Modelling compensatory postural muscle use in subjects with a thoracic spinal cord injury

Willems, M.M.M.

Award date:
1997
Modelling compensatory postural muscle use in subjects with a thoracic spinal cord injury

M.M.M. Willems

Master's thesis

Supervisors: L.H. Braak and A. Huson (TUE)
             H.A.M. Seelen and Y.J.M. Potten (iRv)
             R. Happee (TNO, TUE)

Eindhoven, University of Technology, Faculty of Mechanical Engineering/
Hoensbroek, Institute for Rehabilitation Research

Eindhoven, September 1997
Contents

Summary 3

1 Introduction 4
   1.1 Problem 4
   1.2 Goal 4
   1.3 Preview 4

2 Posture and Movement Research 5
   2.1 Introduction 5
   2.2 Description of the reaching tasks 7
   2.3 Experimental results 9

3 Modelling methods 12
   3.1 Introduction 12
   3.2 Available models, described in literature 12
      3.2.1 GEBOD 12
      3.2.2 Shoulder model with muscle elements 14
      3.2.3 Human model with multisegment spine 16
   3.3 Models used for the present study 19
      3.3.1 Simple model 19
      3.3.2 Detailed model 20

4 Results 23
   4.1 Simple model 23
   4.2 Detailed model 24
      4.2.1 N-model 26
      4.2.2 L-model 27
      4.2.3 H-model 28
      4.2.4 Comparison between ΔCP and ΔCG 29
   4.3 Effect of muscles 29
   4.4 Complete simulation 32

5 Conclusions and discussion 34
   Recommendations 36

References 37

Appendices 2
   A Anatomy and Spinal cord injuries
   B Example of MADYMO data file: the GEBOD model
   C Muscle element in MADYMO
Summary

Spinal cord injured (SCI) subjects suffer from function loss of muscles, caudal to the injury. During rehabilitation, new compensatory postural strategies have to be learned, using muscles which are still controllable, but which normally have a different function. The therapy during rehabilitation is based on experience. The question arose whether the alternative use of muscles to control posture is indeed as efficient as assumed. This is one of the main fields of interest in the Posture and Movement Research at the Institute for Rehabilitation Research. Experiments have been carried out in which EMG data and 3D data were recorded while the seated subject had to perform balance perturbing forward reaching tasks. The experiments included subjects with a complete high thoracic SCI (H-subjects), subjects with a complete low thoracic SCI (L-subjects) and subjects without a SCI (N-subjects). N-subjects can reach further than L- and H- subjects, and primarily use the erector spinae (ES) to restore sitting balance. In L- and H- subjects, the ES has partially lost its function, and these subjects use the latissimus dorsi (LD) and the ascending part of the trapezius (TPA) more than N-subjects to maintain sitting posture. L-subjects extend the upper part of the spine and move the head backwards to counteract the forward displacement of the upper limbs during the aforementioned reaching. H-subjects keep a kyphotic sitting posture and reveal only movement of their upper limbs.

In this thesis, a computer model is described with which moments in all vertebra joints of the spine, due to gravity and muscle activity, can be calculated. With the model, three different sitting postures of the N-, L- and H-subjects, as recorded in the experiments, are simulated.

Firstly, the effect of gravity on the spinal moments was studied. The kyphotic sitting posture of the H-subject resulted in a favourable ‘line of moments’ across the spine. The strategy of the L-subject, as implemented in the model, turned out to be effective in terms of minimizing moments in the spine.

Secondly, the effect of muscle activity of the right LD and right TPA in the model of the H-subject was studied. It turned out that the TPA, as modelled, is able to counteract the influence of gravity on sitting posture, in contrast to the LD. However, both muscles have a larger effect in maintaining lateral spinal stability.

By simulating both gravity and LD and TPA activity in the H-model, the equilibrium position (i.e. resultant moments of 0 Nm) could not be reached. As modelling errors are excluded, more anatomical structures or external influences must play a role.

The current model is only a first step towards an answer to the aforementioned question. It certainly has the potential to become a powerful tool; a number of recommendations for improvement are made.
1 Introduction

1.1 Problem

During their rehabilitation, (thoracic) spinal cord injured subjects have to learn new strategies for postural control and restoration of balance, because they suffer from loss of muscle function, caudal to the injury. The combination of anatomical position and cervical innervation of the latissimus dorsi and the trapezius pars ascendans makes that these muscles seem to be suitable for partially compensating the function loss of the erector spinae. This alternative postural muscle use has been investigated by the Posture and Movement Research (P&MR) group, at the Institute for Rehabilitation Research. A biomechanical model could serve to further contribute to the understanding of the musculoskeletal system and to corroborate the experimental findings. One of the main questions to be answered is: Is the alternative use of the aforementioned muscles indeed as efficient as assumed from a biomechanical point of view? Possibly, in future, predictions can be made with the model about the biomechanical efficiency of (alternative) postural muscle use during and after the rehabilitation of paraplegic subjects. Besides, it could be valuable in ergonomical research (interaction with seating components, task performance, etc.).

1.2 Goal

The goal of this thesis is the development of a biomechanical computer model with which compensatory postural motor function of thoracic spinal cord injured subjects can be described, based on the experimental data from several projects in the P&MR.

1.3 Preview

Preliminary work for this project, i.e. a literature review and finding a suitable modelling method as well as a suitable modelling package, has already been described in an earlier report [Willems, 1996]. The present report concerns the simulation of high, low and non-SCI subjects in several sitting postures, recorded in the P&MR experiments. In chapter 2, the experiments and the results of the P&MR are described. The 3D data from this research serve as input for the model that had to be developed. Two models were created with the computer package MADYMO: a simple model and a detailed model. The latter model was based on a combination of two already existing models, i.e. a model of a complete human body and a model of the shoulder. The models are described in chapter 3 and the results are presented in chapter 4. Finally, in chapter 5, the conclusions are discussed and several recommendations for further improvement of the current model and for future research are made.
2 Posture and Movement Research

2.1 Introduction

The Lucas-Foundation for Rehabilitation is one of the two Cooperating Rehabilitation Centres Limburg (SRL), Rehabilitation Centre 'Franciscusoord', focussed on the treatment of children, being the other one.

The following institutes are part of the Lucas-Foundation:

'Revalidatiecentrum Hoensbroeck' arose in the seventies from a former nursing home. Now, patients with a spinal cord injury, CVA (stroke), amputation, or rheumatic disorder are treated. The rehabilitation team consists of a rehabilitation specialist, nurses, physiotherapists, occupational therapists, a social worker, a psychologist, and sometimes a speech therapist. This team can be extended with other disciplines, if necessary.

Regular activities in the daily schedule of a patient are therapies and exercises, like ADL training (Activities of Daily Living), physiotherapy, occupational therapy, sports, etcetera.

'Audiologisch Centrum Hoensbroeck' supports children and adults with hearing impairments and linguistic and speech problems.

'Instituut Arbeidsexploratie Hoensbroeck' provides counseling regarding educational and working possibilities for subjects with a handicap, with (re-)entering in the employment process as the final goal.

'Arbeidsintegratie Hoensbroeck' is a nationwide operating institute, which offers practically oriented professional trainings for people with a physical handicap.

Several (private) enterprises, which are oriented towards technical supplies for the handicapped (such as prosthesis, orthopaedic footwear, and also car adaptations) which have to lead to more independency, are also situated on the premises of the SRL.

Tasks from the foundation are:
- treatment of patients
- promotion of rehabilitation in general
- education of professional workers
- research in rehabilitation

The foundation cooperates with the Institute for Rehabilitation Research (iRv). The iRv was founded in 1982 by the University Maastricht, the SRL and the Dutch Organization for Applied Scientific Research (TNO). The iRv has two core activities: applied research and dissemination of information. 'Information and Dissemination Services' (DID) within the iRv collects, stores, handles, and supplies information in the field of rehabilitation and care for subjects with a physical handicap. The centre is not only oriented towards the professional worker in rehabilitation, but also towards the handicapped themselves.

The mission of the iRv is to contribute to the improvement of interventions, health and social services and assistive technology for people with long-term functional impairments.
and handicaps. Activities take place at regional, national and European level. The iRv has a staff of approximately 80 from a variety of disciplines: medicine, social sciences, movement sciences, occupational therapy, engineering. The work programme distinguishes four main areas:

- effectiveness of rehabilitation processes
- co-ordination of services
- usability of assistive devices
- information and dissemination

Two specific fields within the work programme 'effectiveness of rehabilitation processes' are of interest: chronic pain and fear of movement, and the development of compensatory muscle use in patients with a spinal cord injury (SCI). This second area is called the Posture and Movement Research (P&MR). Within the P&MR, several projects are carried out, aimed at posture control and control of arm-hand function:

- reorganisation of postural control in SCI subjects during and after their rehabilitation
- task performance related to sitting balance in paraplegic subjects
- development of an arm-hand function task for the evaluation of arm movements in tetraparetic subjects
- effect of orthoses on paretic muscles.

The present report is based upon the research concerning postural control. The goal is to design a computer model describing alternative postural muscle use in thoracic spinal cord injured subjects, as studied by Seelen et al. [1997]. During rehabilitation, new patterns of postural control, involving parts of the sensorimotor system which are still intact, have to be acquired by the SCI patient. This implies compensation of function loss and reduction of disabilities rather than restoration of function. Although a great amount of clinical experience about therapy outcome has been acquired, many of the mechanisms by which movement therapy might influence reorganisation of motor control and more specifically postural control in SCI patients are still largely unknown. Future research will be focussed on the development and improvement of methods for therapy-evaluation.

Subjects without disorders in the sensorimotor system can automatically maintain sitting balance. In subjects with a SCI, this automatism is impaired. Most studies about 'sitting' are of an ergonomic nature and focus on chair adaptations. Little research into (disturbed) postural control during sitting has been done. From several studies [Seelen et al., 1997] it appears that subjects with a high thoracic SCI compensate the loss of muscle function by using non-postural muscles. Non-SCI subjects primarily use their erector spinae (ES) to maintain (sitting) posture. Seelen and coworkers hypothesised that seated thoracic SCI subjects stabilise the trunk and the spine relative to a punctum fixum as near as possible to the base of support, i.e. (externally) the seat and (internally) the pelvis and the adjacent part of the spine, by using the latissimus dorsi (LD) and trapezius pars ascendens (TPA). The anatomical position and cervical innervation of the LD, with its insertion in the thoracolumbar region, and the TPA, with its insertion on the low thoracic vertebral spinous processes, seem to be both biomechanically and neurologically favourable to this concept.

In order to prevent retraction of the scapulae by LD and TPA activity, forces protracting the scapulae should be generated either passively, caused by e.g. gravity during kyphotic
sitting, or actively using muscles still controllable like the pectoralis major (PM) and the serratus anterior (SA).

Main questions addressed in this part of the P&MR are:
• To what extent can subjects with a thoracic SCI maintain their sitting balance?
• Does an increase in activity from the PM and SA occur simultaneously with an increase in LD and TPA activity?
• What is the relation between the pelvic movement, together with the movement of the lower part of the vertebral column, and maintaining sitting balance?

In the aforementioned research a set of reaching tasks is used with which sitting posture is perturbed. Compensation of ES function loss in spinal cord injured subjects was investigated in an experiment in which 15 non-SCI subjects (N-group), 15 subjects with a complete low thoracic SCI at spinal level T9-T12 (L-group) and 15 subjects with a complete high thoracic SCI at spinal level T2-T8 (H-group) participated. Subjects from the N-group will be referred to as N-subjects, from the L-group as L-subjects and from the H-group as H-subjects. The T2 level was chosen to be the upper limit of the SCI, because arm-hand function of the subject is still intact then. In a subject with a SCI below the T12 level, the muscles of the pelvic region and the upper leg can be actively controlled and a major part of the ES is still functioning. See Appendix A (Anatomy and Spinal cord injuries). SCI subjects had completed their active rehabilitation process at least one year before.

2.2 Description of the reaching tasks

The subjects were seated behind a table in a chair, fixed on a force platform. A central start button (SB) was placed on the edge of the table. Also eight target buttons were placed in pairs at locations marking 15%, 30%, 75% and 90% of the individual unsupported maximal reaching distance of the participant, see figure 2.1. This meant that, although absolute button distance could vary between subjects and between groups, relative button distance was kept constant for all subjects for all groups. The 90% reaching distance was chosen to avoid complete balance loss in SCI subjects during task performance. A clear distinction was made between relatively easy coverable reaching distances (15 and 30%) and more difficult coverable distances (75 and 90%).

The subjects were asked to perform reaching movements in the sagittal plane towards the target button pairs, introducing graded systematical sitting balance perturbation. The movement included the displacement of the arms and, if possible, the trunk. During initial sitting position the start button was pressed with both hands. Information about which button pair the subject had to reach to, was presented on a monitor in front of the subject. Then, the subject had to move both hands as fast as possible towards the correct buttons and press them 5 times. This caused a prolonged period in which maintaining sitting balance was thought to be more difficult, especially in SCI subjects. Subsequently the subject had to move back to his initial sitting position as fast as possible and press the start button again. In the experiments, centre of pressure changes and activity of the LD, TPA, PM, SA, ES
and oblique abdominal muscles during task execution were measured. Since non-SCI subjects differ from SCI subjects in their ability to tilt their pelvis during sitting, also the correlation between pelvic movement and postural changes during task execution was investigated by recording 3D motions of the subject.

![Figure 2.1: Experimental set-up [Seelen et al., 1997]](image)

Centre of pressure changes were deduced from the force platform data. Muscle activity was recorded using surface electromyography (EMG). Kinematic data were recorded with the PRéCiSion Motion Analysis System (PRIMAS). Four cameras were positioned in a semi circle around the test setting. These cameras send infrared light which is reflected by retroreflective markers, attached to anatomical landmarks on the femur, the pelvis and the lower part of the spine. In an additional experiment, also data from the upper part of the body of a non-SCI subject were recorded. In this way, movements of the subject can be monitored. The position of the cameras was dictated by the range of the movement; each marker had to be visible for at least two cameras at the same time when recording the movement. The reconstruction and subsequent analysis of the PRIMAS data is very time consuming. Initially, each marker will be given a number. Then, the markers can be tracked automatically, for all four cameras. But when the computer cannot identify a marker, because it crosses another marker or it disappears, marker tracking has to be performed manually. After the marker identification, the 3D reconstruction can be performed and the spatial displacement of the markers can be calculated.

Two main types of data comparison were used: conditions in which relative reaching distance between groups was the same, and conditions in which absolute reaching distance between groups was equivalent.
2.3 Experimental results

Subjects with a (high) thoracic SCI tilt their pelvis towards the backrest of the chair, to compensate for the instability of the pelvis and the lower part of the spine. In this way, the punctum fixum is raised, and more cranial muscles can be active in keeping the trunk erect. During the tasks, the pelvis is kept in this position, see figure 2.2.

![Figure 2.2: Displacement of markers on the pelvis](image)

N-subjects produce more pelvic tilt and flexion of the trunk compared to L- and H-subjects. L-subjects exposed an extension of the upper part of the spine. H-subjects revealed hardly any movement.

EMG results indicate that H-subjects use their LD, TPA, PM and SA more in situations of similarly perturbed sitting balance than N- and L-subjects. The results also show an increase in the use of the high thoracic part of the ES in the H-group, see figure 2.3.
Differences are more diffuse, when the relative disturbances are considered. Since N-subjects can reach further than subjects with a SCI, they develop more phasic muscle activity during, for example, the stretching of the arm.

Figure 2.3: EMG data of the subjects [Seelen et al., 1997]

Gross kinematic differences occur between seated thoracic SCI and non-SCI subjects in pelvic tilt, spinal movement and upper body movement during task execution.

Centre of pressure (CP) data indicate a high inter-subject reproducibility. The range in which N-subjects were able to shift their CP was significantly larger than for SCI subjects. Although CP displacement increased simultaneously with reaching distance in all groups, results indicate that this increase was less in SCI subjects. Results and statistical analysis also show significant differences between the L and the H-group, see figure 2.4.
From the smaller CP displacements for subjects with a SCI, it appears that decreased sitting balance and posture control can be compensated only partially by the alternative strategy.

**Figure 2.4:** CP displacement vs. reaching distance [Seelen et al., 1997]
3 Modelling methods

3.1 Introduction

To corroborate the earlier results from the Posture and Movement Research (P&MR), as described in chapter 2, a biomechanical computer model of the spinal system during the reaching tasks (section 2.2) may be of help. With the multibody computer package MADYMO, in which muscle elements are available, a first start towards such a model is made. With this model, it is possible to calculate resultant loads in the spine for subjects from the different groups (section 2.1) in several static situations during the task execution. For the moment, dynamic situations are not considered because of a lack of input data and because of the complexity.

The moment that can be generated by the muscles which are implemented in the model, can also be calculated.

MADYMO is a general applicable software package which can be used for simulating the dynamic behaviour of (bio)mechanical systems. MADYMO can combine the multibody-technique with the finite element method, so many kinds of systems can be analysed [MADYMO manuals, 1996]. In the present study, only the multibody-technique is used.

MADYMO offers specialised preprocessor programs, like the GEnerator of BOdy Data (GEBOD). GEBOD generates geometric data for adults as well as for children. See section 3.2.1. Standard databases are also available, like 3D dummy databases. Dummies represent the human behaviour in car crash tests, which are often simulated with this package.

To get an impression of the acting moments, a simple model has been created in the present study with GEBOD, see section 3.3.1.

For a more detailed model, two already existing models could be used:
- **SPACAR 3D model of the shoulder mechanism (developed at Delft University of Technology) including muscle elements**, see section 3.2.2.
- **MADYMO 3D human model with a multi-segment spine (developed at TNO Road-Vehicles Research Institute)**, see section 3.2.3.

The combination of these models will be described in section 3.3.2.

3.2 Available models, described in literature

3.2.1 GEBOD

GEBOD was originally developed and documented by Baughman [1983]. With the MADYMO version of GEBOD inertia and geometry properties of humans can be generated in 2 or 3 dimensions. Several subject types are available: a child (2 - 19 years), an adult human male and an adult human female. After selecting the subject type, one or more of the following parameters can be specified: the subject’s weight, height, and, in case of a child, his age. Weight and height values can be given in English units, in SI units or in percentiles.
The model of the human body is also referred to as 'system'. A system in MADYMO can be divided in segments (bodies), connected to each other by kinematic joints. In the case of the human body, examples of segments are the head, the upper arm, and the foot; examples of joints are the elbow and the hip. Only systems with a so-called tree structure can be modelled, see figure 3.1. Closed chains cannot be modelled, in other words: only one way from segment \( i \) to segment \( j \) is allowed.

![Figure 3.1: System with a tree structure](image)

GEBOD generates two files: a (part of the) MADYMO input file and a file that contains tables of the computed body dimensions.

The automatically generated part of the input file contains information about:

**CONFIGURATION** Specification of the branches in the system. Branches are built up of segments. Every segment is given a number. Number 1 is always the reference body. The reference body is either connected to the inertial space, or the motion is prescribed relative to the inertial space.

**GEOMETRY** Every segment has a local coordinate system, with its origin in the joint with the previous segment in that branch. The coordinates of the center of gravity of the segment and the coordinates of the connection with the next segment are given in this local coordinate system.

**INERTIA** Mass and moments of inertia of every segment.

**ELLIPSOIDS** Ellipsoids, and also planes, cylinders and facet surfaces, can be attached to a body to represent its shape. These surfaces are also used to model contact with other bodies. It is possible to attach several ellipsoids to one segment. Semi-axes, coordinates of the centre, and the degree of the ellipsoids must be specified.

This file has to be extended with other information, about e.g. the type of the joints, external forces, and the orientation of body segments. The additional input for the present study will be discussed in section 3.3.1.
3.2.2 Shoulder model with muscle elements

Van der Helm [1991] created a finite element model of the shoulder mechanism, using the computer package SPACAR, see figure 3.2. SPACAR is suited for simulating the kinematic and dynamic behaviour of finite element mechanisms. At each position of the mechanism, motion equations are numerically derived based on deformations of the elements and displacements of its nodes. Contact between elements is obtained by letting them share common nodes.

![Shoulder model of Van der Helm](image)

**Figure 3.2:** Shoulder model of Van der Helm [1991]

The shoulder model was originally developed in order to predict optimal fusion angles for a glenohumeral arthrodesis, to improve the application of an endoprosthesis, and to develop therapies for patients with a habitual glenohumeral subluxation. The goal is to gain insight into the function of morphological structures. Each relevant morphological structure has been represented by an appropriate element. The model consists of four bony structures, three joints, three ligaments, a scapulothoracic gliding plane and twenty muscles and muscle parts, divided over 95 muscle elements. Among these there are the latissimus dorsi and the trapezius, which are also of great interest in the Posture and Movement Research.

The parameters needed for the model were obtained from both shoulders of 7 cadavers:

- location of origin and insertion of all muscles
- shape and position of bony contours determining a muscle path
- location of the attachments of the ligaments
- shape and position of the scapulothoracic gliding plane
- position of well-defined bony landmarks

**Coordinates**
Spatial coordinates were measured by a palpator, a spatial digitizer which consists of four links connected by four hinge joints. Rotations of these joints are recorded and, hence, the position of the endpoint of the final link can be calculated. The 3D coordinates of the datapoints were defined with respect to a global coordinate system, with its origin at the
incisura jugularis.

**Muscles**

For an adequate representation of muscle action, the following items must be considered:

- 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 is 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. 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, then that muscle by itself can exert a moment on a bone, see figure 3.3. Clearly, such a muscle should not be represented by just one force vector.

![Figure 3.3: Representation of the trapezius with several force vectors](Van der Helm, 1991]

Different parts of one muscle can be functionally independent, i.e. one part can be active while another part is inactive.

In order to represent the mechanical effect of large muscles adequately, the complete attachment sites of origin and insertion, and the distribution of muscle bundles in between, have to be mathematically described.

- For several muscles, the muscle line of action cannot be described by a straight line from origin to insertion, but should be defined around a bony contour. If the shape and the position of this bony contour are known, i.e. are 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.

- The magnitude of a muscle force depends on a number of characteristics, e.g. the force-length and force-velocity relations, the architecture of the muscle, and the physiological cross-sectional area (PCSA).

**Joints**

In the finite element model, the sternoclavicular joint, the acromioclavicular joint and the glenohumeral joint are represented as ball-and-socket (spherical) joints. The shoulder
complex is modelled as an open chain. During the studied motions of the humerus, it is assumed that the scapula is always connected to the thorax. Therefore, the scapulothoracic 'joint' is represented by two elements, each constraining a point of the medial border of the scapula to the thorax (trigonum spinae and angulus inferior). The scapulothoracic gliding plane has been modelled as an ellipsoid.

**Simulations**

For a number of cadavers, intraindividual (left-right) differences were even as large as interindividual differences. No further comparisons were discussed and the author has chosen to present the parameters describing the geometry of one median cadaver. The complete parameter set consists of inertia, geometry and muscle parameters. This allows the positioning of muscle force vectors, and calculation of coordinates and moment arms in any position of the shoulder.

Simulations are inverse dynamic:

Input variables are the prescribed (measured) position, velocity and acceleration of the shouldeergirdle and the humerus, and the external load on the humerus.

Output variables are muscle forces and muscle stresses.

**Optimization criteria**

Around a joint, more muscles are present than strictly necessary to control the degrees of freedom: several muscle control schemes can result in the same motion. Motions and EMG patterns are highly reproducible within one subject; it is not likely that muscles are activated coincidentally. Since the physiological way in which muscle forces are distributed over the system is still unknown, an adequate strategy of muscle force calculation is also unknown.

Four different optimization criteria were compared: (a) minimization of the sum of squared muscle forces, (b) minimization of the sum of squared muscle stresses, (c) minimization of the sum of squared muscle forces, normalized to the maximal muscle force, and (d) minimization of the maximal muscle stress in the entire mechanism.

Criterion (a) gives incorrect results, because it does not account for the muscle stress, denoted by the PCSA. Criteria (b), (c) and (d) do not show much differences. Considering the computational efficiency, (b) is preferable.

### 3.2.3 Human model with multisegment spine

Van den Kroonenberg et al. [1997] developed a 3D mathematical model of a seated car occupant to better understand the biomechanical response of the spine and the interaction with the seat during rear-end collisions. Special attention is paid to the modelling of the spine, including the neck, and the interaction with the seat. To obtain insight into its biofidelity, the model's response is compared with rear-end sled experiments. Especially the kinematics of T1 (displacements and accelerations) were of interest, since they serve as input for dynamic models of the neck (for studies on whiplash). The T1 kinematics can be considered a result of several phenomena: lifting of the pelvis from the seat cushion, elongation of the spine, and a change in the shape of the spine.

Starting point for the human model was the 50th percentile male HYBRID III dummy for
frontal car crashes. A database of this dummy is standard available in MADYMO and a short description will be given below.

50th percentile sitting HYBRID III dummy
The HYBRID III is a widely applied dummy for frontal car crashes [MADYMO Database Manual, 1996]. The size and weight of the 50th percentile male HYBRID III represent an 'average' of the USA adult male population. It consists of 32 bodies. The external geometry of the database is described by 51 ellipsoids. Ellipsoids need to be specified for visualisation of bodies and for contact interactions; it is possible to represent one body by several ellipsoids.

![Figure 3.4: Model of the 50% sitting HYBRID III](image)

The reference position of the dummy is the sitting position, see figure 3.4. To simulate other positions of the dummy or other orientations of the dummy segments, the description of the joints should be modified. In the database, abdomen compression is included to describe the deformation in response to lap belt loading, and separate rib and sternum bodies have been specified to describe the stiffness of the ribs and the compliant sternal region. A non-linear stiffness depending on the bending direction has been implemented in the neck model.

The databases were developed and validated using component tests and tests on the complete dummies.

Multisegment spine
In the 50th percentile male HYBRID III dummy database, Van den Kroonenberg replaced the bodies representing the complete spine and the head by a more detailed representation, using several anthropometric sources, a human body model with multisegment spine (up to T2) and a head-neck model (starting with T1), see figure 3.5.

The spine model consists of 16 bodies representing the lumbar and thoracic vertebrae. The intervertebral discs are incorporated in these bodies. The vertebrae are connected to each other by revolute joints (hinges); only motions in the sagittal plane are of interest.
Although the spine model is in principle a 3D model, only two degrees of freedom were used: flexion/extension and axial elongation/compression. Stiffness and damping were defined for each of the joints to represent the soft tissue and muscles acting at the joints.

The so-called 'global model' developed by De Jager (1996) was used to represent the head and neck. The segments (head, seven cervical vertebrae, first thoracic vertebra) are connected by 3D non-linear visco-elastic elements with load-displacement characteristics derived from recent experimental data on cervical motion segment behaviour. For simplicity, also in this model, muscle behaviour is lumped into the intervertebral joint stiffnesses. The head-neck model was validated by comparison with the response of human volunteers to frontal impacts.
3.3 Models used for the present study

For the present study, two models are created. The first is a simple model, based on GEBO. The other is a detailed model, based on a combination of the models of Van der Helm (shoulder with muscles) and Van den Kroonenberg (human model with multisegment spine).
With both models, moments in the spine can be calculated. With the detailed model, muscle activity can be simulated.

3.3.1 Simple model

A model is generated of a 50% adult human male (length 1.77 m, mass 78.8 kg), see figure 3.6. Male data sets are based on the 1980 male stereophotometric study of McConville et al. [1980] and the 1967-68 Air Force flying personnel anthropometric study of Grunhofer [1975].

Figure 3.6: Reference position of the GEBO model

To be able to position the system and to calculate the parameters of interest, the input file (section 3.2.1) had to be extended with:

<table>
<thead>
<tr>
<th>GENERAL INPUT</th>
<th>Simulation time, integration method.</th>
</tr>
</thead>
<tbody>
<tr>
<td>INFORMATION</td>
<td></td>
</tr>
<tr>
<td>JOINTS</td>
<td></td>
</tr>
</tbody>
</table>

A kinematic joint constrains the relative motion of a pair of bodies. Different types of joints allow different relative motions. For example, a revolute joint only allows a rotation around one axis, a spherical joint allows arbitrary rotations, a translational joint only allows translations in one direction.
INITIAL CONDITIONS  Initial positions, orientations and (angular) velocities of all the segments in the system.

FORCE MODELS  An acceleration field (gravity) is defined as a function of time by means of function pairs.

OUTPUT CONTROL PARAMETERS  With 'constraint loads', the resultant forces and moments in the specified joints can be calculated.

For the complete input file, see appendix B.

Resulting forces and moments are calculated (both analytically and numerically) for the joints in the spine for two postures: standing erect and sitting forward bended. The standing position slightly differs (i.e. the position of the arms) from the original position of GEBOD, the sitting position is simulated to gain insight in the load of the spine during the experimental conditions. But this model is too global to study the difference between the subjects from the three groups. Therefore, a more detailed model had to be created, which will be described in the next section.

In literature, often erect or forward bended (lifting) postures are modeled to study the load of the lumbar spine, like for example the multiple-linkage model of Chaffin and Andersson [1991], which is also a very global model and comparable to the GEBOD model.

3.3.2  Detailed model

3.3.2.1  Conversion of the SPACAR shoulder model to MADYMO

A later version of the SPACAR shoulder model, extended with the forearm and hand, is converted into a MADYMO model. The specification of the system in MADYMO differs from that in SPACAR. All coordinates in SPACAR are defined with respect to one global coordinate system with its origin in the incisura jugularis. In MADYMO, the coordinates of the joint are defined with respect to the previous body; the coordinates of the centre of mass and muscle insertions are defined in the body local coordinate systems.

Further, the definition of the different kinds of elements differs. For example, a muscle element in SPACAR is defined by its PCSA, two attachment points, and, in case of a curved muscle, also the geometrical structure of the underlying bone. A muscle element in MADYMO is defined by at least two attachment point, several parameters, like the active state and the reference length, and length- and velocity dependent functions (based on Hill’s equations), see appendix C (Muscle element in MADYMO).

The ligaments and the degrees of freedom of the scapula are not (yet) included in the MADYMO model. The ligaments in the shoulder are not of interest for the present study. If any ligaments would be considered then it would be the ones in the spine. The 'joint' between the scapula and the thorax is very complex: it allows combined translation and rotation in a curved plane. Therefore, it is very hard to model in MADYMO, and for this study no attempt was made. This needs to be done in future research. The other joints are, like in SPACAR, represented as spherical joints.
The MADYMO version of the model of the right shoulder was scaled to fit the human model with a factor 0.9340, based on the y-coordinates of the glenohumeral joint. This version consists of the thorax, clavicula, scapula, humerus, ulna, radius and hand and several (not all) muscles and muscle parts. When a muscle in SPACAR was divided in five elements, in MADYMO the same five elements were defined. In the original shoulder model, muscles are on one side attached to the thorax and on the other side to either the scapula, clavicle or humerus. The thorax is not divided in vertebrae and no further data are available of the vertebral column or the position of the pelvis.

3.3.2.2 Combination of human model and shoulder model
In the multisegment spine model, which has already been created in MADYMO, firstly the right shoulder and arm were replaced by the complete shoulder model with the muscle elements. Thoracal muscle insertions were transformed from the rigid thorax to the separate (vertebral) bodies, using anatomical atlases [Kendall et al., 1971, Platzer, 1989], see figure 3.7. Of the right arm, only the bony structures are depicted, to visualize the muscle insertions. Therefore it looks thin.

Figure 3.7: Detailed model in the original position
The posture of the model, seated in a car, had to be adapted to an erect sitting posture. This mainly has consequences for the orientation of the spine and the legs. Hardly any information was found on the vertebral column of an erect sitting subject, accurately describing the position and orientation of all vertebrae relative to each other. As a result, the only data that could be used in the computer model, were the 3D marker data of the pelvis area with additional visual indications, and information about the absolute reaching distances, recorded during the P&MR research. Data of the positions of the upper part of the spine, the head, and the arms during the task were available for only one healthy subject.

The erect sitting posture differs for all subjects. But there are similarities between the subjects from the same group. For example, H-subjects sit in a kyphotic position. From each group, the data of one 'typical' subject in 3 positions (erect sitting, 30% and 90% reaching position) had been used for the computer simulations. Unfortunately, the coordinate system from the experimental camera set up did not have the same origin and orientation as in MADYMO, so it had to be translated and rotated. After that, the format of the data file had to be adapted into a MADYMO data file. Then, the experimental markers were visualised at the same time with the model of the human body, which could now be fitted.

Weight and height of the subjects were not taken into account, otherwise three different models had to be created. By using one and the same model for all subjects, the effect of a different posture is clear.

Spinal stiffness and damping coefficients, representing the muscles and the soft tissue, are left out of consideration. These parameters are of no relevance, because only static situations are considered for the moment. For dynamic simulations, which are more complicated, more data have to be acquired.

Besides, the aforementioned representation of muscle action cannot even be applied in a model of an SCI subject. The contribution of other anatomical structures (like ligaments) is not clear. Therefore, the model was simplified to rigid structures and muscle elements. Where Van den Kroonenberg is interested in the kinematics of the spine, for this study, the kinematics are prescribed, and the load of the spine is of interest.
4 Results

4.1 Simple model

With GEBOD, a simple computer model was created of an average adult male, as described in section 3.3.1. In order to get an impression of the load of the spine during different tasks, two postures were simulated: standing erect and sitting forward bended, see figure 4.1. The loads were caused only by gravitational forces. The acting moments around joints J2 in the 'spine' and J6 and J8 in the shoulders were calculated using MADYMO.

Figure 4.1: Postures of the simple model

The model was compared to the coplanar multiple-linkage static model for symmetric sagittal plane activities while lifting [Chaffin and Andersson, 1991, pp. 190-195], see figure 4.2. They consider the human body as a set of 6 links, which are connected to each other by the joints (ankle, knee, hip, shoulder, and elbow). In their model, both arms and both legs are represented as one arm and one leg. Different postures were described by varying the angles between the links. For this model, average anthropometric male data were used, but these are not specified. At each joint in the system, the load moments and joint reactive forces are determined in sequence, starting from the elbow (see figure 4.2).

The results of two joints from both the GEBOD model and the model by Chaffin and Andersson are given in table 4.1. The models can be compared because, for the moment around the considered joints, the position of the legs has no influence: it makes no difference whether a standing or a sitting posture is simulated. The only difference is that
Chaffin and Andersson model both shoulders in the same joint, so for the moment in one shoulder, the given value should be divided by two.

![Image of Chaffin and Andersson model](image)

**Figure 4.2:** Postures of the model of Chaffin and Andersson [1991]

<table>
<thead>
<tr>
<th>position</th>
<th>$M_x$ (Nm)</th>
<th>$M_{12}$ (Nm)</th>
<th>$M_2$ (Nm)</th>
<th>$M_{26}$ (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chaffin and Andersson</td>
<td>14.6</td>
<td>19.7</td>
<td>14.6</td>
<td>3.4</td>
</tr>
<tr>
<td>GEBOD</td>
<td>121.0</td>
<td>110.6</td>
<td>14.6</td>
<td>8.1</td>
</tr>
</tbody>
</table>

The moments are not exactly the same for both models. The (small) differences may be caused by differences in anthropometry and posture of the upper part of the body of the models.

When the system is in equilibrium, these are also the moments which have to be generated by one or more muscles around the joints, to lead to resultant moments of 0 Nm. Muscle activity, however, will be discussed in section 4.3.

### 4.2 Detailed model

A second computer model of the human body was created in MADYMO. This model was composed of a model of the right shoulder and a model of the human body with a detailed spine, as described in section 3.3.2.

With this model, the bending moments in all vertebral joints can be calculated. The model was fitted as best as possible to the experimental 3D data of one 'typical' subject from each group (the groups are described in section 2.2). Differences in weight and height were not taken into account and the movement of the subjects was left out of
consideration. For each of the three subjects, three static sitting postures were simulated:
- erect (with the hands at the start button)
- 30% forward reaching position
- 90% forward reaching position

In total, nine sitting postures were simulated with the detailed model. In the following, the three postures of the N-subject, simulated with the detailed model, are referred to as N-model; the postures of the L-subject as L-model, and the postures of the H-subject as H-model.

Positions of the vertebrae, the arms and the head had to be changed for every position. For the N- and the L-model in the 90% reaching position, also the positions of the pelvis and the legs were changed. Changes in the position of the upper body and the legs started from the pelvis. Subsequently, the orientation of every segment had to be adapted.

Changes in the position of the arms only started from the glenohumeral joint. Unfortunately, no data of the motions of the clavicle and the scapula were available. The shoulder girdle in the human body is a closed chain, which cannot be modelled. In the SPACAR version, the shoulder was modelled as an open chain from the thorax via the clavicula and the scapula to the humerus. Motion of all structures was included. However, the motion of especially the scapula could not be transformed to MADYMO. In the present model, the scapula and the clavicula remained fixed to the thorax. This fixation may lead to errors in the position of the arm and, as a consequence, to errors in the acting moments around the vertebral joints. Furthermore, the force distribution by the scapula and the clavicle over the thorax is not conform reality because of this fixation. For the levels between T1 and T8 (the area where the aforementioned bony structures are situated), the calculated moments may contain errors. 3D data of the orientation of these bony structures in the shoulder have to be acquired to be able to improve this part of the model.

In the original shoulder model of Van der Helm [1991], over 20 muscles were included, divided over 95 muscle elements. These were also implemented in the right shoulder of the detailed model of the human body. Among these are also the muscles which are of great interest in the P&MR (LD, TPA, PM and SA).

By splitting the simulation with the detailed model into three steps, the separate effect of (a) gravity against (b) muscle activity can be studied. When (c) these two effects are added, a resultant forward bending moment of 0 Nm in every vertebral joint should be obtained, i.e. when the contribution of other anatomical structures or external influences is left out of consideration.

In a first series of computer simulations, the muscles were not activated: only the load of the spine, due to gravity, was studied. The results of the N-, L- and H-model will be given in sections 4.2.1 to 4.2.3.

In a second series of simulations, the muscles in the computer model were fully activated while gravity was eliminated. In section 4.3 the results are given of those cases in which the right LD and the right TPA were activated in the H-model in the 90% reaching position. The H-model was used, for the simple reason that H-subjects suffer most from muscle function loss, so the less muscles play a role, which makes this initial model not too complex.

Finally, in section 4.4, the results are given of the simulation of the H-model in the 90% reaching position with both gravity and muscle activity. Also the effects of the left LD and left TPA were included.
4.2.1 N-model

Figure 4.3 gives the results of the moments in the spine for the different postures of the N-model. The leap at level T1/T2 is caused by the arms which are attached to the trunk at this level. During erect sitting and reaching to 30%, the moment is nearly constant across the spine, i.e. approximately 34 Nm and 51 Nm respectively. Evidently, the bending in the spine is small. The difference in the resulting moments between erect and 30% is constant at all levels (17 Nm), because only the position of the arms is different in these situations.

In the 90% reaching condition, the moment increases linearly from 41 Nm at level T1/T2 to 133 Nm at level L5/S1. Both the considered mass and the corresponding moment arm increase when the forward bending moment around a more caudal vertebral joint is calculated.
4.2.2 L-model

Figure 4.4 shows the results of the different postures of the L-model. Around the thoracic vertebrae, the forward bending moment in the erect posture is nearly constant: about 27 Nm, which is less than in the N-model. Around the lumbar vertebrae, the forward bending moment decreases to about 3 Nm. Because of the loss of muscle function, of muscles controlling the lumbar spine, a smaller forward bending moment in this part of the spine is favourable. In the L-model, only the position of the arms was changed to simulate the 30% reaching posture, which led to an increase of 7 Nm in every joint. This increase is also less than in the N-model, because of the difference in reaching distance.

There is hardly any increase in moments when reaching to 90% in the L-model, compared to 30%, except for the lower lumbar vertebral joints. When the L-subject has to reach further, he counteracts the forward displacement of the upper limbs by extending the spine and moving the head backwards [Potten et al., 1997]. This can also be seen in figure 4.4. In this way, the center of gravity is kept within the base of support.
4.2.3 H-model

Figure 4.5 shows the results of the different postures of the H-model. Starting at level T1/T2, the forward bending moment first increases; then, from level T11/T12, it decreases.

Figure 4.5: (a) H-model; erect position and 90% reaching position are depicted
(b) Bending moments in all vertebral joints of three postures (i.e. erect sitting, 30% and 90% reaching position) are presented

This is caused by the kyphotic sitting posture, which the H-subject maintains. The decrease can be explained by the decrease in moment arm. The difference between erect and 30% and erect and 90% is constant: approximately 5 Nm and 16 Nm respectively, because here, only the position of the arms differs between these situations.
4.2.4 Comparison of ACP and ACG

To validate the detailed model so far, more experimental data are required. However, a comparison of the experiments and the numerical simulations can be made: there is a relation between the centre of pressure (CP) from the experiments and the centre of gravity (CG) of the upper part of the body in the model (calculated by dividing the forward bending moment around L5/S1 by the mass above that level). They cannot be compared directly, because the CP also takes inertial effects into account and because of the absence of the effect of the legs in the calculation of the CG as used in the model. However, changes in CP and CG can be correlated. During task execution, the subjects were required to press the peripheral buttons five times, thus introducing a short, quasi-static period. The CP signal hardly changes at that moment. In Table 4.1, ΔCP and ΔCG between the erect position and both reaching positions are given. It is assumed that the contribution of the legs in ΔCP and in ΔCG is negligible, because the legs hardly move.

Table 4.1: ΔCP (from the P&MR experiments) and ΔCG (calculated from the forward bending moments around L5/S1)

<table>
<thead>
<tr>
<th>Group</th>
<th>Reaching distance (%)</th>
<th>Reaching distance (cm)</th>
<th>ΔCP (cm) (experimental)</th>
<th>ΔCG (cm) (calculated)</th>
</tr>
</thead>
<tbody>
<tr>
<td>H-group</td>
<td>30</td>
<td>6.5</td>
<td>2</td>
<td>1.3</td>
</tr>
<tr>
<td></td>
<td>90</td>
<td>19.5</td>
<td>5</td>
<td>4.3</td>
</tr>
<tr>
<td>L-group</td>
<td>30</td>
<td>9.0</td>
<td>2</td>
<td>1.9</td>
</tr>
<tr>
<td></td>
<td>90</td>
<td>27.0</td>
<td>8</td>
<td>6.8</td>
</tr>
<tr>
<td>N-group</td>
<td>30</td>
<td>26.3</td>
<td>6</td>
<td>4.7</td>
</tr>
<tr>
<td></td>
<td>90</td>
<td>79.0</td>
<td>23</td>
<td>26.7</td>
</tr>
</tbody>
</table>

It should be noted that no adaptations were made in the model for anthropometry of the subjects. This, together with possible errors in the modelled posture, may be responsible for the differences.

4.3 Effect of muscles

In this section, the results of the second part of the simulations with the detailed model are presented. For the H-model in the 90% reaching position, the activity of the right LD and right TPA were simulated. Muscle data were based on Van der Helm [1991]. This concerns the number of elements per muscle, the coordinates of the origin and the insertion of each element and the maximal isometric force \( F_{\text{max}} \) of each element. In the original SPACAR model, the LD was curved around a geometrical structure. In MADYMO, the curvature had to be modelled by specifying extra attachment points between the origin and the insertion. Otherwise, the muscle elements would go right across the thorax. For the present model, this specification has been an iterative process. For example, one element of the LD has its origin on vertebra T12 and two extra attachment points on the ribs of T8 and T7 (the ribs are assumed to be rigidly connected to the vertebrae). Another
element of the LD has its origin on L2 and the attachment points on T10 and T8. The line, starting from the origin of T8 and going to the first and then to the second attachment point on T8, could be interpreted as (a part of) rib T8.

Figure 4.6: Right LD and right TPA; rear view and side-view of the H-model in the 90% reaching position

When both the muscle element and the rib looked right, the process was stopped. That the curvature of the LD has an effect on the forward bending moment in the vertebral joints is shown in figure 4.7. The influence is most present in the low thoracic and lumbar part of the spine. There, the forward bending moment is smaller in the case of the curved muscle, and the area where a backwards bending moment can be generated, is larger.

In figure 4.8, the results are shown of the cases in which the right TPA, divided in three elements, and the right LD, divided in five curved elements, were activated in the model of a H-subject in the 90% forward reaching position, while gravity was eliminated. All elements in one muscle generated the same $F_{max}$, i.e. approximately 200 N for the TPA elements, and approximately 150 N for the LD elements [Van der Helm, 1991]. Both the forward and the lateral bending moments are shown in figure 4.8.
Figure 4.7: Effect of the curvature of the LD

Figure 4.8: Moments, generated by the LD and the TPA in the model of a H-subject in the 90% reaching position (right bending is positive)
The TPA generates backward bending moments between T1/T2 and T9/T10, thus being able to counteract gravity. In the same part of the vertebral column, the LD generates only forward bending moments, while only in the lower part backward bending moments are generated. Both muscles have a much larger effect in the lateral direction.

Because of the fixation of the humerus, the scapula and the clavicula to T1 in the present model, the value of the bending moments in the upper part of the thoracic spine (T1-T8) might be wrong. In reality, forces are distributed over the thorax, especially by the scapula.

When also the left LD and the left TPA would be activated, the forward bending moment would double, while the lateral bending moment would become zero.

### 4.4 Complete simulation

Finally, both muscle activity and gravity were simulated, and the forward bending moments in the spine were again calculated for the H-model in the 90% reaching position, see figure 4.9.

![Figure 4.9: Forward bending moments in case both muscle activity and gravity are simulated (Nm)](image)

At most levels, the resultant moment has only increased due to muscle activity, compared with the case that only gravity was active. This result was to be expected, because it is a matter of linear superposition. Varying some muscle parameters (like $F_{\text{max}}$) could lead to
different results. However, the activity of only the LD and the TPA is not enough to reach an equilibrium position; more anatomical structures and external influences must play a role.
5 Conclusions and discussion

The GEBOD model is, because of its limited data set, a good model to start with. It is easy to understand; results can be checked analytically. The simple model was compared with a model of Chaffin and Andersson [1991]. The bending moments in the spine and in the shoulder were of the same order in both models for two different postures. The (small) differences are probably the result of differences in anthropometry and posture of the upper part of the body.

Simulations with the detailed model were split into three steps: (a) only gravity, (b) only muscle activity and (c) both gravity and muscle activity.

In the first series of computer simulations with the detailed model, the muscles were not activated: only the load of the spine, due to gravity, in different sitting postures was studied. N-subjects move the whole upper part of their body during reaching, while H-subjects only move their arms. H-subjects also differ from N-subjects in their ability to tilt the pelvis. To compensate for the instability of the pelvis and the lower part of the spine, H-subjects tilt their pelvis backwards, thus assuming a kyphotic sitting posture [Potten et al., 1997]. N-subjects keep their spine more erect.

In the N-model (section 4.2.1), the moments in the erect and in the 30% reaching position are nearly constant over the spine. Evidently, the bending in the spine is small. In the 90% reaching position, the moment increases linearly over the spine towards a value of about 130 Nm around L5/S1. This is of the same order as the moment around J2 in the GEBOD model (section 4.1). In the erect position, the moment is almost twice as large in the N-model. The masses of the upper parts of both models cannot account for this difference: in the simple model it is about 39 kg, in the detailed model it is about 37 kg. The observed difference is most likely caused by the forward position of the head and the upper part of the spine. It can be concluded that the GEBOD model supports the detailed model.

In the L-model (section 4.2.2), the forward bending moment over the spine in the erect posture is smaller than in the N-model, because of the forward position of the head and the corresponding forward position of the cervical and high thoracic spine of the N-model. In the lumbar part, the bending moment decreases further. When reaching to 90%, the subject moves his head backwards to compensate for the forward stretching of his arms. This hardly leads to an increase in load, compared with the 30% reaching position. It can be concluded that this is a successful strategy.

Given the fact that postural muscle function loss is most present in the lower part of the spine, the H-subject has no alternative but to (passively) tilt his pelvis backwards. The resulting kyphotic sitting posture in all positions, simulated with the H-model (section 4.2.3) leads to a 'curved' line of moments across the spine, with a maximum near T7/T8 and a minimum at L5/S1. From this model, it can be concluded that this is a favourable sitting posture for the H-subject.

The detailed model was fitted as best as possible to the available 3D data (i.e. marker positions and reaching distances) from the P&MR experiments. Since these data only were
complete for one N-subject, it is possible that the postures of the H- and L-model differ from reality.

In the second series of simulations, the muscles in the computer model were fully activated while gravity was eliminated. This was done to gain insight in the moments that can be generated by the muscles. It was expected that the muscles would be able to counteract the calculated moments from the first series of simulations (gravity) in all vertebral joints.

N-subjects primarily use their ES to control sitting posture while leaning forward. In spinal cord injured patients, the ES has (partly) lost its function. During rehabilitation, these subjects are trained to use non-postural muscles like the LD and the TPA to compensate the function loss of the ES. It is studied to what extent those muscles can take over the function of the ES. In section 4.3 the results were given for those cases in which the LD and the TPA were activated in the H-model in the 90% reaching position.

Both muscles have larger effects in the lateral direction than in the frontal direction. It should be noted, that the moments in the part of the spine where the scapula and the clavícula are effective (T1-T8), are not correct, because of the fixation of both structures to T1 in the model. In reality, both the scapula and the clavícula can move, and forces are therefore distributed over the thorax and thus over the vertebrae. This is also the part of the spine where the largest bending moments occur. Given the further values, it is not expected that the sign of the moments is wrong, but the calculated values may be too large. Varying some muscle parameters (like $F_{\text{max}}$) could lead to different results. Below level T8, the fixation has no influence.

However, in the present model, the function of the LD during forward bending seems to be very small. But as a consequence of the aforementioned error (and possibly other modelling errors), the function of the LD could be underestimated.

In the final simulations, both gravity and muscle activity were present. At most levels, the resultant moment has only increased due to mainly LD activity, compared with the case in which only gravity was active. To reach an equilibrium position, more anatomical structures and/or external influences must play a role.
Recommendations

• For validation and further improvement of the present model, more experimental data should be acquired, especially concerning the position and orientation of body segments and bony structures.

• Attention should be payed to the scapulothoracic joint. The combined translation and rotation of the scapula on the curved thorax makes it a very complex joint. For the present study, there was no time left for improvement.

• More trunk muscles should be implemented in the model and when the shoulder joints are improved, also the effect of the shoulder and arm muscles, which are already implemented, can be studied. The curvature of muscles plays an important role.

• One of the most important external influences, as mentioned above, is most likely the backrest of the chair. SCI subjects need a backrest for support, but its contribution is unknown.

• By adapting a part of the data file of the present model, dynamic situations can be simulated. For the moment, however, the additional value will be small. Improvement of the model is most important.
References


Appendices
A Anatomy and Spinal cord injuries

The spine consists of 7 cervical, 12 thoracic, and 5 lumbar vertebrae, and the sacrum. A vertebra consists of a vertebral body and an arch with facets. The vertebral bodies with the intervertebral discs form the so-called anterior column, which transfers axial compression forces. The arches with the facet joints form the posterior column. The canal, which is located between the arches, protects the spinal cord; the foramina give way to the spinal nerves.

A spinal cord injury (SCI) results in an interruption of the ascending and descending tracts. This implies motor loss, sensory loss, and vegetative disorders, distal from the SCI. The most frequent cause of a SCI is a traffic or diving accident, but it can also be caused by an infection, a haemorrhage, or a tumour. The extent of the sensorimotor function loss depends on the level in the spinal cord and on the completeness of the lesion. The more caudal the lesion, the less the functional impairment.

A SCI is named after the last intact spinal cord segment. Spinal cord segments do not correspond with the vertebral body, see figure A.1. When the injury is caudally from the first thoracal segment, there are disorders in the trunk and the lower extremities. When the damage is cranial to that level, there are also disorders in the upper extremities.

**Figure A.1:** Relationship of vertebra, spinal cord and spinal nerves

The rehabilitation of a SCI subject requires a multidisciplinary approach. Both rehabilitation approach and functional outcome depend on the level of the SCI, the subsequent function loss and residual capacities/potential of a patient.
Simpel model om momenten in joints te kunnen berekenen.

Augustus 1997

GENERAL INPUT

TO 0.0000
TE 0.0000
INT RUKUS
TOL 1.0E-4
TSMAX 2.0E-4
RAMP 0.0000 0.1000
RACO 0.0100 0.1000

END GENERAL INPUT

SYSTEM

zithouding

CONFIGURATION

12 11 10 1
15 14 13 1
16 7 6 3 2 1
17 9 8 3 2 1

END CONFIGURATION

GEOMETRY

!adult male, 50% standing height

zithouding

!1:5

0.112E+02 0.988E-01 0.906E-01 0.114E+00
0.229E+01 0.156E-01 0.860E-02 0.234E-01
0.239E+02 0.476E-02 0.350E-00 0.295E-00
0.102E+00 0.170E-02 0.200E-02 0.250E-02
0.429E+01 0.204E-01 0.233E-01 0.155E-01

!6:10

0.190E+01 0.125E-01 0.132E-01 0.250E-02
0.134E+01 0.840E-02 0.850E-02 0.120E-02
0.190E+01 0.125E-01 0.132E-01 0.250E-02
0.134E+01 0.840E-02 0.850E-02 0.120E-02
0.957E+01 0.297E-01 0.297E-01 0.170E-02

!11:15

0.377E+01 0.573E-01 0.582E-01 0.660E-02
0.945E+00 0.800E-03 0.420E-02 0.440E-02
0.957E+01 0.297E-01 0.297E-01 0.170E-02
0.502E+00 0.110E-02 0.110E-02 0.400E-03
0.502E+00 0.110E-02 0.110E-02 0.400E-03

END INERTIA

ORIENTATIONS

3 2 1 3.0 0.000000
5 2 1 1.0 0.069813
12 2 1 1.0 0.069813
15 2 1 1.0 -0.069813
16 2 1 3.0 -0.244346
17 2 1 3.0 0.244346

JOINTS

!chair/lower torso
1 %RAC
!lower torso/spine
2 REVO
!spine/upper torso
3 REVO
!upper torso/neck
4 REVO
!neck/haad
5 BRAC
!shoulder left
6 SPHE
!elbow left
7 REVO
!shoulder right
8 SPHE
!elbow right
9 REVO
!hip left
10 SPHE
!knee left
11 REVO
!ankle left
12 BRAC
!hip right
13 SPHE
!knee right
14 REVO
!ankle right
15 BRAC
!wrist left
16 BRAC
!wrist right
17 BRAC

END JOINTS

INERTIA

END GEOMETRY
Muscle element in MADYMO

A muscle element in MADYMO consists of a chain of connected muscle segments. The two end points of a segment are called attachment points. At least two attachment points per muscle are required. These attachment points can be connected to bodies. The intermediate attachment points are frictionless sliding points through which the muscle can pass. In this way, the curvature of a muscle can be simulated.

The muscle element is based on Hill's description and consists of a contractile element CE and a parallel elastic element PE, see figure C.1. CE describes the active force, generated by the muscle. PE describes the elastic properties of mainly muscle fibers and the surrounding tissue. The forces on the segmented muscles are based on the total lengthening and the lengthening velocity of the whole muscle. The muscle tension force is the same for all of the segments. The muscle force acting on an attachment point is the vectorial sum of the tension forces acting on that attachment point.

On both sides, a muscle element could be expanded with other MADYMO-elements such as lumped masses, which describe muscle mass, and series elastic elements, which describe the elastic properties of the tendons and the aponeurosis.

Muscles are visualized as ellipsoids with a circular cross-section (of which the radius can be specified) and with a constant volume.

The muscular tension force (figure C.1) is expressed as:

\[ F = F_{CE} + F_{PE} \]  

where:

\[ F_{CE} = A \cdot F_{\text{max}} \cdot f_H(v_r) \cdot f_L(l_r) \]  

\[ F_{PE} = F_{\text{max}} \cdot f_P(l_r) \]

- \( A \) = active state signal (see section C.2)
- \( F_{\text{max}} \) = force exerted with maximal activation in isometric conditions \((v = 0)\) and at the reference length
\( f_{il}, f_L, f_p = \text{length- and velocity dependent functions of CE and PE (see section C.1)} \)

\( I_r = \text{dimensionless muscle length:} \)

\[
I_r = \frac{1}{I_{ref}} \tag{4}
\]

\( I_{ref} = \text{reference length (optimum), at which active force generation is most efficient} \)

\( v_r = \text{dimensionless lengthening velocity of the muscle (} v_r \text{ is positive if the muscle lengthens):} \)

\[
v_r = \frac{v}{v_{maxA}} \tag{5}
\]

where:

\[
v_{maxA} = f_V(A) \cdot v_{max} \tag{6}
\]

\( v_{max} = \text{maximum shortening velocity (positive if the muscle shortens)} \)

\( f_V = \text{dependency of the effective maximum shortening velocity} v_{maxA} \text{ on the active state (see section C.1)} \)

Standard functions in MADYMO can be used for \( f_{il}, f_L, f_p \) and \( f_V \), but it is also possible to use other, user defined, functions in tabular form. The standard functions will now be discussed.

**C.1 Standard Functions**

\( f_H \), the force-velocity relation of CE ('Hill curve') is expressed as:

\[
f_H(v_r) = \begin{cases} 
0 & (v_r \leq -1) \\
\frac{1 + v_r}{1 - \frac{v_r}{CE_{sh}}} & (-1 < v_r \leq 0) \\
\frac{1 + \frac{v_r}{CE_{shl}} \cdot CE_{shl}}{1 + \frac{v_r}{CE_{shl}}} & (v_r > 0)
\end{cases} \tag{7}
\]

Parameters \( CE_{sh} \) and \( CE_{shl} \) determine the shape of the curve, \( CE_{ml} \) represents the ultimate tension during lengthening relative to the isometric force with maximal activation (see
figure C.2). \( v_r \) is the **lengthening** velocity; in literature mostly the **contraction** velocity is used. In other words: the figure is reflected around \( v_r = 0 \).

In the isometrical situation, \( f_H \) is equal to 1. If the muscle contracts \((v_r < 0)\), \( f_H \) decreases quickly; if the muscle lengthens \((v_r > 0)\), \( f_H \) increases quickly, maximal to \( CE_{ml} \). Usually, the force-velocity relation has a positive slope; force increases during lengthening and reduces during shortening.

![Figure C.2: Force-velocity relation of CE](image)

\( f_L \), the force-length relation of CE, is expressed as:

\[
f_L(l_r) = e^{-\left(\frac{b_r}{S_r}\right)^2}
\]

(8)

\( S_r \) determines the width of the active force-length curve (figure C.3). \( f_L \) is highest if the muscle length is equal to the optimum.

The slope of the force-length relation determines a stiffness coefficient. The slope of the active force-length relation is negative for a relative length larger than one. So in this range, the active force-length relation introduces a negative stiffness. This will be partially compensated by the passive force-length relation. However, when the resulting stiffness is negative, feedback may be required in order to stabilise the model.

\( f_P \), the passive force-length relation of PE, is given by:

\[
f_P(l_r) = \begin{cases} 
0 & (l_r \leq 1) \\
(k_f e^{k_2(l_r - 1)} - 1) & (l_r > 1)
\end{cases}
\]

(9)

with constants:
\[ k_1 = \frac{1}{e^{PE_{sh}} - 1} \]
\[ k_2 = \frac{PE_{sh}}{PE_{on}} \]

PE_{sh} determines the shape of the curve, PE_{on} is the relative elongation \((l - l_{ref})/l_{ref}\) inducing a passive muscle force equal to \(F_{max}\) (figure C.3).

The standard function for \(f_v\) is the unity function: the effective maximum shortening velocity is independent of the active state. In the following, \(f_v\) will not be considered further, and so \(v_{max} = v_{max}'\).

In table C.1, as an example, function values are given for default parameter values in MADYMO. The values for \(v_r\) and \(l_r\) are arbitrarily chosen.

<table>
<thead>
<tr>
<th>(v_r)</th>
<th>(l_r)</th>
<th>parameter (CE)</th>
<th>default value</th>
<th>(f_H)</th>
<th>(f_L)</th>
<th>(f_P)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-0.2</td>
<td>-</td>
<td>(CE_{sh})</td>
<td>0.25</td>
<td>0.44</td>
<td>--</td>
<td>--</td>
</tr>
<tr>
<td></td>
<td>+0.2</td>
<td>(CE_{sh})</td>
<td>0.05</td>
<td>1.4</td>
<td>--</td>
<td>--</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(CE_{on})</td>
<td>1.5</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>1.5</td>
<td>(S_{sh})</td>
<td>0.54</td>
<td>--</td>
<td>0.42</td>
<td>--</td>
</tr>
<tr>
<td></td>
<td>1.5</td>
<td>(PE_{sh})</td>
<td>10.0</td>
<td>--</td>
<td>--</td>
<td>0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(PE_{on})</td>
<td>0.8</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Figure C.3:  Force-lenght relations of CE and PE

Table C.1:  Function values for default parameter values
C.2 Active state signal $A$

The active state signal $A$ is a measure of the neural excitation of the muscle. A maximal activation is usually represented by $A = 1$. The active state may depend on time or on the (relative) motion of bodies. Some possibilities are:

'Reference Signal': $A$ is prescribed as a function of time: $0 \leq A \leq 1$, maximal activation is represented by $A = 1$. De Jager [1996] used this signal in a detailed model of the human neck for simulations of frontal and lateral car crashes. Activation of the neck muscles as a reaction to the impact load, starts after a 'reflex time'. The period between that moment and maximal activation is called the 'muscle contraction time'. Due to dynamical processes the active state does not change instantaneously: it takes about 100 ms for the active state to transfer from the rest state to maximal activation, see figure C.4.

![Time signals of neural input $u$, neural excitation $E$ and active state $A$ for $t_r = 50$ ms, $T_e = 30$ ms and $T_{ac} = 10$ ms. Signal $A$ is used in the model to describe muscle contraction.](image.png)

**Figure C.4:** Example of the active state signal $A$ [De Jager, 1996]

'Joint Sensor Signal': $A$ is the value of a degree of freedom (translation or rotation) or the first derivative of that d.o.f. in a 'joint'.

'Body Sensor Signal': $A$ is the component of a motion quantity (position, velocity or angular velocity) of an arbitrary point P1 with respect to:
- the global coordinate system (of the reference body),
- the local coordinate system (of the segment), or
- a second, arbitrary, point P2.