

A three-dimensional model of the shoulder and elbow.

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Abstract - A three-dimensional musculoskeletal model of the shoulder and elbow has been developed, in order to increase the functional insight in the mechanical behavior of the shoulder and neural control issues, to assist in diagnosis and treatment of shoulder disorders, and to estimate the load on morphological structures for prevention of injuries. In the model all bones and joints of the upper extremity, i.e. the shoulder girdle joints, glenohumeral, elbow and forearm joints, are represented, as well as the 31 muscles crossing the joints. Muscle dynamics are represented by a third-order muscle model. Morphological data have been obtained in cadaver studies. The model has an inverse dynamic mode, in which muscle forces are calculated from the recorded motion and external forces, as well as a forward dynamic mode, in which the resulting motion from the neural input to the muscles is calculated. A few examples for a standardized motion (abduction), clinical application (glenohumeral endoprosthesis) and ergonomic application (reaction forces in the glenohumeral joint) are demonstrated.

INTRODUCTION

In medical textbooks, the knowledge about the mechanical function of morphological structures of the shoulder is very limited. When the sources of this knowledge are traced back in history, most information appears to stem from anatomical studies of German anatomists around the turn of this century (e.g. Mollier, 1899; Fick, 1911). Mollier (1899) built a physical model of the shoulder girdle, in which the muscles were replaced by ropes connected to a 'keyboard'. The shortening range and moment arms of muscles could be calculated by moving the bones and measuring the excursion of the keys: inverse kinematic analysis. On the other hand, when the keys were pressed, a motion resulted: forward kinematic analysis. Basically, this is the knowledge in medical textbooks: if a muscle shortens during a motion, it is assumed to be active and contributing to the motion. The muscle function is called after the motion: abductors, levators, etc. However, a dynamic analysis is missing in which the muscle forces and other forces in the system are calculated. Especially, the function of poly-articular muscles is very difficult to assess. Muscle forces can not be measured directly. Therefore, musculoskeletal models are necessary to simulate motions, and calculate the forces required for those motions. In Delft, a three-dimensional model of the shoulder has been developed (Van der Helm, 1994^{ab}). Recently, this model has been extended with an elbow representation (flexion/extension, pro/supination). The goals of the shoulder-elbow model are threefold. In the first place, it serves to enhance the functional insight in the mechanics of the upper extremity, and the role of the various morphological structures. Next, some clinical questions about the diagnosis and

treatment of shoulder and elbow complaints can be investigated. In the third place, the loading of morphological structures during work, ADL and other activities can be calculated in order to relate the loading with persistent shoulder complaints. Subsequently, if it is revealed what is causing the high loads and/or what are the vulnerable structures, the tasks can be adjusted in order to prevent the complaints. In this paper, these three goals will be illustrated.

The shoulder and elbow model

For modeling the shoulder mechanism a finite element method is used which is implemented in the computer program SPACAR (Werff, 1977).

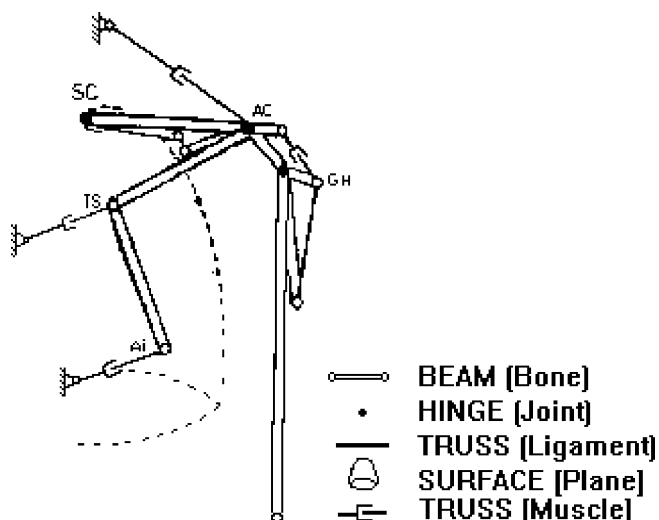


Figure 1 - Schematic view of the elements used for the shoulder model. Bones are represented by rigid bodies, joints by a combination of HINGE joints, muscles and ligaments by (curved) TRUSS elements, and the scapulothoracic gliding plane by a SURFACE element.

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 elements of which the kinematic and dynamic behaviour is known. Then, the behaviour of the complete mechanism can be calculated by simply connecting the elements. It is very easy to change the structure of the model, and/or to change parameters. In Figure 1 a basic sketch of the finite-element mechanism is shown.

In the finite element method each bony structure is represented by a multi-node MNODE element: a local co-ordinate system in which the muscle and ligament attachment sites and joint rotation centers are fixed.

The shoulder and elbow model consists of seven bony structures: the thorax, clavicle, scapula, humerus, ulna, radius and hand.

Two MNODE elements are connected by sharing one position node, i.e. the joint rotation center. The sternoclavicular, acromioclavicular and glenohumeral joints are represented as spherical joints by three orthogonal HINGE joints allowing all possible rotations. The elbow and forearm joint are each represented as a one degree-of-freedom HINGE joint. The wrist joint is represented as a spherical joint because there is not sufficient information about the location of the rotation axes. Most important of the wrist representation is to

have an accurate positioning of the application point of external forces at the hand. With the addition of 6 degrees of freedom for the thorax moving in space, the shoulder/elbow model contains a total of 17 DOF.

The scapula slides over the thorax over the so-called scapulothoracic gliding plane. 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 double connection of the scapula to the thorax (by the clavicle and the scapulothoracic gliding plane) results in a closed-chain mechanism: the motion constraints are limiting the potential orientations of the scapula, and the motion equations of the sternoclavicular and acromioclavicular joints are coupled through the reaction forces at the scapulothoracic connection. The motion constraints of the scapulothoracic gliding plane are represented by two SURFACE elements, each constraining one point of the medial border of the scapula to slide over the surface of the thorax, modeled as an ellipsoid.

There are three extra-capsular ligaments: the costoclavicular ligament crossing the sternoclavicular joint, and the coracoclavicular ligaments (the conoid and the trapezoid ligament) crossing the acromioclavicular joint. Furthermore, the sternoclavicular joint has a strong capsule, whereas the acromioclavicular joint capsule is relatively weak. The glenohumeral capsule is quite lax in order to allow for the large motion range in the glenohumeral joint. The glenohumeral capsule is strengthened by a number of ligaments. At the elbow joint a number of intra-capsular ligaments are present. However, one may assume that these intra-capsular ligaments are only stressed at the boundaries of the motion range, otherwise the muscles would have to overcome large passive forces during normal physiological movements. Hence, they can be modeled as a non-linear passive stiffness of the joint. The extra-capsular ligaments are represented by flexible TRUSS elements, of which the exerted force depends on the elongation (strain). However, since deformations of a few millimeters can result in large forces, the ligament forces are very sensitive to measurement errors of the recorded motion (in an inverse-dynamic analysis (Pronk et al., 1994)). Therefore, it is chosen to incorporate the conoid ligament as a rigid TRUSS element only, of which the forces only depend on the loading by the muscles.

Seventeen muscles are driving the motions of the shoulder, and seven cross the elbow joints. Only two mono-articular muscles are crossing the sternoclavicular joint: m. subclavius and the clavicular part of m. trapezius. Furthermore there are five bi-articular scapulothoracic muscles (mm. trapezius (scapular part), levator scapulae, rhomboideus,

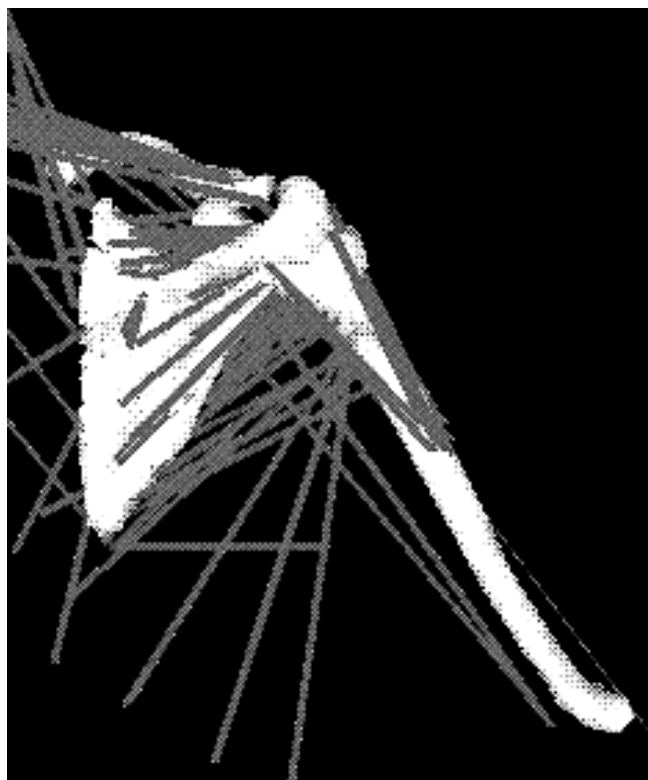


Figure 2 - *Ninety-five muscle elements of the shoulder model (visualization by SIMM).*

pectoralis minor and serratus anterior), two muscles are crossing all three joints (mm. pectoralis major and latissimus dorsi) and additionally nine muscles originate at the scapula and cross the glenohumeral joint (mm. deltoideus, teres major, subscapularis, supraspinatus, infraspinatus, teres minor and coracobrachialis), of which two muscles (mm. biceps and triceps (long head)) also exert an elbow moment. At the elbow joint, there are a few mono-articular flexor-extensor muscles (mm. brachialis, triceps (medial and lateral head), anconeus) and one mono-articular pro-supination muscle (m. pronator quadratus). Most muscles are of a mixed mono/bi-articular nature (mm. pronator teres, supinator) or are poly-articular (m. brachioradialis, wrist/finger flexors and extensors). 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 (e.g. thorax, humeral shaft, radius). 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, 115 muscle elements are used in the model to represent 24 muscles (m. subclavius is not included). In Figure 2 a number of muscle elements is shown.

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., 1991; van der Helm et al., 1992; Van der Helm & Veenbaas, 1991). Recently, elbow data have been measured (Veeger et al., 1997) and muscle optimum length parameters (Klein Breteler et al., 1997).

MODEL SIMULATIONS

SPACAR allows for the numerical evaluation of the motion equations in any position. In the inverse dynamic mode, the position, velocity and acceleration of the DOF, as well as external forces, are input variables to the model. Muscle forces can be calculated in an optimization procedure. Any optimization criterion can be chosen, but most often the sum of squared muscle activation α_i weighed by muscle volume V_i is used:

$$J = \sum_{i=1}^N V_i \cdot \alpha_i^2 = \sum_{i=1}^N V_i \cdot \left(\frac{F_i}{F_{i\max}(l,v)} \right)^2, \quad (1)$$

where:

N is the number of muscles,

F_i is the muscle force and $F_{i\max}(l,v)$ is the maximal muscle force as a function of the force-length and force-velocity relationship.

The reaction force in the glenohumeral joint is constrained to point from the rotation center inside the glenoid cavity, otherwise dislocation of the joint would occur. Several options are available for additional constraints. Muscle dynamics can be included using an inverse muscle model (Happee & Van der Helm, 1995). Quasi-static stability requirements, including a desired endpoint stiffness, can be included using an eigenvalue calculation of the joint stiffness matrix. For a full dynamic stability analysis, including reflexive feedback, the linearized state-equations are analyzed (Rozendaal, 1997). In Figure 3 a flow-chart of the inverse-dynamic simulation is shown.

In the forward dynamic mode, neural inputs to the muscles are the input variables. Muscle dynamics are represented by a third-order muscle model, with neural input and muscle length as input and muscle force as an output (Winters & Stark, 1985). The musculoskeletal model is driven by these muscle forces. Thus, the output variables are the motions of the mechanism. A severe problem of these simulations is the optimization of the neural input such that a desired motion results: The model requires 95 neural input signals to the muscles, parameterized in time, e.g. every 12 msec a value is to be calculated. For one second of simulation time this results in almost 8000 variables to be optimized!! Needless to say that even with the current computer power it is impossible to find an acceptable solution.

Output of the shoulder/elbow model

Once the muscle forces are known, all other forces in the mechanical linkage system can be calculated: ligament forces, joint reaction forces, compression forces at the scapulothoracic gliding plane. All mechanical variables in the model can be accessed and used for analysis. For post-processing of the data, two programs are used: Matlab (The Math Works, Inc.) and SIMM (Musculographics, Inc.). In Matlab all variables of interest can be viewed using a screen with pull-down menus. The variables are bone rotations, joint rotations, muscle lengths, muscle moment arms, muscle forces, forces in passive structures, joint reaction forces, and the moment equilibrium's around all joints. The pull-down menu structure is very user-friendly and robust. SIMM is being used as an visualization tool: the joint rotations are input for SIMM, and the resulting motion is shown to the user. Also the muscle elements can be shown on request. Muscle force magnitude is shown by the intensity of the muscle element. The user is able to change

his/her viewpoint, zoom in and out, and inquire about data. Figure 2 is made using SIMM.

Validation

In the strict sense, the shoulder/elbow model has not been validated since the predicted output (muscle forces) can not be recorded. Comparison with EMG measurements is difficult, since there is no absolute relation between EMG amplitude and force magnitude, for instance because the force-length and force-velocity characteristics are simply unknown. Van der Helm (1994^a) showed that most non-linear optimization criteria resulted in the same set of active muscles in each position, though magnitude of muscle forces were different. Then,

EMG measurements can not distinguish between these optimization criteria. EMG measurements can be used for validation in two other ways. In the first place, the time-patterns of EMG can be compared with the timing of muscle activation as predicted by the model (Happee & Van der Helm, 1995, unpublished wheelchair data). In the second place, the 'principal action' of the muscles can be determined with EMG, and can be compared with model predictions (De Groot, 1992). The results of both ways of force-EMG comparisons were reasonably good.

APPLICATIONS OF THE MODEL

Applications of the shoulder part of the model thus far have included standardized arm motions (abduction, forward flexion; Van der Helm, 1994), analysis of a glenohumeral arthrodesis (Van der Helm & Pronk, 1994), goal-directed arm motions (Happee & Van der Helm, 1995), manual wheelchair propulsion (Van der Helm & Veeger, 1996) and sensitivity analysis of an glenohumeral endoprosthesis design and operation technique (De Leest et al., 1996). A few examples will be shown.

Functional insight in shoulder mechanics.

In a palpation experiment, the 3D rotations of the shoulder bones have been recorded during abduction and forward flexion (Van der Helm & Pronk, 1995). These rotations were used as input for simulations with the shoulder model (Van der Helm, 1994^b). In Figure 3 the moment equilibrium about the ventral/dorsal axis of the sternoclavicular joint is shown during abduction from 0° to 180°. M. serratus anterior is the main muscle to exert a moment opposing the external weight, though the largest moment is due to the reaction force between Angulus Inferior and the thorax. It is striking that m. trapezius does not exert a major moment about this axis during abduction. It is concluded that the combined lateral rotation and protraction movement of the scapula results in a energetically very efficient position to counterbalance the external moments.

CLINICAL APPLICATIONS.

The functional results of a glenohumeral endoprosthesis are not very good. Usually, the elevation angles after surgery are limited to 90°. In a sensitivity analysis the effect of design parameters (radius and thickness of glenoid component, radius and thickness of humeral component) and operation technique parameters (orientation and positioning of glenoid and humeral component) on the required muscular effort was investigated (De Leest et al., 1996). Therefore, these parameters were altered in the model, affecting the joint rota-

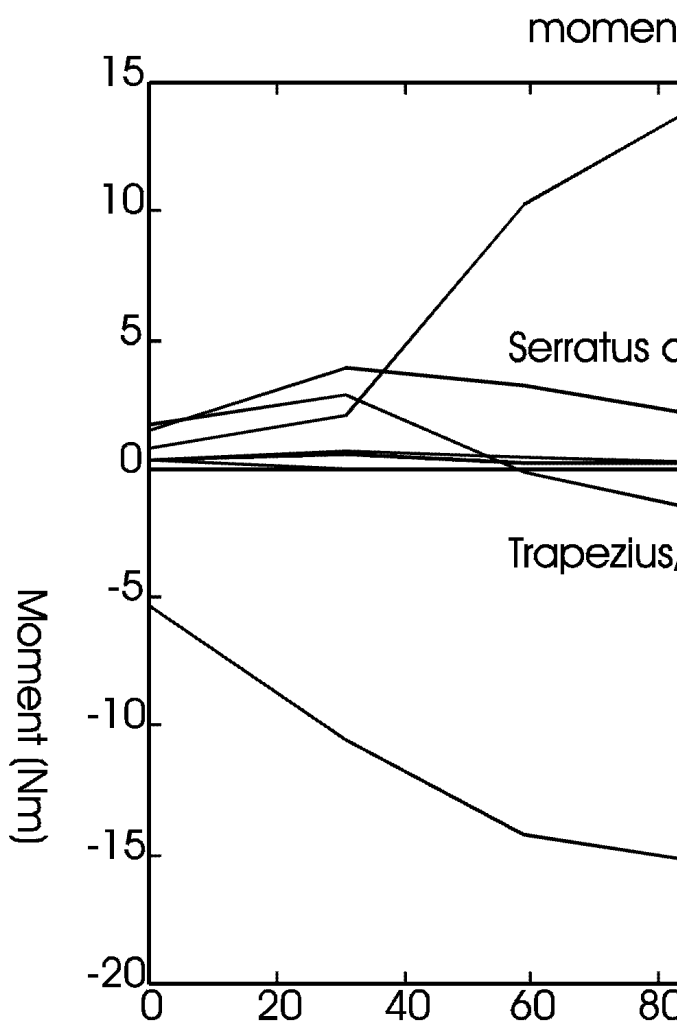


Figure 3 - The moment equilibrium about the ventral/dorsal axis of the sternoclavicular joint during abduction. The external moment is due to the weight of the arm, counterbalanced primary by the reaction forces at Angulus Inferior in the scapulothoracic gliding plane, and the mm. serratus anterior and trapezius.

tion center, muscle moment arms and wrapping of the muscles. It was concluded that medialization of the rotation center and posterior/superior redirection of the glenoid orientation was favourable.

Loading of morphological structures.

The shoulder is a very complicated 3D mechanism, in which some smaller muscles are very important to maintain stability, i.e. to compensate for unwanted moments about secondary axes by larger muscles. Consequently, the additional muscular effort for stabilization will result in higher compression forces at the joints. The internal loading of some specific structures may be much higher than an analysis based on the net moments would have revealed. If these structures are vulnerable, shoulder complaints may occur.

In Figure 4 the magnitude of the glenohumeral joint reaction force is shown during a (quasi-static) wheelchair push. The loading of the joint may be as high as almost three times body weight. Another propulsion technique, or redesign of the wheelchair might be necessary to decrease this joint reaction force.

CONCLUSIONS

With the shoulder/elbow model a very powerful tool is developed to analyze the mechanical function of the upper extremity. Also the control efforts of the Central Nervous System (CNS) can be

analyzed, on a level of complexity which somewhat approaches the control problem that faces the CNS. The quality of predictions of the shoulder model is considered to be very good, though a strict validation cannot be done.

REFERENCES

Fick R (1911). *Handbuch der Anatomie und Mechanik der Gelenke*. Teil 3, Gustav Fischer, Jena.
 De Groot JH, Van der Helm FCT, Arwert HJ (1992). The effect of the direction of arm force on muscle forces and the configuration of the shoulder mechanism. *Abstr. 8th Meeting of the European Society of Biomechanics*:78.
 Happee R, Van der Helm FCT (1995). The control of shoulder muscles during goal directed movements. An inverse dynamic analysis. *J. Biomech.* 28(10):1179-1191.
 Klein Breteler MD, Spoor CW, Van der Helm FCT (1997). Measuring muscle and joint geometry parameters for a shoulder model. In: Veeger HEJ, Van der Helm FCT, Rozing P (ed.), *Proc. 1st Conf. of the International Shoulder Group*, Delft, The Netherlands.
 De Leest O, Rozing PM, Rozendaal LA, Van der Helm FCT (1996). The influence of glenohumeral prosthesis geometry and placement on shoulder muscle forces. *Clin. Orthop. Rel. Res.* 330, 222-233.
 Mollier S (1899). *Ueber die Statik und Mechanik des menschlichen Schultergürtels unter normalen und pathologische Verhältnissen*. Festschr. f. C. v. Kupffer, Jena.
 Pronk GM, Van der Helm FCT, Rozendaal LA (1994). Interaction between the joints in the shoulder mechanism: The function of the costoclavicular, conoid and trapezoid ligaments. *J. Engineering in Medicine (Proc. Instn. Mech. Engrs)* 207, 219-229.

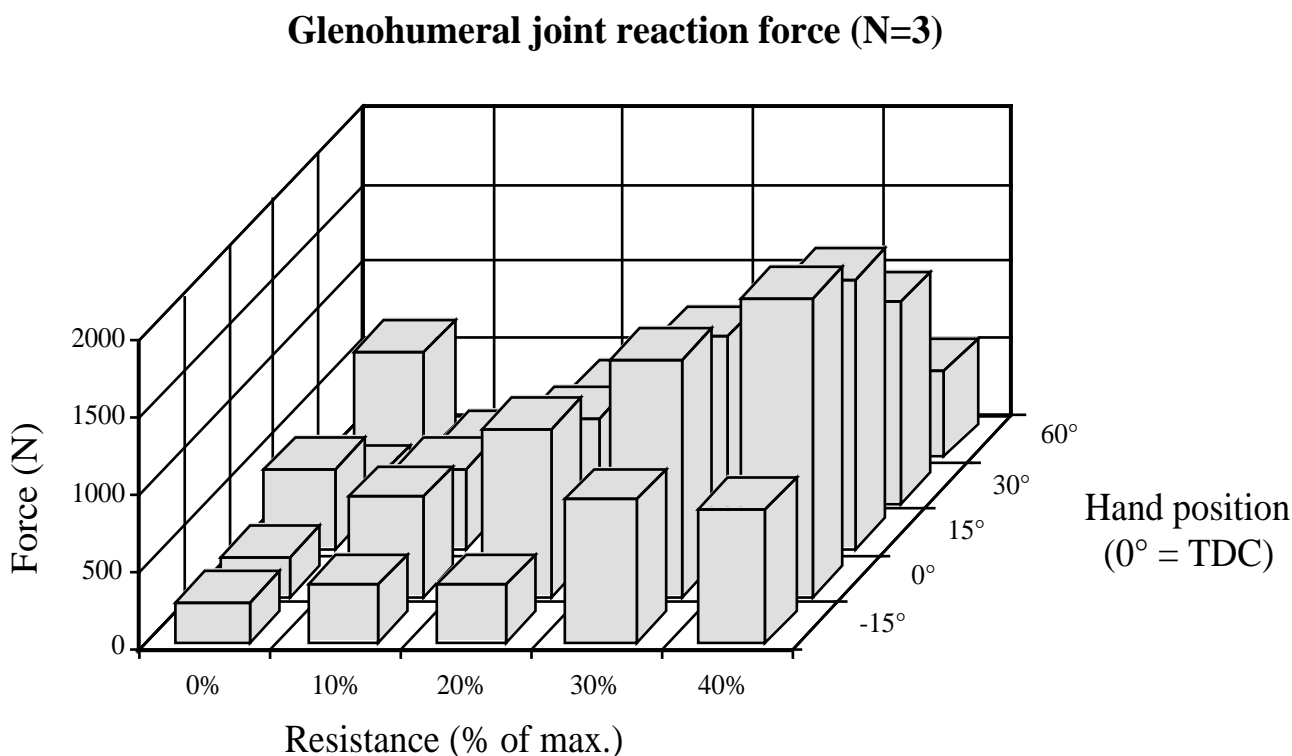


Figure 4 - Glenohumeral reaction forces for different load levels and hand positions: quasi-static calculation.

- Rozendaal LA (1996). The reflexive control of the shoulder muscles .In: Veeger HEJ, Van der Helm FCT, Rozing P (ed.), *Proc. 1st Conf. of the International Shoulder Group*, Delft, The Netherlands.
- Van der Helm FCT, Pronk GM, Veeger HEJ, Van der Woude LHV, Rozendal RH (1992). Geometry parameters for musculoskeletal modelling of the shoulder mechanism. *J. Biomech.* 25, 129-144.
- Van der Helm FCT, Veenbaas R (1991). Modelling the mechanical effect of muscles with large attachment sites: application to the shoulder mechanism. *J. Biomech.* 24(12), 1151-1163.
- Van der Helm FCT (1994). A finite element musculoskeletal model of the shoulder mechanism. *J. Biomech.* 27(5), 551-569.
- Van der Helm FCT (1994). Analysis of the kinematic and dynamic behavior of the shoulder mechanism. *J. Biomech.* 27(5), 527-550.
- Van der Helm FCT, Pronk GM (1994). The loading of shoulder girdle muscles in consequence of a glenohumeral arthrodesis. *Clinical Biomechanics* 9, 139-148.
- Van der Helm FCT, Pronk GM (1995). Three-dimensional recording and description of motions of the shoulder mechanism. *J. Biomechanical Engineering* 117(1), 27-40.
- Van der Helm FCT, Veeger HEJ (1996). Quasi-static analysis of muscle forces in the shoulder mechanism during wheelchair propulsion. *J. Biomech.* 29(1):39-52.
- Veeger HEJ, Van der Helm FCT, Van der Woude LHV, Pronk GM, Rozendal RH (1991). Inertia and muscle contraction parameters for musculoskeletal modelling of the shoulder mechanism. *J. Biomech.* 24(7), 615-629.
- Veeger HEJ, YU B, An KN, Rozendal RH (1997). Parameters for modeling the upper extremity. *J. Biomech.* (in press)
- Werff K van der (1977). *Kinematic and dynamic analysis of mechanisms: A finite element approach*. PhD thesis, Delft University of Technology, The Netherlands.
- Winters JM, Stark L (1985). Analysis of fundamental human movement patterns through the use of in-depth antagonistic muscle models. *IEEE Trans. Biomed. Eng.:* 32(10), 826-839.