees) when the hand was in top position on the rim. Rim diameter was 52 cm. Five hand positions are discerned (Fig. 2): Position 1 - 15 degrees before top position, position 2 - top position, position 3 - 15 degrees after top position, position 4 - 30 degrees after top position, position 5 - 60 degrees after top position. These hand positions describe approximately the whole trajectory during the push phase of wheelchair propulsion. Order of hand positions was randomized. In a separate session on a preceding day, maximal voluntary moments (MVM) had been recorded for each hand position, with intervals of 20 minutes. Five load levels were imposed in each hand position: zero load, approximately 10% MVM, 20% MVM, 30% MVM and 40% MVM. For each subject a total of 25 sessions was planned. Between sessions a 5 minutes break was allowed. Load levels were imposed by adjusting the virtual slope on the wheelchair ergometer.
Table 1: Characteristic of subjects participating in the experiment and the cadaver used for the model parameters. Mmax: Maximal one-sided moment around the wheel axis.

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<th>Weight (kg)</th>
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</table>

which would result in a backward turning of the wheels. Subjects were asked to keep the wheels steady with their hands in the specified position. For higher load levels the virtual slope was built up gradually.

Total recording time during each session was maximally 40 seconds. For each hand position the subject was free to seek an adequate trunk position to exert force. Since the palpation technique requires that the subject does not move, the subject’s position was subsequently standardized by a pointer touching the sternum. For every load level at the same hand position the position of this pointer was identical.

Figure 2: Stationary wheelchair ergometer. Hand positions 1 to 5 are indicated on the rim, which can be seen in the mirror.
II.3 Force measurement

Direction and magnitude of force applied to the right rim and the moment exerted around the right wheel axis were sampled independently at a sample frequency of 50 Hz. On the onset of sampling a timing signal was sent to an available tape recorder channel (TEAC SR-70). Accuracy of the force signal is 4.5 N (standard deviation) over a range of 400 N in all three directions, and accuracy of the moment recordings is 0.5 Nm (standard deviation) over a range of 100 Nm.

II.4 EMG measurement

For ten muscles of the shoulder mechanism surface EMG recordings were collected (Sentry 1000 electrodes, DISA 15C01 amplifier; bandwidth 10-500 Hz) and stored on tape (TEAC SR-70). Table 2 shows the muscles involved and electrode location. Electrode center-to-center distance was 4 cm. Recordings were off-line band filtered and AD converted (filter 10-200 Hz, sample frequency 500 Hz). After rectification, EMG’s were averaged over a 10 second interval which embraced the instant of camera recordings.

II.5 Position recording

Two types of position recording methods were used simultaneously: Photocameras were used for recording 3D coordinates of markers glued to the skin and the palpator, a spatial digitizer, was used for recording the position of scapula and clavicle underneath the skin.

Two photocameras and one mirror were used to provide three views of the subject. Both cameras (Canon T-90) were operated with one trigger. All three views were calibrated using the DLT method (Marzan & Karara, 1975), resulting in a mean prediction error of 1.2 mm per coordinate (standard deviation). Table 3 shows which bony landmarks were marked with markers glued to the skin. In addition, to both upper arm and forearm a manchet equipped with at least 7 markers was firmly attached. The positions of these manchet with respect to markers on bony landmarks were calibrated at the start of the experiment.

Positions of scapula and clavicle were recorded using a palpation technique (Pronk, 1987; Van der Helm & Pronk, 1991). Four bony landmarks on the scapula, i.e. most dorsal point on the acromioclavicular joint (AC), angulus acromialis (AA), trigonum spinae (TS) and angulus inferior (AI), and two landmarks on the spine, i.e. the spine of 7th cervical vertebra (C7) and 8th thoracic vertebra (T8), were digitized using the palpator (Pronk & Van der Helm, 1991). Mean distance between these bony landmarks was used to check on-line whether or not bony landmarks were palpated correctly, using rigid body assumptions. This procedure resulted in a measurement error
Table 2: Electrode locations at the muscles involved in EMG recordings.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Electrode location</th>
</tr>
</thead>
<tbody>
<tr>
<td>M. biceps</td>
<td>In the midst of the muscle belly.</td>
</tr>
<tr>
<td>M. deltoideus, p. anterior</td>
<td>2 cm below the anterior rim of the acromion.</td>
</tr>
<tr>
<td>M. deltoideus, p. medialis</td>
<td>2 cm below the lateral rim of the acromion.</td>
</tr>
<tr>
<td>M. infraspinatus</td>
<td>In the midst between Trigonum Spinae and Angulus Inferior, 2 cm of the medial border of the scapula.</td>
</tr>
<tr>
<td>M. pect. major, p. clavic.</td>
<td>In the midst between sternoclavicular joint and processus coracoideus, 2 cm below the clavica.</td>
</tr>
<tr>
<td>M. pect. major, p. thor.</td>
<td>6 cm above the nipple.</td>
</tr>
<tr>
<td>M. trapezius, p. ascendens</td>
<td>2/3 on the line between Trigonum Spinae and 8th thoracic vertebra, 4 cm from the edge of the muscle.</td>
</tr>
<tr>
<td>M. trapezius, p. descend.</td>
<td>In a vertical line above Trigonum Spinae, 2 cm below the edge of the muscle</td>
</tr>
<tr>
<td>M. trapezius, p. transv.</td>
<td>1 cm medial and 4 cm cranial to TS.</td>
</tr>
<tr>
<td>M. triceps, c. longum</td>
<td>2 cm below the edge of m. deltoideus posterior.</td>
</tr>
</tbody>
</table>

for each bony landmark of 3 - 5 mm standard deviation in a previous experiment (Van der Helm & Pronk, 1991). Accuracy of the palpation technique is comparable with accepted standards for cinecamera techniques, since in the latter method skin displacements reduce the accuracy of bony landmarks reconstruction (Capozzo et al., 1988).

Measurement error of the palpator is 0.96 mm standard deviation per coordinate. The coordinate system of the palpator was fitted to the coordinate system of the photocameras using a calibration frame containing nine markers and the help of an algorithm obtained from Veldpaus et al. (1988).

III Data Processing

III.1 Dynamic model of the shoulder mechanism

An extensive description of the shoulder mechanism model has been presented elsewhere (Van der Helm, 1991). For modelling the shoulder mechanism a finite element method is used which is implemented in the computer program SPACAR (Werff, 1977; Werff & Jonker, 1983; Jonker, 1988). This finite element method is specially developed for multi-degree-of-freedom spatial mechanisms with flexible bodies. In this application morphological structures are represented by
### Table 3: Bony landmarks used to reconstruct the local coordinate systems of the bones.

<table>
<thead>
<tr>
<th>Bone</th>
<th>Bony landmark</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Incisura Jugularis (IJ)</td>
</tr>
<tr>
<td></td>
<td>Processus Xiphoideus (PX)</td>
</tr>
<tr>
<td></td>
<td>Cervical vertebra 7 (C7), palpated.</td>
</tr>
<tr>
<td></td>
<td>Thoracic vertebra 8 (T8), palpated.</td>
</tr>
<tr>
<td>Clavicula</td>
<td>Incisura Jugularis (IJ)</td>
</tr>
<tr>
<td></td>
<td>Most dorsal point on acromioclavicular joint (AC), palpated.</td>
</tr>
<tr>
<td>Scapula</td>
<td>Most dorsal point on acromioclavicular joint (AC), palpated.</td>
</tr>
<tr>
<td></td>
<td>Angulus Acromialis (AA), palpated.</td>
</tr>
<tr>
<td></td>
<td>Trigonum Spinae (TS), palpated.</td>
</tr>
<tr>
<td></td>
<td>Angulus Inferior (AI), palpated.</td>
</tr>
<tr>
<td>Humerus</td>
<td>Gap between acromion and humerus (AH)</td>
</tr>
<tr>
<td></td>
<td>Epicondylus Medialis (EM)</td>
</tr>
<tr>
<td></td>
<td>Epicondylus Lateralis (EL)</td>
</tr>
<tr>
<td></td>
<td>Upperarm manchet</td>
</tr>
<tr>
<td>Forearm</td>
<td>Epicondylus Medialis (EM)</td>
</tr>
<tr>
<td></td>
<td>Epicondylus Lateralis (EL)</td>
</tr>
<tr>
<td></td>
<td>Processus Styloideus Radialis (SR)</td>
</tr>
<tr>
<td></td>
<td>Processus Styloideus Ulnaris (SU)</td>
</tr>
<tr>
<td></td>
<td>Forearm manchet</td>
</tr>
<tr>
<td>Hand</td>
<td>Processus Styloideus Radialis (SR)</td>
</tr>
<tr>
<td></td>
<td>Processus Styloideus Ulnaris (SU)</td>
</tr>
<tr>
<td></td>
<td>Metacarpale II (MII)</td>
</tr>
</tbody>
</table>

Elements of which the kinematic and dynamic behavior is well-known. Then, the behavior of the complete mechanism can be calculated by simply connecting the elements. In the finite element method bones are represented by rigid BEAM elements: The clavicula by one BEAM between the sternoclavicular and acromioclavicular joint; the scapula by two BEAM elements, between the acromioclavicular joint and trignon spinae (TS) and between TS and angulus inferior (AI), respectively; the humerus by one BEAM from the glenohumeral joint to the elbow joint; the forearm bones by one BEAM from the elbow joint to the wrist and the hand by one BEAM from the wrist to the point of force application at the second metacarpal (MII). All joints, i.e. sternoclavicular, acromioclavicular, glenohumeral, elbow and wrist joint are represented as spherical joints by three orthogonal HINGE joints allowing all possible rotations. The elbow and wrist joint have been represented as spherical joints, because the axial rotation of the proximal
bone could not be assessed sufficiently accurate to allow description of the rotations of the distal bone around two rotation axes fixed in the proximal bone. Ligaments are represented by flexible TRUSS elements. The medial border of the scapula is pressed against the thoracic wall by the combined action of the m. serratus anterior and m. rhomboideus. The motion constraints of the so-called scapulothoracic gliding plane are represented by two SURFACE elements, each constraining one point of the medial border to slide over the surface of the thorax, modelled as an ellipsoid. Muscles are represented by one or more force generating 'active' TRUSS or CURVED-TRUSS elements, of which the latter are wrapped around the surface of bony contours (Van der Helm, 1991c). Each muscle element represents a muscle line of action, generating force at its attachments on the bone. Generally, more than one muscle element is necessary for representing the mechanical effect of muscles with large attachment sites and complex muscle architectures (Van der Helm & Veenbaas, 1991). In total, 95 muscle elements are used in the model to represent 16 muscles. Morphological data, including the geometry of bones and muscles, muscle physiological cross-sectional area (PCSA) and the inertia tensor of the segments, were recorded in a cadaver study (Veeger et al., 1991a; van der Helm et al., 1991; Van der Helm & Veenbaas, 1991). In Table 1 the characteristics of the cadaver used in this simulation is presented. It was a median cadaver in the cadaver population. The performance of 4 young subjects which were above average compared with a larger subject population (Veeger et al., 1991b), is simulated with the morphology of an older man. Size and weight were quite different and pilot simulations revealed that the external force as recorded for the subjects could not be exerted with muscle PCSA as recorded in the cadaver experiment. Therefore, it was decided to scale muscle PCSA by enlarging the force per unit area from 30 N/cm² to 100 N/cm².

Input variables of the model are the position (and eventually velocity and acceleration) of the bones and external forces, output variables are muscle forces subject to an optimization criterion. Van der Helm (1991a) showed that the results of a number of non-linear criteria were almost similar. In this study minimization of the sum of squared muscle stresses has been chosen.

Position of the shoulder mechanism and upper extremity is determined by independent generalized coordinates comparable to the Lagrangian approach. The number of generalized coordinates is equal to the number of degrees of freedom (DOF). In the present model there are twelve DOF: three at the shoulder girdle, three at the glenohumeral joint, three at the elbow and three at the wrist joint. The shoulder girdle is a closed-chain mechanism. The sternoclavicular and acromioclavicular joints yield six DOF. Two DOF are restricted by the scapulothoracic gliding plane, resulting in a total number of four DOF. However, the axial rotation of the clavicle could
not be measured, since only two bony landmarks were available (Table 3). Therefore, the number of DOF is reduced by one by assuming a rigid conoid ligament which only affects the axial rotation of the clavicle. Then, three DOF remain at the shoulder girdle. Generally, for musculoskeletal systems rotations of the joints are used as generalized coordinates. However, due to the constraints imposed by the clavicle and thorax, orientation of the scapula as input variable would result in strange positions on the thorax. Therefore, the y- and z-coordinate of AC and x-coordinate of AI have been chosen as generalized coordinates. For the glenohumeral, elbow and wrist joints Euler angles are used as generalized coordinates.

### III.2 Input variables

In order to simplify model computations, the thorax is fixed in a global coordinate system. Incisura jugularis (IJ) is the origin of this coordinate system and the X-axis points from medial to lateral, Y-axis from caudal to cranial and Z-axis from ventral to dorsal. Only right shoulders were measured. All position coordinates are presented with respect to this global coordinate system. Position data recorded in each session have been transformed to the global coordinate system using a transformation function obtained from fitting the recorded thorax position in the same session to the mean thorax position in the global coordinate system.

The local coordinate system of the thorax is defined as follows:

**Thorax (T):**
- **y-axis:** pointing from processus xiphoideus (PX) to IJ.
- **x-axis:** perpendicular to the plane through C7, IJ and PX, pointing laterally.
- **z-axis:** perpendicular to the x- and y-axis.

Mean orientation of the thorax ($T_0$) with respect to the global coordinate system for ten subjects has been derived in a previous motion recording study (Van der Helm & Pronk, 1991)

\[
T_0 = \begin{bmatrix}
0.9954 & -0.0051 & 0.0350 \\
0.0162 & 0.9457 & -0.3247 \\
-0.0315 & 0.3251 & 0.9452
\end{bmatrix}
\]
Wheelchair propulsion

For each session i the orientation of the local coordinate system $T_i$ was calculated. Then all the vectors $p_j$ containing position coordinates in the measurement coordinate system were transformed to vectors $a_j$ in the global coordinate system according to:

$$a_j = R_i . (p_j - U) ;$$

$$R_i = T_0 . T_i^T ,$$

where the superscript $^T$ means 'transposed' and $U$ is the position vector of bony landmark $U$, recorded in the measurement coordinate system.

The y- and z-coordinate of AC and x-coordinate of AI, calculated in the global coordinate system, have been used as generalized coordinates to determine the position of the scapula and clavicle.

The local coordinate systems of the upper arm (U), forearm (F) and hand (H) are defined as follows:

**Upper arm (U):** y-axis: Pointing from the gap between acromion and humerus (AH) to the midpoint between epicondylus medialis (EM) and lateralis (EL).

z-axis: Perpendicular to the plane determined by AH, EM and EL, pointing ventrally.

x-axis: Perpendicular to the y- and z-axis.

**Forearm (F):**

y-axis: Along the line from the midpoint between EM and EL to the midpoint between processus styloideus radialis (SR) and ulnaris (SU).

z-axis: Perpendicular to the plane determined by the midpoint between EM and EL and the landmarks SR and SU, pointing ventrally.

x-axis: Perpendicular to the y- and z-axis.

**Hand (H):**

y-axis: Along the line from SR to the distal end of second metacarpal (MII).

z-axis: Perpendicular to the plane through SR, SU and MII, pointing ventrally.

x-axis: Perpendicular to the y- and z-axis.

Since often markers, glued to bony landmarks, are obscured from at least two camera views, some additional calculations are necessary in order to obtain the local coordinate systems $U_j$ and $F_i$. In the calibration position in which all markers could be recorded, the local coordinate system $U_0$ was determined as well as the position vectors $b_j$ of the markers on the upper arm manchet ($j = 1, \ldots, N$; $N$ is the number of markers on the manchet). The manchet has been measured again in recording
session i and the position vectors $q_i$ of the markers are derived in accordance to Eq. (2). The relation between vectors $b_i$ and $q_i$ is determined by

$$q_i = a_0 + v_i + R_i (b_i - a_0) ;$$

where $a_0$ is the centroid of the markers on the manchet in the calibration position, $v_i$ is the translation vector and $R_i$ is the rotation matrix. $R_i$ and $v_i$ are estimated using a least squares criterion (Veldpaus et al., 1988). Then, the orientation of local coordinate system $U_i$ of the upper arm in session i can be calculated:

$$U_i = R_i U_0 .$$

Local coordinate system $F_i$ is calculated analogously.

Simulations started from the anatomical position of the arm. Orientations of the local coordinate systems $U_a$, $F_a$ and $H_a$ in the anatomical position are defined by:

$$U_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & -1 & 0 \\ 0 & 0 & -1 \end{bmatrix} ; \quad F_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & -1 & 0 \\ 0 & 0 & -1 \end{bmatrix} ; \quad H_a = \begin{bmatrix} 1 & 0 & 0 \\ 0 & -1 & 0 \\ 0 & 0 & -1 \end{bmatrix} .$$

Then, rotations in the glenohumeral joint are defined with respect to the anatomical position (Van der Helm & Pronk, 1991a):

$$U_i = R_a U_a = R_a U_i U_a^T ;$$

Rotation matrix $R_u$ is decomposed into Euler angles around the global Y-axis (pole angle), local z'-axis (elevation angle) and local y''-axis (axial rotation):

$$R_u = R_y(\beta_u) R_z(\gamma_u) R_y(\beta'_u) =
\begin{bmatrix}
\cos(\gamma_u) & 0 & -\sin(\gamma_u) \\
0 & 1 & 0 \\
-\sin(\gamma_u) & 0 & \cos(\gamma_u)
\end{bmatrix}
\begin{bmatrix}
\cos(\beta_u) & 0 & \sin(\beta_u) \\
0 & 1 & 0 \\
-\sin(\beta_u) & 0 & \cos(\beta_u)
\end{bmatrix} .$$
Wheelchair propulsion

from which $\beta$, $\gamma$ and $\beta'$ can be calculated. Single and double quotes denote transformation of the rotation axes by one or two prior rotations, respectively (Paul, 1983). Calculated Euler angles are used as input variables to the model. For the elbow and wrist, the Euler angles can be calculated analogously.

For the elbow rotations:

$$F_i = R_u R_f F_a \rightarrow R_f = R_u^T F_i F_a^T;$$

and rotation matrix $R_f$ is decomposed into Euler angles around subsequently the global X-axis (flexion/extension), the local $y'$-axis (pro/supination) and the local $z''$-axis:

$$R_f = R_x(\alpha_f) R_y(\beta_f) R_z(\gamma_f).$$

For wrist rotations, rotation matrix $R_h$ is decomposed into Euler angles subsequently around the global X-axis (flexion/extension), local $z'$-axis (ulnar/radial abduction) and local $y''$-axis:

$$H_i = R_u R_f R_h H_a \rightarrow R_h = R_f^T R_u^T H_i H_a^T;$$

$$R_h = R_x(\alpha_h) R_z(\gamma_h) R_y(\beta_h).$$

**Force input**

Direction and magnitude of the force exerted by the hand on the rim were recorded by three force transducers at the suspension of the wheel. In addition, the moment around the wheel axis was recorded independently by strain gauges at the wheel axle. From the force at the hand and the hand position the moment around the wheel axis was calculated. This calculated moment was always somewhat different from the recorded moment. The difference is likely due to a moment exerted by the hand onto the rim. This hand moment was calculated for each position. Thus, the hand is exerting both a moment and a force at the rim. Force and moment vector were transformed to the global coordinate system attached to the thorax, and included in the model, with metacarpal II (MII) as point of application. Moreover, external force vectors due to gravitational forces at the upper- and forearm were similarly transformed and included in the model, with the center of gravity of the segment as point of application. Hence, in all subsequent force calculations gravitational forces are not strictly along the global Y-axis.
IV Results
Due to the complexity of simultaneously recording EMG, force and positions, data are incomplete for a number of sessions. Table 4 summarizes available data from EMG, position and force recordings for each condition. A total number of 79 out of the maximal 100 sessions were completely recorded for all four subjects.

IV.1 Kinematics
In this study only static positions have been recorded due to the obvious complications of recording the position of scapula and clavicle. Hence, in this study the word 'motion' refers to a quasistatic motion: The difference between one position and the next. Still, it is interesting to compare a sequence of static positions with positions obtained in a dynamic recording of a propulsion cycle using high-speed film cameras (Veeger et al., 1991d), see Figs 3A and 3B. Only the positions of the trunk and arm segments are shown, because the position of the shoulder girdle can not be measured in dynamic situations. The similarity between static and dynamic positions of the trunk and arm suggests that in dynamic situations the positions of scapula and clavicle may be extracted from assumptions based on these static measurements.

Main thorax rotation was around the frontal (X-)axis (Figs 4A to 4D). Other rotations were negligible since a symmetric two-sided motion was required. Thorax position was quite different for the subjects. One subject (FK) had a large but steady forward rotation during all hand positions.

**Figure 3**: Stick diagrams showing the position of the upper extremity during wheelchair propulsion. In addition, the propulsion force vector at the rim is shown.
A: Static experiment.
B: Dynamic experiment.
Table 4: Number of sessions for each condition in which EMG, motions (Mot) and Force (For) recordings were successful. Number of subjects was four.

<table>
<thead>
<tr>
<th></th>
<th>position 1</th>
<th>position 2</th>
<th>position 3</th>
<th>position 4</th>
<th>position 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>0% MVM</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 3</td>
<td>EMG: 1</td>
</tr>
<tr>
<td></td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 3</td>
<td>Mot: 2</td>
</tr>
<tr>
<td></td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 3</td>
<td>For: 2</td>
</tr>
<tr>
<td>10% MVM</td>
<td>EMG: 2</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 3</td>
<td>EMG: 3</td>
</tr>
<tr>
<td></td>
<td>Mot: 2</td>
<td>Mot: 3</td>
<td>Mot: 4</td>
<td>Mot: 3</td>
<td>Mot: 3</td>
</tr>
<tr>
<td></td>
<td>For: 2</td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 3</td>
<td>For: 3</td>
</tr>
<tr>
<td>20% MVM</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 2</td>
<td>EMG: 3</td>
</tr>
<tr>
<td></td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 1</td>
<td>Mot: 3</td>
</tr>
<tr>
<td></td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 2</td>
<td>For: 3</td>
</tr>
<tr>
<td>30% MVM</td>
<td>EMG: 3</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 3</td>
<td>EMG: 3</td>
</tr>
<tr>
<td></td>
<td>Mot: 3</td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 3</td>
<td>Mot: 2</td>
</tr>
<tr>
<td></td>
<td>For: 3</td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 3</td>
<td>For: 3</td>
</tr>
<tr>
<td>40% MVM</td>
<td>EMG: 3</td>
<td>EMG: 4</td>
<td>EMG: 4</td>
<td>EMG: 2</td>
<td>EMG: 3</td>
</tr>
<tr>
<td></td>
<td>Mot: 3</td>
<td>Mot: 4</td>
<td>Mot: 4</td>
<td>Mot: 2</td>
<td>Mot: 3</td>
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<tr>
<td></td>
<td>For: 3</td>
<td>For: 4</td>
<td>For: 4</td>
<td>For: 2</td>
<td>For: 3</td>
</tr>
</tbody>
</table>

Subjects EL and WV showed smaller thorax rotations, whereas subject EM seemed to show some effect of load level on thorax rotation. The motions of clavicula, scapula and humerus will be described with respect to a coordinate system fixed to the thorax.

Scapular orientation is described as a sequence of rotations with respect to a virtual reference position in which the spine (AC - TS) is along the frontal axis and the scapular plane (AC - TS - AI) parallel to the frontal plane. Rotations are described by three Euler angles, subsequently around the global Y-axis (pro/retraction), local z'-axis (medial/lateral rotation) and local x''-axis (tipping forward/backward); (Van der Helm & Pronk, 1991a). Figs 5A, 5B and 5C show mean rotations for each combination of hand position and load level. Using multiple regression, predictions of scapular orientation from thorax forward rotation and humeral orientation were highly significant but had a generally moderate predictive value (Veeger et al., 1991c). With respect to the mean rest position of the scapula as recorded in a previous study (Van der Helm & Pronk, 1991a), the scapula is slightly laterally rotated and retracted.

IV. 2 Inverse dynamics

Fig. 6 shows the motion of the upper extremity with respect to a fixed thorax for five subsequent
hand positions (subject EM, 20% MVM). Also shown are the propulsion force vectors exerted by the hand at the wheel. The largest external moments are around the frontal (X-) axis of the glenohumeral joint (Figs 7A to 7D). Due to the propulsion force mainly a negative (retroflexing) external moment results around the glenohumeral joint which is counterbalanced by muscles generating an anteflexing moment. The largest moment is in hand position 2 in which the force vector is nearly perpendicular to the line between the point of application (the hand) and the glenohumeral joint (Fig. 6). Near hand position 5 the moment approaches zero since the force vector points roughly towards the glenohumeral joint. Subject EL experienced even an anteflexing external moment, urging retroflexing muscles around the glenohumeral joint to become active. Since the rim position is more lateral than the rotation center of the GH-joint (about 18 cm), also external moments around the vertical (Y-) and sagittal (Z-) axis occur. External forces and moments (and inertial forces in case of dynamic situations) are counterbalanced by muscular and ligamental forces and moments. In the case of the shoulder girdle, also forces between thorax and scapula are important.

For ten muscles surface EMG was recorded (Table 2). EMG amplitude can be compared with forces of the most nearby muscle elements as calculated in the model simulations. However, since EMG-force relationship is unknown for shoulder muscles, and in addition EMG amplitude depends on muscle length (Heckathorne & Childress, 1981), some precautions are needed in the EMG-force

![Figure 4: Forward rotation (in degrees) of the thorax around the frontal axis for subject EM (A), FK (B), EL (C) and WV (D) for hand position 1 to 5 at the rim.

---: 0% MVM; -----: 10% MVM; ........: 20% MVM; -.-.-.-: 30% MVM; -o-o-o-: 40% MVM.]
comparison. Only on/off patterns in EMG and force should be compared (Van der Helm, 1991a). Since for a number of sessions data are missing, mean values depend on which subjects are available and show an irregular pattern. As a typical example, Figs 8A to 8J show EMG and force patterns in relation to load and hand position for subject FK. Increasing load normally results in increasing EMG activity, as would be expected. Since the net external moment around the frontal axis of the GH-joint is almost zero in hand position 5, most EMG and force patterns decrease towards this hand position. Most muscles have their major activity around hand position 2 in which the largest external moments occur. In comparison, muscle force and EMG patterns agree well for m. biceps, m. deltoideus, pars anterior, m. infraspinatus, m. pectoralis major, thoracic part, m. trapezius, pars ascendens, transversalis and descendens. For m. pectoralis major, clavicular part, and m. deltoideus, pars medialis, and m. triceps EMG activity has been recorded while no muscle force is predicted by the model.

Figs 9A to 9T show forces calculated for all active muscles, summed over all muscle elements, for subject FK. Since the main external moment decreases with hand position 1 to 5, all muscle forces tend to decrease accordingly. Despite differences in technique, resulting in a different position of thorax and upper extremity, muscles forces and moments were quite similar for all subjects. With exception of zero load level, different load levels did not change the muscle activation patterns, compared over five hand positions. As an example, for all hand positions at 20% MVM (subject EL), Figs 10A to 10C show the moment equilibrium around the X-, Y- and Z-axis of the SC-joint, respectively. Moments which are less than 1 Nm are omitted for

![Graphs](image)

**Figure 5:**
Mean rotations (in degrees) of the scapula for hand position 1 to 5 at the rim. A: Pro/retraction; B: Lateral rotation; C: Tipping forward/backward.
———: 0% MVM; ————: 10% MVM;
........: 20% MVM; -.--.-: 30% MVM.
0-0-0-0-0: 40% MVM.
Figure 6: Stick diagram showing the position of the upper extremity with respect to a fixed thorax for hand position 1 to 5 (subject EM, 20% MVM).

- - - - - : hand position 1; - - - - : hand position 2; - - - : hand position 3; - - - - : hand position 4; -o-o-o- : hand position 5.

Figure 7: Moments (in Nm) due to the propulsion force vector around the frontal axis of the glenohumeral joint for subject EM (A), FK (B), EL (C) and WV (D) for hand position 1 to 5 at the rim.

- - - - : 0% MVM; - - - - - : 10% MVM; - - - - - - : 20% MVM; - - - - - - : 30% MVM; -o-o-o- : 40% MVM.
Figure 8: EMG (in \%EMG_{max}) and force (in \%F_{max}) patterns for ten muscles for hand position 1 to 5 (subject FK).

- : 0\% MVM; - - - - : 10\% MVM; - - - - - : 20\% MVM; - - - - - - : 30\% MVM; - o - o - : 40\% MVM.
Figure 9: Force patterns (in N) for 20 muscles and muscle parts for hand position 1 to 5 (subject FK), summed over all muscle elements per muscle.

---: 0% MVM; ----: 10% MVM; ------: 20% MVM; ---: 30% MVM; -o-o-o-: 40% MVM.
Wheelchair propulsion

presentation purposes. The main external moment is around the X-axis, which is counterbalanced mainly by m. serratus anterior, aided by m. trapezius, scapular part and m. pectoralis, thoracic part. The reaction force of the thorax at Al is adding to the external moment. The moment balance around the Y-axis consists of m. serratus anterior and m. pectoralis major, thoracic part, on the one hand, and m. trapezius, scapular part, and the external moment on the other hand. When the propulsion force vector becomes more vertical towards hand position 5, the external moment around the Z-axis increases. Main counterbalancing moments are due to the weight of the arm and m. pectoralis major, thoracic part (Fig. 10C).

For the same load condition, the moment equilibrium of the AC-joint is shown in Figs 11A to 11C

Figure 10: Moment balance (in Nm) around the joint axes of the sternoclavicular joint. Only moments larger than 1 Nm are shown (subject EL, 20% MVM).
A: frontal (X)-axis; B: vertical (Y)-axis; C: sagittal (Z)-axis.
1: m. trapezius, scapular part; 2: m. trapezius, clavicular part; 3: m. levator scapulae; 4: m. pectoralis minor; 5: m. rhomboideus; 6: m. serratus anterior; 7: m. deltoideus, scapular part; 8: m. deltoideus, clavicular part; 9: m. coracobrachialis; 10: m. infraspinatus; 11: m. teres minor; 12: m. teres major; 13: m. supraspinatus; 14: m. subscapularis; 15: m. biceps, caput longum; 16: m. biceps, caput breve; 17: m. triceps, caput longum; 18: m. latissimus dorsi; 19: m. pectoralis major, thoracic part; 20: m. pectoralis major, clavicular part; 21: reaction force thorax -> TS; 22: reaction force thorax -> Al; 23: lig. conoideum; 24: lig. trapezoideum; 25: lig. costoclavicularare; 26: weight of the arm; 27: propulsion force; 28: internal moment exerted by the hand at the rim.
Figure 11:
Moment balance (in Nm) around the joint axes of the acromioclavicular joint. Only moments larger than 1 Nm are shown (subject EL, 20% VMV). For explanation symbols see Fig. 10.
A: frontal (X-)axis; B: vertical (Y-)axis; C: sagittal (Z-)axis.

Figure 12:
Moment balance (in Nm) around the joint axes of the glenohumeral joint. Only moments larger than 1 Nm are shown (subject EL, 20% VMV). For explanation symbols see Fig. 10.
A: frontal (X-)axis; B: vertical (Y-)axis; C: sagittal (Z-)axis.
for X-, Y- and Z-axis, respectively. The main muscle counterbalancing the external moment is m. serratus anterior. M. trapezius hardly exerts a moment around the X-axis. This muscle mainly serves to position the clavícula in order to provide a supporting point for the large lever arm of m. serratus anterior (Van der Helm, 1991b). Considering the moments around the Y-axis, the external moment presses the scapula against the thorax. The main counterbalancing moments are exerted by reaction forces between the thorax and the medial border of the scapula. The external moment around the Z-axis is generally low in the first three hand positions, since the propulsion force vector has a small angle with this axis. In hand position 4 and 5 the propulsion force vector becomes more vertical and will exert a larger moment around the Z-axis (and in addition a smaller moment around the Y-axis!). M. serratus anterior, m. trapezius and the scapulothoracic force at AI have a moment component around the Z-axis, in the same direction as the external force. This moment is counterbalanced by the weight of the arm, m. pectoralis major, thoracic part, (though this muscle has no attachments on clavícula or scapula!), and the conoid ligament which prevents torsion of the capsule of the AC-joint (Van der Helm, 1991b).

Figs 12A to 12C show the moment equilibrium of the GH-joint around the X-, Y- and Z-axis, respectively. The main external moment is around the X-axis with a peak at hand position 2. In the retroflexed position of the humerus no dominant muscle is present which can exert the necessary anteflexing moment. M. deltoideus, pars anterior, m. pectoralis major, thoracic part, m. biceps, caput longum, m. supraspinatus and m. infraspinatus contribute to the counterbalancing moment. Muscle forces often rise to peak levels in hand position 2 (Fig. 9). Surprisingly, m. pectoralis major, pars clavicularis, a muscle considered in most textbooks as anteflexor, is hardly active in the model simulations (Fig. 9T), since its muscle lines of action are almost parallel to the frontal (X-)axis and therefore this muscle part can not exert a moment around this axis. In some subjects, the external moment switches from a retroflexing moment to an anteflexing moment in hand position 5 (Figs 7C and 7D). Then, muscles like m. latissimus dorsi, m. teres major en m. teres minor become active to exert a retroflexing moment around the glenohumeral joint. Considering the moment around the Y-axis, m. subscapularis is almost the only counterbalancing muscle. Other muscles crossing ventrally the glenohumeral joint do not exert a moment around the Y-axis. Muscle lines of action of m. deltoideus, pars anterior, are mainly vertical and not oblique, whereas m. pectoralis major, pars clavicularis, lies caudally to the joint rotation center, due to the retroflexed position of the humerus, and therefore it has no moment arm around the vertical axis. In hand positions 3, 4 and 5 m. pectoralis major, thoracic part, supports m. subscapularis. Moments around the Z-axis are mainly due to side-effects of muscles contributing to moments
around the other axes. M. pectoralis major, pars sternalis, exerts the main moment counterbalancing the external moment.

V Discussion
During wheelchair propulsion the largest net joint moments and net joint power are generated at the shoulder mechanism (Cerquiglini et al., 1981; Veeger et al., 1991b). Goal of the current research project is to investigate which muscles are active in the power generation. In order to achieve this goal, a musculoskeletal model is necessary which can predict the distribution of net joint moments and net joint powers over the muscles. Recently, such a musculoskeletal model of the shoulder mechanism has been developed and applied to loaded and unloaded humeral abduction and anteflexion (Van der Helm, 1991ab). In this study it is verified whether the model provides reliable results in more complex situations such as in wheelchair propulsion in which an unusual combination of humeral position and external force exertion occurs.

Main problem in dynamic simulations of wheelchair propulsion is that motions of the shoulder girdle can not be recorded dynamically. Therefore, in this study the musculoskeletal model has been verified under static conditions. Positions have been recorded by means of a combination of optical methods and a palpation method. Using a musculoskeletal model of the shoulder mechanism which comprehends all relevant gross morphological structures, muscle forces and moments have been calculated. The motion recording method and musculoskeletal model have been verified by comparing on/off patterns of EMG and muscle force (Figs 8A to 8J). For seven out of ten muscles EMG and force patterns were consistent. For m. pectoralis major, pars clavicularis, and m. deltoideus, pars medialis, EMG activity has been recorded whereas no muscle force is predicted by the model. Since other parts of these muscles are very active, it could be that some crosstalk between electrodes occurs or that neural control is not sufficiently specific to inhibit these parts while other parts are active. EMG activity of m. triceps, caput longum, is likely due to the necessary extending moment at the elbow joint. In the model no muscle force is predicted for m. triceps, caput longum, since in the model only the moment balance of the shoulder mechanism is studied and not the moment balance of more distal joints.

Positions, external forces and EMG have been recorded at four young healthy male subjects whereas simulations have been performed with the morphology of a 90-year old male (Table 1). Though this cadaver was a more or less median cadaver in the cadaver study (Veeger et al., 1991a; Van der Helm et al., 1991), discrepancies between the physical abilities of the subjects and the cadaver are inevitable. For example, forces exerted by the subjects were too high to simulate with
the maximal muscle force of the cadaver, as derived from the physiological cross-sectional area (PCSA). Therefore, the PCSA has been scaled by setting the force per unit area from 30 N/cm² to 100 N/cm². 30 N/cm² is a conservative factor. In literature, values up to 100 N/cm² have been reported (see Weijts & Hillen, 1985, for review). Whenever the morphology can be recorded in the same subject as in which the positions, external forces and EMG are recorded, muscle force predictions will be presumably even more accurate. Generalization of results obtained from healthy subjects to wheelchair users is hazardous, since the latter group is very heterogeneous (Veeger et al., 1991; Brown et al., 1990). However, the same experimental method can be applied for wheelchair users.

The use of a mathematical optimization criterion is a rather arbitrary way of force distribution over the muscles in the system. Though the criterion reflects an interpretable physiological phenomenon, i.e. high muscular stresses are not desirable, some principles of neuromuscular control may have been ignored. For instance, in the model it is assumed that all muscle elements are independent. However, maybe neuromuscular organization does not permit some muscle parts to contract independently, and will instead activate the whole muscle. This could probably explain the discrepancy between EMG and force patterns of m. deltoideus, pars medialis, and m. pectoralis major, pars clavicularis. On the other hand, the consistency mentioned above between EMG and force patterns increased our confidence in the model. Whenever the position of the bones and external forces can be recorded accurately, the model will give reliable predictions of muscle forces in the shoulder mechanism.

In static experiments, irrespective of the optimization criterion used the desired external moment is mainly decisive which muscles will become active to exert a counterbalancing moment or to fix joints by neutralizing undesired actions of other muscles. During dynamic wheelchair propulsion, it is also important whether or not the muscles contribute to the positive power development which depends on the joint angular velocity, and hence the muscle contraction velocity, as well. The trajectory of the hand is prescribed by the rim. Hence, the trajectory of the joint angles and their angular velocity are determined within narrow limits. Muscles can only generate positive power when they shorten while exerting force. Shortening of the muscles and hence the amount of power generated depends on the angular trajectory and angular velocity of the joints. Therefore, muscle activation patterns can be different in static and dynamic experiments.

Power generation at the shoulder mechanism is a complex event. External forces can be exerted in almost any direction and are usually applied by the hand which can result in large moment arms with respect to the shoulder. Hence, external moments can be large, even if external forces are
relatively small. Design parameters of the wheelchair, e.g. seat position with respect to wheel axle, rim diameter and camber angle, are important for the ability to generate power in an unnatural type of locomotion like wheeling. In the static push, the propulsion force vector is directed roughly tangential to the rim, pointing upwards in hand position 1 and pointing downwards in hand position 5 (Fig. 3A). The largest external moment is in hand position 2. In dynamic situations patterns of external moments are somewhat different (Fig. 3B). At the start of the propulsion cycle the arm must be accelerated until the hand reaches the velocity of the rim. Therefore, the force and external moment in hand position 1 and 2 will be smaller than in static experiments. In hand position 2, 3 and 4 the force vector is directed more downwards instead of tangential to the rim. In hand position 5 the force direction is along the stretched arm. In this position the angular velocity around the glenohumeral joint is close to zero and it will not contribute to power generation. At most, forward flexion of the trunk will add to the power.

In this simulation study the sum of squared muscle stresses is minimized. Van der Helm (1991) showed that other non-linear criteria such as minimization of maximal muscle stress in the system did not lead to different results. Main factor for muscle activity is whether or not the resulting moment exerted by the muscle is effective which depends on the muscle moment arm. Because of the large number of muscles crossing the joints of the shoulder mechanism and considering the six DOF, there seems to be much redundancy in the system. For instance, the glenohumeral joint is modelled with three DOF whereas ten muscles are crossing the joint. However, since muscles can only exert traction forces, the number of muscles needed is doubled. In addition, the upper extremity is multi-functional. The desired external moment vector in the glenohumeral joint can point anywhere in the eight quadrants of the coordinate system. A muscle is probably only active when its resulting moment vector is in the opposite direction of the external moment vector. If the angle between both vectors becomes smaller, compensatory muscle force is needed which is not favoured by the optimization criterion. In this simulation study, especially the moment around the vertical (Y-)axis of the glenohumeral joint was critical. Only m. subscapularis of which the important role in the shoulder mechanism is only recently fully appreciated (Van der Helm, 1991), has a large moment arm with respect to this axis. The upper part of this muscle provides the necessary moment, since the lower part has a large moment arm around the frontal axis as well which would add to the already large external moment. The largest external moments are around the X-axis of the glenohumeral joint. A number of muscles is counterbalancing this external moment: M. deltoideus, pars anterior, m. pectoralis major, thoracic part, m. biceps, caput longum, m. supraspinatus and m. infraspinatus. These muscles are likely to contribute to external power.
in dynamic situations. Moments around the SC- and AC-joint and forces of the thoracoscapular muscles are not critical. Largest moments around these joints are due to the activity of m. serratus anterior. Since the scapula hardly changed position during these static experiments, the contribution of thoracoscapular muscles to the external power would be negligible. However, maybe in dynamic situations scapular motions are larger, which can be deduced from translations of the acromion. For the time being, the main function of the thoracoscapular muscles seems to be to stabilize the scapula in order to provide a stable base for the action of muscles crossing the glenohumeral joint. Results of this study show that the model can be used for dynamic situations. For dynamic simulations it is necessary to have a verified model of the shoulder mechanism. In addition, motions and forces should be recorded in dynamic situations. Comparison of EMG recordings and force predictions shows that the model provides reliable predictions of muscle forces in static situations. Input variables of the model should be expanded with velocity and acceleration of the generalized coordinates. In addition, an optimization criterion should be used in which muscle power is optimized. Therefore, a model describing the dynamic behavior of muscles must be included in the optimization procedure. Propulsion forces and moments at the rim and movements of the arm can be dynamically recorded using the wheelchair ergometer and two cinecameras. The manchets equipped with markers are necessary for accurately recording of the 3D position and orientation of upper arm and forearm, which include the axial rotation of these segments. For the time being, motion recording of the clavicle and scapula in dynamic situations is impossible. However, the position of the lateral end of the clavicle can be recorded using a marker on top of the acromion. Hence, position, velocity and acceleration of two input variables (ACy and ACz) are known and only one input variable is missing. Consequently, it can be assumed that in addition velocity will have no effect on scapular position, which means that the position of AI can be recorded in static situations. Finally, muscle forces calculated by model simulations can be compared with EMG recordings.

Dynamic simulations of wheelchair propulsion could demonstrate which muscles are important in power generation around the shoulder mechanism joints, and which muscles are important for stabilization. Then, the influence of factors affecting mechanical efficiency like discontinuity of force application, recovery phase, large muscle force, etc., can be assessed. Design parameters like seat height and wheel camber can be adjusted. Maybe other types of propulsion mechanisms can be developed which will be more efficient than rim-type wheelchairs.

It is concluded that the musculoskeletal model of the shoulder mechanism can be used in static simulations, as is demonstrated in this study for wheelchair propulsion, and presumably in dynamic
simulations. In order to be useful for analysis of the shoulder mechanism in ergonomics and rehabilitation, dynamic simulations are necessary.

Concluding remarks
1. Using a wheelchair ergometer, static positions of the trunk, shoulder girdle and upper extremity were recorded simultaneously with forces on the rim and surface EMG of ten muscles. The data have been used for simulations with a musculoskeletal model of the shoulder mechanism.
2. The similarity between recorded EMG and predicted muscle force patterns shows that model predictions are reliable.
3. The contribution of muscles to the moment balance around the shoulder mechanism joints during static wheelchair propulsion is determined.
4. The experimental set-up of this study can be expanded to dynamic situations. It is suggested that scapular and clavicular positions may be extracted from assumptions based on these static measurements.
5. The musculoskeletal model of the shoulder mechanism can be very useful in ergonomic studies to determine the mechanical load on morphological structures.
Chapter 10
Reflection

I Introduction

The shoulder mechanism is one of the most complex musculoskeletal systems of the human body. Thorax, clavicle, scapula and humerus are interconnected by three joints which allow a considerable range of motion, but the scapula is constrained by the thorax which results in forced rotations. Accurate positioning of the proximal bones of the upper extremity is essential for accurate positioning of the hand. Small positioning errors will be amplified by the length of the arm, resulting in larger errors at the hand. In addition, human beings are able to exert force with their hands in almost any direction which is accomplished by exerting sometimes large moments around the joints of the shoulder mechanism. A total of 17 muscles controls the 3D motions of the shoulder girdle and humerus.

Until now, it is not known how humans control their motions. Fig. 1 shows a speculative block

![Figure 1: Block diagram of the forward dynamic model.](image-url)
diagram of a control model of these motions, incorporating muscle dynamics and proprioceptive control. Motor programs for the muscles, generated in the brain, are the input to this model. These motor programs are converted into pulse trains (α-innervation) in the nervous system, thereby taking into account the dynamics of the neural system (Hatze, 1976; Audu & Davy, 1985; Winters & Stark, 1985). These pulse trains result in the so-called 'active state' of the muscle (Hatze, 1976; Hof & Berg, 1981). The force which is finally generated by the muscle depends on the muscle dynamics which can be described by the force-velocity and force-length relationship. The effect of these muscle forces depends on the motion equations describing the mechanical behavior of the skeletal system, i.e. the shoulder mechanism. The result of muscle forces are motions of the bones. A number of feedback systems is presented in the model. Motions of the bones will result in length changes and velocities of the muscles. Due to the force-length and force-velocity dependency of muscles, length and velocity will have an effect on the muscle force generated. Length and velocity of intrafusal muscle fibers which are individually innervated by γ-innervation, are detected by muscle spindles. Forces at the tendons are detected by Golgi tendon organs. These variables are fed back to the central nervous system (CNS), obviously to adapt the active state in order to arrive at the desired motor action. It is unknown how exactly these proprioceptive feedback loops function in the control of motions. Other feedback loops are due to joint capsule receptors, skin receptors and the visual system. These feedback loops are believed to control motions on the level of coordination of complete muscle groups instead of individual muscles, and are considerably slower systems (Voorhoeve, 1984).

The model as shown in Fig. 1 is a forward or direct dynamic model with feedback loops. Main difficulty in studying the behavior of the entire system is that it is impossible to measure in vivo pulse trains and muscle forces. Only the output variables, the resulting motions and external forces, can be measured. In combination with the limited knowledge of the model structure and the complexity of the model, at present parameter estimation of a possible model is not feasible.

The behavior of the model can be studied through simulations. An input signal (motor program) is adjusted iteratively until the desired motion results. However, this type of simulation requires much computing time. In the research project presented in this thesis, the inverse model of the motion equations describing the mechanical behavior of the musculoskeletal model has been used. This inverse dynamic model is presented in Fig. 2. In the inverse dynamic model muscle forces are calculated which are most likely responsible for the recorded motions. Neural activation, muscle dynamics and feedback loop dynamics have been neglected. Advantage of the inverse dynamic model is that computing time is decreased considerably since the differential equations
have been reduced to algebraic equations. Results of the inverse dynamic model can be used to assess the correctness of the motion equations and for calculations in static situations and in slow motions. The model structure has been based on an analysis of the mechanical relevance of gross morphological structures. The output variables, i.e. muscle forces, can not be recorded accurately enough for direct estimation of the model parameters. Therefore, model parameters must be physically interpretable so that they can be derived in a morphological study. Then, the model can be verified by comparing model predictions with the system behavior, e.g. by comparing muscle force with recorded EMG, or by comparing predicted maximal force output in a number of force directions with recorded maximal external force.

Since generally there are more unknown muscle forces than motion equations in the model, muscle forces can only be calculated using an optimization criterion. Interpreted in the forward dynamic case, it means that several activation patterns can result in identical motions. Since motion patterns and EMG patterns are highly reproducible (Van der Hoeven, 1988; Van der Helm & Pronk, 1991), apparently it may be concluded that the CNS has optimized its control actions. Knowledge of the optimization criterion is necessary for correctly determining muscle forces.

1.1 Model structure

In this study a dynamic model of the shoulder mechanism has been developed based on the finite element method, implemented in the computer program 'SPACAR'. Gross morphological structures have been represented by mechanical equivalents (~ elements) of which the kinematic and dynamic behavior is well-known. The behavior of the entire mechanism can be calculated by connecting these elements. The structure of the model is physically well interpretable, because the position and orientation of each element are known. The same model structure can be applied in a number of

![Figure 2: Block diagram of the inverse dynamic model.](image-url)
situations: forward dynamic, inverse dynamic, static and kinematic.

Some newly developed, special purpose elements were necessary for the correct modelling of the shoulder mechanism using the finite element method. Muscles with curved muscle lines of action have been represented by CURVED-TRUSS elements. The mechanical effect of the scapulothoracic gliding plane has been represented by SURFACE elements of which the internal stresses are constrained to compression stresses. Because of the small articular surface of the glenoidal cavity, only a subset of reaction forces is possible in the glenohumeral joint. The reaction force calculated in the model is restricted to this subset, otherwise the joint would dislocate.

Modelling ligaments is difficult in an inverse dynamic analysis. Since the position of the mechanism is input for the model, the length of the ligament is determined and so is the force in the ligament. Due to the unknown zero-force length and stress-strain relationship, and inaccuracies in motion recording, odd ligament forces could result. However, even in a forward dynamic analysis the unknown zero-force length and stress-strain relationship would affect the calculated motion.

1.2 Model parameters

Of late, in the spirit of the time, several research teams all over the world have concluded almost simultaneously that a morphological study is necessary to obtain accurate model parameters for developing a musculoskeletal model of the shoulder (Hogfors et al., 1987; Wood et al., 1989ab; Bassett et al., 1990). It is acknowledged that the coherence between geometry, physiological cross-sectional area and inertia could be important and hence all parameters should be recorded within one human body. In this thesis a cadaver study consisting of 14 shoulders of 7 cadavers has been described. Thus far, for model simulations only the morphology of one, more or less median, cadaver has been used. The recording of motions and EMG in healthy young subjects whereas muscle forces are calculated using the morphology of a cadaver of a 90-year old man, can probably result in discrepancies. For instance, in simulating static wheelchair propulsion the physiological cross-sectional area of the muscles had to be scaled in order to enable the high external forces as exerted by the subjects. Maybe some typical results of the simulations are due to the specific morphology of this cadaver.

The correct determination of muscle lines of action representing the mechanical effect of the muscles with large attachment sites is important for the simulation results. In contrast to the authors mentioned above who divided the muscles a priori into functionally independent parts, in this study the complete attachment site and architecture have been mathematically described which allowed
derivation of the muscle lines of action a posteriori.

1.3 Motion recording

Motion recording of the shoulder girdle bones is the most problematical part in developing a musculoskeletal model. Due to the complex motion constraints of the scapula imposed by the clavicle and thorax, three-dimensional (3D) motion recording is necessary. Two-dimensional motion recording methods like roentgenography which is the method most applied, will never yield enough information to reconstruct 3D motions. For 3D motion recordings using two roentgen cameras it is necessary to implant markers into the bones in order to achieve sufficient accuracy. Only Peterson et al. (1985) reported such measurements but did not include any results. In this thesis the use of a palpation technique has been described. In this palpation technique bony landmarks are palpated and subsequently digitized. With at least three non-collinear bony landmarks the position of the bone can be reconstructed. Disadvantage of the palpation technique is that it is only applicable in static situations.

Since no 3D recordings of the shoulder girdle bones were available, thus far in literature no attention was paid to an unambiguous description of the 3D motions. In this thesis local coordinate systems of the bones have been defined using bony landmarks. Motions of the bones have been defined by rotations of these local coordinate systems. A number of possible motion definitions exists: bony rotations vs. joint rotations, choice of reference position, choice of local or global coordinate systems relative to which the rotations can be defined. The rotation of these local coordinate systems with respect to the global coordinate system or with respect to other local coordinate systems defines the motion of the bones or the motion in the joints, respectively. Particular problem in describing the motions of the shoulder girdle is that no anatomical position of the bones has been defined. Therefore, a virtual reference position of the local coordinate systems has been defined in which the axes coincide with the axes of the global coordinate system. Advantage of this definition is that the position of the bones with respect to the global coordinate system can be compared between subjects. In addition, rotations around axes of the global coordinate system are better interpretable than around axes of a local coordinate system of which the initial orientation is hardly conceivable. Though, for describing rotations in the joints the rest position of the bones, i.e. the position with the humerus hanging down aside the body, is preferred as reference position. Rotations with respect to this position should be described with respect to the global coordinate system. Advantage of this definition is that in the initial position rotation in the joints is zero, whereas subsequently joint rotations can be interpreted as rotations around the
axes of a global coordinate system. Anyhow, well-defined coordinate systems are necessary for an unambiguous description of the bony motions which enables renewed calculation using any motion definition.

Rotations can be parametrized in two ways: Using Euler angles and using helical axes. Three-dimensional rotations are non-linear, so in any parametrization the variables will be mutually dependent. This implies that one variable can only be interpreted and used for comparisons considering the values of the other variables. The palpation technique is not accurate enough to assess translations in the joints. In addition, since only static positions are recorded, position changes are large and the helical axes do not approximate the instantaneous screw axes. Therefore, in this thesis Euler angles have been chosen to describe rotations.

1.4 EMG recordings

Since the output variables of the model, i.e. muscle forces, can not be recorded directly, EMG has been used to assess muscle activity. Generally, the amplitude of rectified integrated EMG is considered to represent the so-called active state of the muscle (Hof & Berg, 1981; Bogert, 1989). However, this implies that maximal EMG should be independent from muscle length, which is in contrast to the conclusions of Heckathorne & Childress (1981) that EMG represents active state and an additional, unknown factor. Force development can be described by a first-order model in which the force-velocity and also the force-length relationship is important. When static positions are simulated, the effect of velocity can be neglected. The effect of muscle length is more cumbersome. The force-length relation is usually approximated by a parabole which is defined by the length at which the muscle can generate maximally active force (muscle optimum length) and by the width of the parabole which determines the length at which the muscle still can exert force (Woittiez, 1984). To date, both factors can not be assessed in vivo. If the passive force-length relation is taken into account, more unknown factors are added.

Two methods have been applied to incorporate muscle length into the force-EMG comparison. In the first method, a linear relation has been assumed between the quotient of recorded EMG and maximal EMG, and the quotient of muscle force calculated by the model and maximal muscle force. Maximal muscle force has been estimated using the muscle length calculated by the model and a normalized force-length relationship with muscle optimum length assumed to be equal to the initial muscle length, i.e. the length in the initial (rest) position with the arm hanging aside the body. In this relation it is implicitly assumed that maximal EMG is a representation of the active state and that it is independent from muscle length. In the second method, a linear relation has been
assumed between recorded EMG divided by maximal EMG recorded in the same position, and calculated muscle force divided by maximal muscle force which is calculated by the model when maximal external force is simulated. However, neither of these methods resulted in a linear relation. It is concluded that the unknown factors, i.e. EMG-active state relationship, muscle optimum length and width of the force-length relationship, impede a quantitative validation of the output variables with EMG. Only simultaneous changes (on/off patterns) in muscle force and EMG due to e.g. cyclic motions or changing force directions, can be used for verification purposes. In this thesis, it is concluded that the similarity between muscle force calculations and EMG recordings during static wheelchair propulsion shows that the muscle force predictions of the model are reliable, though only in a qualitative sense.

II Simulations
As for most human joints, in the shoulder mechanism more muscles are present than strictly necessary to control the degrees of freedom (DOF). In other words, several muscle control schemes can result in the same motion. An important question for submaximal force exertion is which distribution factor amongst muscles is used by the human body. Bony motions and EMG patterns in the shoulder mechanism are highly reproducible within one subject. Thus, it is not likely that muscles are activated coincidentally. The distribution factor or optimization criterion which is used by the human body is likely to be task-dependent. For long-lasting submaximal tasks probably the energy used by muscles is minimized or fatigue resistance is maximized.

In the shoulder mechanism model the excess of muscles with respect to the DOF leads to a mathematical situation in which an infinite number of solutions exists: The indeterminacy problem (Hardt, 1978; Dul, 1984\textsuperscript{ab}). This problem can be solved mathematically using an optimization criterion. If this criterion resembles the real criterion used by the human body, the calculated muscle forces are likely to resemble the real muscle forces. However, one only can guess at the real criterion. Non-linear criteria are preferred in order to enforce synergism between muscles (Dul, 1984\textsuperscript{a}). In this thesis, four different non-linear criteria have been discussed: Minimization of the sum of squared muscle forces, minimization of the sum of squared muscle stresses, minimization of the sum of squared quotients between muscle force and length-dependent maximal muscle force, and minimization of the maximal muscle stress in the system. Any of these criteria will penalize antagonistic action. Though, antagonistic action can be necessary for joint stabilization as is demonstrated by the constraint at the reaction force vector of the glenohumeral joint, which resulted in the activation of antagonistic muscles with small moment arms. Resultant muscle forces,
subject to these criteria, only differ in a quantitative sense. Hence, EMG amplitude can not
distinguish between these criteria, since its amplitude depends on muscle length, active state,
muscle optimum length and width of force-length relation. In addition, the criteria will result in
similar on/off timing patterns of muscle force, so even if on/off patterns are compared with EMG
patterns, conclusions can only be drawn in a qualitative sense.
In comparing the four optimization criteria, a few conclusions can be drawn. Using the first
criterion, minimization of the sum of squared muscle forces, calculated muscle forces will heavily
depend on the number of muscle lines of action used to represent the muscles. The fourth criterion,
minimization of the maximal muscle stress in the system, is numerically unstable. If one muscle
is required to exert high forces, because no other muscle can aid to its specific action, the solution
for all other muscles is still indeterminate, since their stress will always remain under the maximal
stress as calculated for the first muscle. The results for the second and third criteria are similar.
However, when the third criterion, minimization of the sum of squared quotients of calculated
muscle force and maximal muscle force, is used, the results are unreliable in high elevation
positions where the length of some muscles decreases down to less than 60% of the initial
(optimum) length which restricts these muscles from force exertion. The lack of knowledge about
muscle optimum length and force-length relationship hampers the use of the third criterion.
Therefore, it is concluded that for the time being the second criterion is preferred. It is well
interpretable, because high muscular stresses are penalized. However, the relation of this criterion
with minimization of energy consumption remains obscure. It is implicitly assumed that high
muscular stress will result in large energy consumption.
Distribution of muscle forces mainly depends on the moment which can potentially be exerted by
the muscle. Figs 3A and 3B show x-, y- and z-components of the unit direction vector of the
external moment around the glenohumeral joint in a certain position during static wheelchair
propulsion, in comparison with the components of all 95 unit direction vectors of muscle moments.
The external moment vector can point into any of the eight quadrants of the global coordinate
system. In comparison with only eleven muscles crossing the glenohumeral joint, the redundancy
in the system is not large. Only muscles of which the moment vector is close to the external
moment vector, will contribute to the resultant muscle force vector. Muscle moment vectors with
large angles with the external moment vector must be compensated by muscle moment vectors with
a similar deviation into the opposite direction which is definitely not preferred by any of the
optimization criteria.
Since the shoulder mechanism is multifunctional, the direction of the external moment vector can
be anywhere in the eight quadrants. The distribution of the muscle moment vectors over the eight quadrants is determined by the position of origin and insertion. Muscle origins and insertions are likely to be the result of a slowly evolutionary process, instead of fast adaptation to momentary functions. Therefore, it is impossible to say that the muscle moment vector distribution is anywhere near an optimal distribution. On the contrary, functions of the upper extremity are the result of more or less coincidentally situated muscles. Researchers are destined to analyse only how these functions come about and which muscles are responsible, and they are incapable to answer the question why these muscles are situated here and there.

It is concluded that the direction of the muscle moment vector is decisive for the function of the muscle, whatever of the four optimization criteria is used. Since the muscle moment vector depends on the position of origin and insertion with respect to the rotation center, it will change due to motions of the bones. Therefore, it is necessary to include motions of the shoulder girdle bones for deriving muscle functions in the shoulder mechanism.

The scapulothoracic gliding plane is very important for the dynamic behavior of the shoulder girdle. Since the shoulder girdle is a closed chain, the scapula is subject to forced rotations and the moment balances around the sternoclavicular and acromioclavicular joints are coupled. Reaction forces between thorax and scapula contribute to the moment balance of these joints. These reaction

Figure 3: X-, Y- and Z-components of the external moment vector (tot) and muscle moment vectors around the glenohumeral joint, normalized to unit vectors.
forces are due to forces exerted by muscles which in turn have their contribution to the moment balance according to their moment vector.

Results of the inverse dynamic approach are valid to assess the static force distribution in the system, within the limitations imposed by the assumptions discussed before. In the static case, muscle dynamics can be neglected. Only the force-length relation is of interest. In addition, the effect of feedback loops for motion control can also be neglected. Muscle force calculations in static situations provide insight in which muscles are well located to exert moments. It is the first step in analyzing muscle functions. Also for slow motions, in which the dynamics of the system are of minor importance, (quasi)static predictions are valid. If the motions are faster, the dynamics of the system can no longer be neglected.

III Concluding remarks
The finite element musculoskeletal model provides good insight in the mechanics of the shoulder mechanism. Simulations result in a reliable prediction of muscle function for most muscles. It is a three-dimensional model and includes a representation of the scapulothoracic gliding plane and a representation of muscles which are wrapped around bony contours. The reaction force vector in the glenohumeral joint was constrained to point inside the glenoid cavity, otherwise the joint would dislocate. This constraint resulted in activity of antagonistic muscles with small moment arms, i.e. the rotator cuff muscles which are located in a half circle around the joint.

Parameters of the model, i.e. inertia, geometry and muscle contraction parameters, have all been obtained from the same cadaver. Hence, possible important features as the location and size of the muscles with respect to each other have been preserved. Recording of the muscle bundle direction made a new theory possible for the representation of the mechanical effect of muscles with large attachment sites by multiple lines of action.

The computer program 'SPACAR', based on the finite element method, offers a powerful tool for modelling multibody systems. Changes in model parameters and model structure, i.e. addition or omission of elements, are simply to realize. The choice of generalized coordinates can easily be adapted.

Muscle functions have been analyzed by deriving a moment balance around the axes of the sternoclavicular, acromioclavicular and glenohumeral joint. In this analysis the position change of the bones of the shoulder girdle has been taken into account. The location of muscles with respect to the joint rotation centers, and with respect to each other, is important for the resulting muscle functions, especially when the large moment arms due to the triangular shape of the scapula are
Reflection

considered.

It is concluded that the function of thoracoscapular muscles can not be assessed when the scapulothoracic gliding plane is not included in the musculoskeletal model. Since no mono-articular muscles are crossing the acromioclavicular joint, it can be argued that the connection between the medial border of the scapula and the thorax is necessary for the stability of the shoulder girdle. In fact, during humeral abduction the reaction force of the thorax at angulus inferior results in the main moment around the sagittal axis of the sternoclavicular joint.

It is concluded that model predictions of muscle force can not be validated using the amplitude of surface EMG, due to the unknown force-length relationship and EMG-length dependency. Nevertheless, model predictions are considered to be valuable, because the origin of the predictions can be traced since all model parameters are physically interpretable.

It is concluded that results of the model simulations as presented in this thesis, provide a reliable prediction of the function of muscles during static humeral abduction and anteflexion, and during static wheelchair propulsion. The limitation to static situations originates from the nature of the motion recording experiments. For the time being, it is impossible to record clavicular and scapular motions in dynamic situations.

Three-dimensional positions of the shoulder mechanism bones have been recorded using a palpation technique. Two methods of motion description have been proposed using Euler angles. The position of the bones have been described with respect to a virtual reference position. Joint rotations have been described starting from the initial, rest position.

The model proved to be useful in predicting the effect of fusion angles in a glenohumeral arthrodesis. The model can be used for improving the diagnosis of shoulder problems by analyzing the normal situation and for predicting the effect of operations like tendon transfers or implantation of an endoprosthesis. The model can be useful in ergonomic situations, as is partly proved by the application to the wheelchair user.

IV Future research

Good research generates more questions than answers. The development of a reliable musculoskeletal model allows for an extensive and more detailed analysis of the shoulder mechanism.

In the near future model parameters of the other bodies recorded in the cadaver study will be used to assess the sensitivity of the results for the morphology which probably can explain the sometimes large interindividual differences encountered in motion recording and EMG studies. In order to use
the model to predict the behavior of the shoulder mechanism in individual subjects or patients, the morphology of these subjects must be recorded. Recently, a project has been initiated for automatic detection of the contours of morphological structures from Magnetic Resonance Imaging (MRI) recordings (Kuntz, 1990). From these contours the parameters needed for the shoulder mechanism model can be derived.

Muscle functions have been analysed in a few static situations, i.e. humeral abduction and anteflexion and wheelchair propulsion. For a complete analysis of muscle functions and further verification of the model, motions will be recorded for the complete motion range of the humerus, including force exertion by the hand in multiple directions for each position. Then, in each position muscle force predictions can be compared with EMG recordings, since the effect of muscle length has been excluded from the analysis. In addition, maximal muscle force predictions by the model in each position can be compared with actual recordings.

Angular velocity of the humerus is likely to affect the scapulohumeral rhythm. In static situations the scapulohumeral rhythm is likely to be affected by the force-length relationship of the muscles and probably by external load. In dynamic situations the force-velocity relationship of the muscles could favor some muscle to exert most of the power needed. The problem will be twofold: Neuromuscular dynamics should be incorporated in the model and the scapular motions probably will change, which implies that dynamic motion recordings should be performed. Recently, Groot (1991) has implemented a dynamic model of the neuromuscular system in the finite element method, based on the work of Winters & Stark (1985) and Happee (1991a). Theoretically, this model can be used in two ways for the shoulder mechanism. In a forward dynamic simulation the optimal control can be derived for all muscle elements, comparable with the work of Hatze (1976), Davy & Audu (1987), Pandy et al. (1990) and Pandy & Zajac (1991). However, this would require much computing time and is therefore nowadays not feasible. Happee (1991b) developed a faster algorithm using an inverse dynamic muscle model for rapid convergence. In the near future, this algorithm will be implemented in the finite element method which will hopefully result in acceptable computing effort.

Muscular contraction velocity has its effect on muscular dynamics, and will probably has its effect on the load sharing problem between muscles. In the optimization criterion which is used to solve the load sharing problem, the energy which is consumed by the muscles will be incorporated. Energy consumption depends on the muscular contraction velocity. This will be especially important in order to assess the role of bi-articular muscles, since these muscle have been proven to be efficient when a contradictory situation occurs in which the moment needed around a joint
axis is opposite to the angular velocity of the joint (Ingen Schenau, 1989).
The admissibility of extrapolation of static positions to dynamic trajectories will be investigated.
Therefore, the dynamic motions of the shoulder girdle must be recorded. Recording the dynamic
motions of the shoulder girdle is only possible using roentgenography. Using 2D roentgenograms,
it can be revealed whether or not velocity affects the scapulohumeral rhythm. Quantification of the
effect is impossible due to projection distortions. For 3D roentgen cinegraphy, markers must be
implanted to achieve sufficient accuracy, which is presently not allowed in healthy subjects.
However, the use of dissolvable markers will be investigated which will probably be permitted in
future.
An ultimate goal of the research is to detect why the scapula moves as it does. Given the position,
velocity and load on the humerus, the scapula shifts over the thorax to seek for an optimal position.
It is unknown which factors determine optimality, but likely similar factors as included in the
optimization criterion, e.g. energy consumption, are important, extended with coordination factors
as mechanical stability and muscular control. The scapular position can be optimized with respect
to such a criterion. The resulting positions can be compared with actual recordings. In the end this
would result in prediction of the scapular position based on the position, velocity and load on the
humerus. This would eliminate the necessity of recording the scapular position.
A dynamic extension of the musculoskeletal model will be useful in ergonomics, e.g. in wheelchair
propulsion. External load, position of the humerus and trunk, EMG as well as oxygen uptake can
be recorded in dynamic situations. From a recording of the shoulder top, i.e. the acromion, and
probably an extrapolation from static recordings or a prediction based on an optimization criterion,
the position of the shoulder girdle can be reconstructed. Then, the shoulder mechanism model can
be used in a power balance during cyclic movements. The consumed energy (oxygen uptake) per
unit of time is transformed by the muscles into external power and heat. The relation between
consumed energy and external power determines the mechanical efficiency of the system. Since
the mechanical efficiency of arm work is low compared with leg work, it is interesting to analyse
on a musculoskeletal level how much energy is used for stabilization. Probably, this research will
lead to better propulsion mechanisms for wheelchairs.
A biomechanical model of the shoulder mechanism has been developed with physically well-
interpretable parameters. The next step will be to extend the muscle elements in the model with
proprioceptive sensors and feed-back loops in order to analyse possible strategies of the central
nervous system (CNS) to stabilize the shoulder mechanism and to control motions. Special attention
will be paid to the mechanical stability of the glenohumeral joint. Since the joint capsule is very
lax and the glenoid cavity is small, mechanical stability is preserved by the activity of the rotator cuff muscles. The activity of these muscles will partly be feed-forward controlled, e.g. during fast movements, and partly be feed-back controlled, e.g. when pertubations in the load of the arm occur. The contribution of the feed-forward and feed-back control will be investigated. Results of this research is important for patients with a habitual subluxation of the glenohumeral joint which presumably lack adequate control of the rotator cuff muscles, and application of a glenohumeral endoprosthesis in which presumably the lack or distortion of proprioceptive information results in an overactivity of the rotator cuff muscles which leads to immobilisation of the joint.
References

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C
Clarjijs JP, Marfell-Jones MJ (1986). Anatomical segmentation in humans and the prediction of
Clauser CE, McConville JT, Young JM (1969). Weight, volume and center of mass of segments of
the human body. AMRL-TR-69-70, Wright-Patterson Air Force Base, Ohio.
Rev. 11: 421-432.
D
Davy DT, Audu ML (1987). A dynamic optimization technique for predicting muscle forces in the
DeLuca CJ, Forrest WJ (1973). Force analysis of individual muscles acting simultaneously on the
shoulder during isometric abduction. J. Biomech. 6: 385-393.
559-585.
Dempster WT (1955). Space requirements of the seated operator. WADC-TR-55-159, Wright
Patterson Air Force Base, Ohio.
Doody SG, Freedman L, Waterland JC (1970a). Shoulder movements during abduction in the
Rehab. 51: 711-713.
sharing between synergistic muscles. J. Biomech. 17(9): 663-673.
fatigue criterion for load sharing between synergistic muscles. J. Biomech. 17(9): 675-684.
Biomechanics XI-A, De Groot G, Hollander AP, Huijing PA, Ingen Schenau GJ van
(Eds.), Free University Press, Amsterdam: 471-476.
3: 124-128.
J. Biomech. 11: 219-225.
E
Engen, TJ, Spencer, WA (1968), Method of kinematic study of normal upper extremity
for determination of the shoulder complex sinus. J. Biomech. 20: 103-117.
Entken PJM, Wijgergangs GK (1984). Three-dimensional definition of both the shape of the
thoracic wall and the positions of the scapula and clavicle at humerus abduction. Rep.
F
Netherlands (In Dutch).
Freedman L, Munro R (1966). Abduction of the arm in the scapular plane and glenohumeral
Fukuda K, Graig EV, An KN, Cofield RH, Chao EY (1986). Biomechanical study of the
G

H

I

J
Johnston TB (1937). The movements of the shoulder joint. A plea for the use of the plane of the


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References


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References

neurophysiology, Karger, Basel: 607-622.

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Summary

Patients with a brachial plexus lesion sometimes undergo a glenohumeral arthrodesis in order to prevent the joint from subluxation and to achieve some function restoration of the arm. As a result of the arthrodesis, the scapula and humerus start to move as one bone and the thoracoscapular muscles are the actuators. This study was initiated to predict the optimal fusion angles for a glenohumeral arthrodesis. The altered muscle functions and subsequently changed motion range and ability to exert forces was the main research goal. Very little quantititative as well as qualitative knowledge existed about the mechanics of the shoulder mechanism. During the project the aim of the research became more generally to get insight into the mechanics of the shoulder mechanism, and especially into the function of muscles and ligaments. Therefore, a inverse dynamic model of the shoulder mechanism has been developed.

The shoulder mechanism is one of the most complex musculoskeletal systems of the human body. Thorax, clavicula, scapula and humerus are interconnected by three joints which allow a considerable range of motion. Motions of the scapula are constrained by the scapulothoracic gliding plane which turns the shoulder girdle, i.e. clavicula and scapula, into a closed-chain mechanism. Three extracapsular ligaments are crossing the joints and seventeen muscles control the motions of the shoulder girdle and humerus.

The shoulder mechanism model is based on the finite element method as implemented in the computer program 'SPACAR'. It is an inverse dynamic model: Input variables are the position, velocity and acceleration of the bones, and external forces. Output variables are muscle forces calculated subject to an optimization criterion. All gross morphological structures have been represented by elements: The bones have been represented by BEAM elements, joints by HINGE elements, ligaments by TRUSS elements, the scapulothoracic gliding plane by two specially developed SURFACE elements, and muscles by force-generating TRUSS or CURVED-TRUSS elements which represent the muscle lines of action. CURVED-TRUSS elements are wrapped around the surface of bony contours. Four optimization criteria have been compared: Minimization of the sum of quadratic muscle forces, minimization of the sum of quadratic muscle stresses, minimization of the sum of squared quotients of muscle force and length-dependent, maximal muscle force, and minimization of the maximal muscle stress in the entire system. Additional constraint to the optimization was the requirement that the resultant force vector in the glenohumeral joint must point from the rotation center inside the glenoid cavity, otherwise the joint will dislocate.

Parameters for the model have been derived in an extensive cadaver study consisting of 14
shoulders of 7 cadavers. In the model parameters of one more or less median cadaver have been used. From anthropometric measurements the inertia of the segments has been estimated. The physiological cross-sectional area (PCSA) and weight of the muscles have been recorded in order to estimate the maximal muscle force. The so-called palpator, a kind of spatial digitizer, has been used to record the three-dimensional (3D) positions of morphological structures, including muscle and ligament attachments, articular surfaces, bony contours, the scapulothoracic gliding plane and bony landmarks. Geometric forms have been fitted to the data points of the morphological structures for implementation into the model. In addition, a new theory has been developed for representation of the mechanical effect of muscles with large attachment sites by multiple lines of action. In the model one to six lines of action have been used for each muscle.

Positions of the shoulder mechanism bones were recorded in 10 subjects during unloaded and loaded abduction and anteflexion. Mean values were used as input for the model. Since only static positions could be recorded, the model has also been used for static calculations only. The length change of shoulder muscles during humeral elevation is very large compared with leg muscles during locomotion. This large length change would imply that some muscles could not exert any force at 180 degrees humeral elevation when a normal force-length relation between 60% and 140% of muscle optimum length is assumed. It has been concluded that the amplitude of rectified and integrated EMG could not be used to validate the model, due to the length-dependency of EMG and the unknown force-length relationship. Only on/off comparisons can be made which was not sufficient to distinguish between the four optimization criteria. Muscle functions have been analysed by establishing moment balances around the axes of the three shoulder mechanism joints. Then, forces at the scapulothoracic gliding plane are part of the balance. M. serratus anterior exerts the largest moment around the sternoclavicular (SC-) and acromioclavicular (AC-)joint. M. trapezius merely serves to position the clavicle in order to provide a point of support for the large lever arm of m. serratus anterior. Next to m. deltoideus, m. subscapularis was the second important abductor of the glenohumeral (GH-)joint. Rotator cuff muscles were active in controlling the stability of the GH-joint by pointing the resultant force vector to the articular surface of the glenoid. Lig. conoideum counterbalances the axial rotating forces at the clavicle of the clavicular parts of m. trapezius and m. deltoideus. As a result, the axial rotation of the clavicle closely follows the lateral rotation of the scapula, with minimal rotations of the AC-joint. During abduction, the largest moment around the sagittal axis of the SC-joint was due to the reaction force of the thorax at Angulus Inferior. The laterally rotated and protracted scapular position is a stable position which requires little muscle activity. Since no mono-articular muscles are crossing the AC-joint, the
connection between the medial border of the scapula and the thorax is necessary for the stability of the shoulder girdle.

Three-dimensional positions of the shoulder mechanism bones were recorded in 10 subjects using a palpation technique in which bony landmarks were palpated and subsequently digitized using the palpator. Hence, the position of the bones could be reconstructed. Accuracy of the palpation method is estimated to be in the range of common used 3D cinegraphic methods. Bony rotations have been described using Euler angles. Motion definitions are hampered by the lack of an anatomical position of the shoulder girdle bones and by the lack of well-interpretable rotation axes like a flexion/extension or abduction/adduction axis. Several 3D motion definitions have been compared. Finally, two definitions have been chosen since they provide the best interpretable rotations. In the first definition rotations of the bones have been described around axes of the global coordinate system with respect to a virtual reference position in which the bony local coordinate system coincides with the global coordinate system. In the second definition joint rotations have been described around axes of the global coordinate system with respect to the initial (rest) position of the bones at 0 degrees humeral elevation.

The shoulder mechanism model has been used to analyse the effect of different fusion angles on mobility area of the hand and maximal lifting forces after a glenohumeral arthrodesis. Scapular motions have been recorded in 10 subjects, mean fusion angles have been recorded in 18 patients and the morphology of one cadaver has been used. Maximal lifting force can be exerted when the scapula is in a protracted and laterally rotated position, due to the large moment arm of m. serratus anterior in that position. The mean fusion angles (i.e. subsequently 60 degrees endorotation, 9 degrees abduction and 13 degrees anteflexion) provided a mobility area of the hand in the midsagittal plane.

During wheelchair propulsion the largest external moments are exerted at the shoulder mechanism joints, but little is known about individual muscle contribution. Bony positions, external forces and EMG have been recorded simultaneously on a stationary wheel-chair ergometer. Muscle force patterns predicted by the model did reasonably agree with EMG-patterns. Muscle forces and moments around joint axes have been analysed. Extension of the recordings to dynamic situations is discussed.

It is concluded that a useful biomechanical model of the shoulder mechanism has been developed. Simulations result in a reliable prediction of muscle function for most muscles, and in addition ligament forces and joint reaction forces are calculated. The model allows for a wide application, e.g. improving the diagnosis of shoulder problems by analyzing the normal situation, and for
prediction the effect of operations like implanting an endoprosthesis. In ergonomic situations the model can predict the load on morphological structures, which can be helpful to improve the design of man-machine interfaces like the propulsion mechanism of a wheelchair.
Samenvatting

Bij patiënten met een beschadiging van de plexus brachialis wordt soms een glenohumerale arthrodeze uitgevoerd, met als doel te voorkomen dat het gewricht subluxeert en om enig functieherstel van de arm te bereiken. Als een gevolg van de arthrodeze zullen de scapula en humerus als één botstuk gaan bewegen, waarbij de thoracoscapulaire spieren als actuatoren functioneren. Dit onderzoek is gestart om de optimale fixatiehoeken te voorspellen bij een glenohumerale arthrodeze. Het voornaamste onderzoeksdoel was de veranderde spierfuncties en de daarmee samenhangende verandering van bewegingsruimte en vermogen om krachten uit te oefenen. Er waren zeer weinig kwantitatieve en kwalitatieve gegevens over de mechanica van het schoudermechanisme bekend. Tijdens het verloop van het onderzoek werd het doel van het onderzoek algemener gesteld, namelijk het verkrijgen van inzicht in het mechanisch functioneren van het schoudermechanisme, met name in het functioneren van spieren en ligamenten. Om die reden is een inverse dynamisch model van het schoudermechanisme ontwikkeld.

Het schoudermechanisme is een van de meest complexe spierskeletsystemen van het menselijk lichaam. De thorax, clavicula, scapula en humerus zijn onderling verbonden door drie gewrichten met een aanzienlijke bewegingsruimte. De bewegingen van de scapula worden beperkt door het scapulothoracale glijvlak, waardoor de schoudergordel, d.w.z. de clavicula en scapula, een gesloten keten vormt. Drie extracapsulaire ligamenten lopen over de gewrichten, en zeventien spieren sturen de bewegingen van de schoudergordel en de humerus.

Het model van het schoudermechanisme is gebaseerd op de eindige elementen methode zoals geïmplementeerd in het computerpakket 'SPACAR'. Het is een inverse dynamisch model: Ingangsvariabelen zijn de positie, snelheid en versnelling van de botstukken, en uitwendige krachten. Uitgangsvariabelen zijn spierkrachten, die berekend worden met behulp van een optimalisatiecriterium. Alle grote morfologische structuren worden geregistreerd door elementen: De botstukken worden geregistreerd door BEAM elementen, gewrichten door HINGE elementen, ligamenten door TRUSS elementen, het scapulothoracale glijvlak door twee speciaal ontwikkelde SURFACE elementen, en spieren door kracht-genererende TRUSS en CURVED-TRUSS elementen, die de werklijnen van spieren voorstellen. CURVED-TRUSS elementen zijn gekromd rond het oppervlak van botcontouren. Vier optimalisatiecriteria zijn vergeleken: Minimalisatie van de som van gekwadratierde spierkrachten, minimalisatie van de som van gekwadraterde spierspanningen, minimalisatie van de som van de gekwadraterde quotiënten van spierkracht en maximale, lengte-afhankelijke spierkrachten, en minimalisatie van de maximale spierspanning in het gehele systeem. De voorwaarde, dat de resulterende krachtvector in het
glenohumerale gewricht vanaf het rotatiepunt binnen de cavitas glenoidalis moet wijzen, is een extra beperking in de optimalisatieprocedure. Als aan deze voorwaarde niet zou zijn voldaan, zou het gewricht disloceren.

De modellparameters zijn verkregen in een uitgebreide kadaverstudie, waarin 14 schouders van 7 kadavers zijn gemeten. Voor het model zijn de parameters van één min of meer mediaan kadaver gebruikt. Met behulp van antropometrische metingen zijn de segmentstraagheden geschat. De fysiologische dwarsdoorsnede en het gewicht van de spieren zijn gemeten om de maximale spierkracht te schatten. De drie-dimensionale (3D) posities van de morfologische structuren zoals de aanhechtingen van spieren en ligamenten, gewrichtsvlakken, botcontouren, het scapulothoracale glijvlak en markante botpunten, zijn gemeten met de zogenoemde palpator, een instrument om ruimtelijke posities te meten. Door de meetpunten van de morfologische structuren zijn geometrische vormen geschat, die in het model geïmplementeerd kunnen worden. Er is een nieuwe theorie ontwikkeld om het mechanisch effect van spieren met grote aanhechtingsplaatsen te representen met behulp van meerdere werklijnen.

De posities van de botstukken van het schoudermechanisme zijn gemeten bij 10 proefpersonen tijdens onbelaste en belaste abductie en anteflexie van de humerus. De gemiddelde waarden zijn gebruikt als ingangsvariabelen voor het model. Omdat slechts statische posities gemeten konden worden, is het model ook slechts gebruikt voor berekeningen in statische situaties. De lengteveranderingen van spieren tijdens elevatie van de humerus zijn zeer groot vergeleken met de lengteveranderingen van beenspieren tijdens het lopen. Deze grote lengteverandering zou betekenen dat sommige spieren niet in staat zouden zijn om kracht uit te oefenen bij 180 graden elevatie van de humerus, wanneer uitgegaan wordt van een normale kracht-lengte relatie tussen 60% en 140% van de optimum spierlengte. Er is geconcludeerd dat de amplitude van het gerecificeerde en geïntegreerde EMG niet gebruikt kan worden om het model te valideren, omdat EMG lengteafhankelijk is en de kracht-lengte relatie van de spieren onbekend is. Er kunnen slechts aan/uit patronen vergeleken worden, wat niet voldoende is om onderscheid te maken tussen de vier optimalisatiecriterium. Spierfuncties zijn geanalyseerd door momentenbalansen rond de assen van de drie gewrichten van het schoudermechanisme op te stellen. In dat geval vormen de krachten in het scapulothoracale glijvlak onderdeel van de momentenbalans. M. serratus anterior oefent het grootste moment uit rond het sternoclaviculaire (SC-) en acromioclaviculaire (AC-)gewricht uit. M. trapezius dient voornamelijk om de clavícula zodanig te positioneren dat dit botstuk een steunpunt vormt voor de grote hefboom van de m. serratus anterior. Na de m. deltoideus bleek de m. subscapularis de tweede belangrijke abductor van het glenohumerale (GH-)gewricht. De spieren
van de rotatorenmanchet zijn aktief om de stabiliteit van het GH-gewricht te regelen door de resulterende spierkrachtvector te richten binnen het gewrichtsvlak van de cavitas glenoidalis. Het ligamentum conoideum vormt een tegenwicht voor de axiaal roterende krachten op de clavicula van de claviculaire delen van de m. deltoideus en m. trapezius. Door het ligamentum conoideum volgt de axiale rotatie van de clavicula de laterorotatie van de scapula, waardoor de rotaties in het AC-gewricht minimaal zijn. Tijdens abductie wordt het grootste moment rond de sagittale as van het SC-gewricht geleverd door de reactiekrachten van de thorax op de Angulus Inferior. De laterorotatie en protractie van de scapula resulteert in een stabiele positie, die weinig spierkracht vereist. Omdat er geen mono-articulaire spieren over het AC-gewricht lopen, is de verbinding tussen de mediale rand van de scapula en de thorax noodzakelijk voor de stabilité van de schoudergordel.

De drie-dimensionale posities van botstukken van het schoudermechanisme zijn gemeten bij 10 proefpersonen met behulp van een palpatietechniek, waarbij markante botpunten worden gepalpeerd en vervolgens gemeten met de palpator. Zo kan de positie van de botstukken worden gereconstrueerd. De nauwkeurigheid van de palpatiemethode is vergelijkbaar met algemeen gebruikte 3D film- of videomethoden. De rotatie van botstukken wordt beschreven met behulp van Euler hoeken. Het definieren van de bewegingen wordt bemoeilijkt door het ontbreken van een anatomische positie van de botstukken van de schoudergordel en door het ontbreken van goed gedefinieerde rotatieassen, zoals een flexie/extensie as of een abductie/adductie as. Een aantal 3D bewegingsdefinities is met elkaar vergeleken. Uiteindelijk zijn twee definities gekozen omdat ze de best interpreteerbare rotaties opleveren. In de eerste definitie worden de rotaties van de botstukken beschreven rond assen van het globale assenstelsel, uitgaande van een virtuele referentiepositie waarin het lokale assenstelsel van het botstuk samenvalt met het globale assenstelsel. In de tweede definitie worden de rotaties in de gewrichten beschreven rond de assen van het globale assenstelsel, uitgaande van de initiële (rust)positie van de botstukken (bij 0 graden humeruselevatie).

Het model van het schoudermechanisme is gebruikt om het effect van verschillende fixatiehoeken op de bewegingsruimte van de hand en op de maximale tilkrachten na een glenohumerale schouderarthrodese te analyseren. De bewegingen van de scapula zijn gemeten bij 10 proefpersonen, de gemiddelde fixatiehoeken zijn gemeten bij 18 patiënten en de morfologie van één kadaver is gebruikt. De maximale tilkracht kan worden uitgeoefend wanneer de scapula is gelateroroteerd en geprotraheerd. In die positie heeft de m. serratus anterior een grote momentarm. De gemiddelde fixatiehoeken (achtereenvolgens 60 graden endorotatie, 9 graden abductie en 13
graden anteflexie) resulteren in een bewegingsruimte van de hand in het midsagittale vlak. Tijdens het aandrijven van een rolstoel worden de grootste momenten uitgeoefend rond de gewrichten van het schoudermechanisme, maar er is slechts weinig bekend van de bijdrage van individuele spieren. Op een rolstoelergometer zijn de positie van de botstukken, uitwendige krachten en EMG gelijktijdig gemeten. Spierkrachtpatronen zoals die berekend zijn met het model kwamen redelijk overeen met de EMG-patronen. De spierkrachten en de momenten rond de gewrichtsassen zijn geanalyseerd. De mogelijkheden van metingen in dynamische omstandigheden worden besproken. De conclusie luidt dat een bruikbaar model van het schoudermechanisme is ontwikkeld. Modelsimulaties leveren voor de meeste spieren betrouwbare voorspellingen van hun functies op. Tevens kunnen de krachten in ligamenten en gewrichten worden berekend. Het model kan op een groot aantal gebieden toegepast worden, b.v. voor het verbeteren van de diagnose van schouderklachten door de normale situatie te bestuderen, en voor het voorspellen van het effect van operaties zoals het implanteren van een endoprothese. Op ergonomisch gebied kan het model de belasting op de morfologische structuren voorspellen, hetgeen gebruikt kan worden voor het verbeteren van het ontwerp van mens-machine interfaces zoals het aandrijvingsmechanisme van een rolstoel.
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Curriculum Vitae

I was born on the first of August, 1960 in Rijswijk, The Netherlands. I started my career in the state educational system in 1964 at the kindergarten, and passed through primary school from 1966 until 1972. After that I spent the years from 1972 until 1978 at the Lodewijk Makeblijde College in Rijswijk, and I was offered a gymnasium-B certificate. From 1978 till 1985 I studied at the Interfaculty of Physical Education (presently: Faculty of Human Movement Sciences) at the Free University in Amsterdam, and graduated in Functional Anatomy under the supervision of Prof.dr. R.H. Rozendal. In the period 1985 - 1986 I fulfilled my duty to the nation as a conscientious objector to the Military Service at the Man-Machine Systems Group, Lab. for Measurement and Control, Dept. of Mechanical Engineering and Marine Technology of the Delft University of Technology. During that time I cooperated in the development of an expert system for the diagnosis of brachial plexus lesions. From 1987 till 1990 I continued to work in the same research group as a PhD-student, but changed subject to the development of a dynamic model of the shoulder mechanism. The research was conducted under supervision of Prof.dr.ir. H.G. Stassen and Prof.dr. R.H. Rozendal. The results of this research are subject of this thesis. Presently I am employed as an assistant professor at the Man-Machine Systems Group at the Delft University of Technology.

Frans C.T. van der Helm