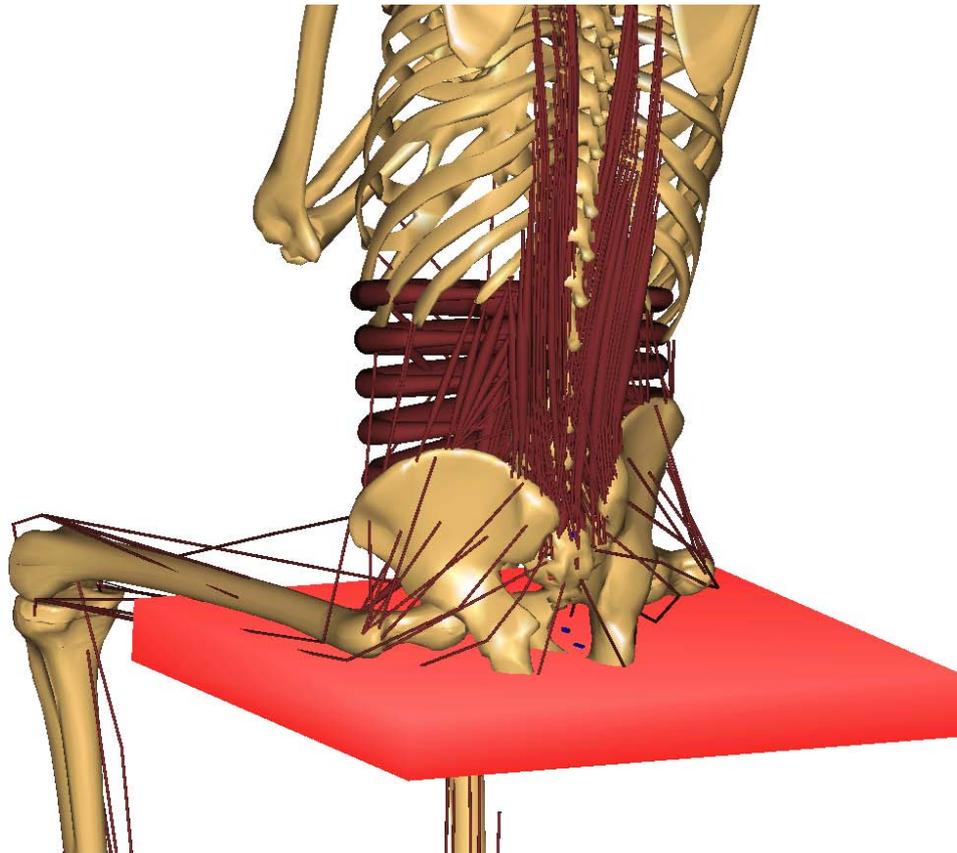


MUSCULOSKELETAL ANALYSIS OF LIGHT POSTURAL MOVEMENTS DURING SITTING

A PILOT STUDY ABOUT THE EFFECT OF LIGHT POSTURAL
MOVEMENTS DURING SITTING ON THE MUSCLE ACTIVITY IN
THE BACK



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Prologue

The report that you are about to read has been established during an internship at the Center for Sensory-Motor Interaction (SMI), Department of Health Science and Technology, Aalborg University, Denmark. I am thankful to Christian Gammelgaard Olesen, Morten Enemark Lund and Michael Andersen Skipper for assisting me setting up the AnyBody model and answering all sort of questions related to the project. I am most grateful to my supervisor Mark de Zee for giving me the opportunity to do an internship at Aalborg University and for giving me valuable advice during the project.

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1 Introduction

Cerebral palsy (CP) is a common cause of disability in childhood [1]. The classical symptoms are spasticity's, spasms, other involuntary movements, unsteady gait and problems with balance. For example scissors walking and toe walking are common among people with CP.

There is no cure for CP, but therapy can help a person to function more effectively and makes life easier. Treatment may include physical therapy, drugs to relax muscle spasms, surgery, hippo therapy, braces, orthotic devices and so on. Furthermore clinical experience with CP-children shows that a combination of light movements and changing sitting-posture reduces spasticity in neck and shoulder muscles and makes it possible for the child to sustain attention for a longer period of time.

A pillow is developed by Aalborg University in Denmark which can generate light movements of the pelvis while sitting on it. It contains four cells that can be inflated and deflated using a pneumatic system. A study is done to investigate how spasticity, upper-limb fine motor skills, and attention are affected by light postural movements in sitting position [2]. In this study it is concluded light movements during sitting position may have a potential application in daily academic settings and may have potential for clinical treatment.

To understand why this light movement of the pelvis during sitting reduces spasticity, it is a prerequisite to understand what the effect of this movement is on the muscles. The goal of this study is to investigate the muscle activity in the back during light movement provided by the moving pillow for different movement patterns and to validate the model used to determine this muscle activity. In this pilot-study results are obtained from six healthy subjects who participated in several experiments.

An already existing model of the seated human built in the AnyBody Modeling System [3] is modified and used to analyze the movement of the body provided by the pillow and is used to estimate the muscle activity.

2 Cerebral Palsy

2.1 Introduction

Cerebral palsy is an umbrella term for a range of non-progressive syndromes of posture and motor impairment that results from an insult to the developing central nervous system. The motor disorders are often accompanied by disturbance of sensation, perception, cognition, communication and behavior. These disturbances are often caused by epilepsy and by secondary musculoskeletal problems. Cerebral palsy is the most common cause of severe physical disability in childhood [1].

2.2 Prevalence

The worldwide prevalence and incidence in children aged from 3-10 years is 2.4 per 1000 children with variability in girls and boys [5,6]. The prevalence is still increasing due advances in neonatal care [7]. The survival of babies with low birth weights has increased, and these babies are more likely to develop cerebral palsy. The economical costs for people with CP in the USA were estimated at \$8.2 billion [8].

2.3 Classification

CP is divided into four major classifications to describe different movement impairments.

Spastic CP

Spastic CP is the most common type, occurring 70% to 80% and it often accompanies one of the other types. Patients with this type of CP are hypertonic. This class is further classified by topography, dependent on the region of the body affected.

- Diplegia: significant leg involvement with little effect on the arms
- Hemiplegia: involvement of the ipsilateral arm and leg
- Quadriplegia: involvement of all four limbs

Ataxic

Ataxic CP is less common occurring in at most 10% of all cases. Individuals can have hypotonia and tremors. They may have problems with skills like writing, typing as well as balance. It is also common for these patients having difficulty with visual and/or auditory processing.

Athetoid/dyskinetic

Patients with athetoid or dyskinetic CP have trouble holding themselves in an upright, steady position (for example sitting, walking), and often show involuntary movements because they have mixed muscle tone. It occurs in 10% to 20%.

Hypotonic

People with hypotonic CP can only move little or cannot move at all, they appear limp.

2.4 Epidemiology

Cerebral palsy is caused by damage to the motor control centre of the developing brain, which can occur in utero, during delivery or during the first two years of life [9]. Examples of factors that increase the chance having a baby with CP are multiple birth,

maternal infection and fetal infection [4]. The severity depends on magnitude, extent and location of the insult in the developing brain and can range from subtle motor impairments to involvement of the whole body. 25% of the patients cannot walk, 30% are mentally impaired, 35% have neurological problems and include seizures and 20-40% of the patients have impairment of visual perception [7,10,11,12]. A typical posture of a child with left hemiparesis is shown in Figure 1.

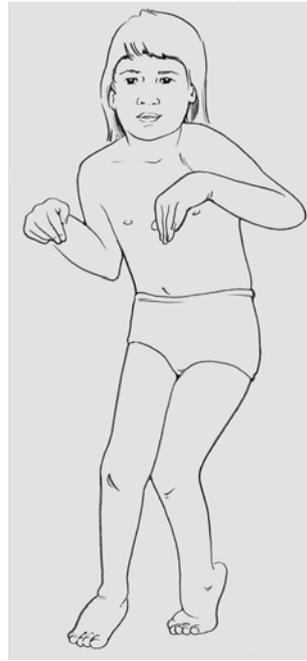


Figure 1: Typical posture of a child with left hemiparesis [3]

2.5 Treatment

There are various forms of treatment that can help a person with cerebral palsy, but there is no cure. Examples of treatments are physical therapy (nearly all patients), drugs to control seizures or alleviate pain or relax muscle spasms, surgery to correct anatomical abnormalities or release tight muscles, hippo therapy and orthotic devices. Also dynamic seating systems, acupuncture and music therapy seemed to help people with CP. Furthermore clinical experience with CP-children shows that a combination of light movements and the possibility to change sitting-posture reduces spasticity in neck and shoulder muscles and makes it possible for the child to sustain attention for a longer period of time.

The availability of new methods of physical therapy, new techniques and new drugs improve function of CP patients and positively affect their quality of life.

3 AnyBody Modeling System

3.1 Introduction

The AnyBody Modeling System, initiated at Aalborg University, is a software system for simulating the mechanics of the live human body working with its environment [3]. It is a general modeling system for making musculoskeletal models. From AnyBody it is possible to obtain results on individual muscle forces, joint forces and moments, metabolism, elastic energy in tendons, antagonistic muscle actions and more making use of inverse dynamics. Furthermore AnyBody can also scale the models to any individual.

3.2 The AnyBody software

The AnyBody software consists out of a Windows graphical user interface and a console application. Although they both have the same modeling facilities, they differ in the ways they can be used. The console application can be called from other programs whereas the graphical user interface contains much more facilities for viewing the model and its results.

Modeling is done by a text-based input by the modeling language named AnyScript. In this way it is easy to develop and maintain the models. It allows users to construct models from scratch, or use and modify already existing models. Besides that it is easy to exchange models and it facilitates cooperation on model development.

3.2.1 *AnyScript language*

AnyScript language is a declarative, object-oriented language for development of multibody dynamics models, particularly for models of the musculoskeletal system [13]. A model in AnyScript is divided into two main sections: the model section containing the definition of the mechanical system, that is the body and its environment and the study section containing operations that can be performed on the model such as a kinematical analysis and kinetic analysis. Also the visualization of the model in the viewer of the graphical user interface is programmed by the user in AnyScript.

3.2.2 *Structuring models*

Structuring becomes very important when models getting large, for example like a full body model. In AnyScript it is possible to structure models in a tree structure similar to a file system. Large models are organized in several files which can be simply combined by C-style including statements. In this way it is possible building a model by several modelers and model parts can easily be exchanged.

However, models can be structured in many different ways, so model merging can become a problem. To make merging possible the model is split into two distinct parts: the body model and the application model. The body model contains anatomical data like segments, joints and muscles but no boundary conditions. These are in the application model which contains data about movements, loads and external objects like tools or a chair. In this way users can exchange body parts and connect them with any application they want. The interface between two body models or between body and environment can then be handled by addition of the necessary elements outside the basic objects declaration.

3.3 Inverse dynamics

Musculoskeletal models can be divided in three groups: forward dynamics, inverse dynamics and a combination of these two called the Kane's –method (TMT-method). Forward dynamics computes the motion based on (predicted) forces and moments. Although it is attractive in detailed modeling of various physical phenomena, it is computationally demanding and requires time consuming optimization to make the model perform a specific task. In inverse dynamics, the motion is known and the forces and moments are computed. This puts many restrictions on the model but it has also a number of attractive features such as computational speed and stability. AnyBody makes use of inverse dynamics, therefore this computational method will be described in more detail. Inverse dynamics makes use of the Newton-Euler equations stated as:

$$F = m \cdot \ddot{r} \quad (1)$$

When this equation is applied on a simple segment as shown in Figure 2 the equations are:

$$m \cdot \ddot{x} = F_x \quad (2)$$

$$m \cdot \ddot{y} = F_y - m \cdot g \quad (3)$$

$$J \cdot \ddot{\theta} = -r \sin \theta \cdot F_y - r \cos \theta \cdot F_x \quad (4)$$

where the segment is hinged at the top and J is the mass inertia. AnyBody uses a generalized concept of this example.

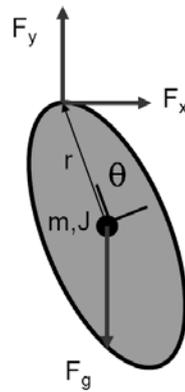


Figure 2: Simplified body segment

In biomechanics all segments are modeled as rigid bodies. Every segment has 6 degrees of freedom, therefore the force vector consists of 3 forces and 3 moments. When this force vector is subdivided in forces and moments of the muscles ($g_i^{(M)}$), reaction forces ($g_i^{(R)}$) and known applied forces ($g_i^{(app)}$), the Newton-Euler equation for each segment yields [13]:

$$g_i^{(M)} + g_i^{(R)} = g_i^{(app)} - \begin{bmatrix} m_i I & 0 \\ 0 & J'_i \end{bmatrix} \dot{v}_i - \begin{bmatrix} 0 \\ \tilde{\omega}_i J'_i \omega'_i \end{bmatrix} \quad (5)$$

However, the inverse dynamics in a musculoskeletal model is subject to a redundancy

problem, there are not enough equations available to determine all the muscle forces. The solution for this problem will be shown in the next paragraph.

3.4 Muscle recruitment

A fundamental problem of inverse dynamics is the indeterminacy of individual muscle forces. In the body there are more muscles available than necessary to perform a certain movement; in fact the body is statically indeterminate. There are more muscle recruitment patterns possible to drive a given motion. The central nervous system (CNS) resolves this redundancy problem; it chooses a set of muscle actions that realize the desired motion. It is assumed the CNS tries to minimize the load on the muscles and therefore the unknown muscle and joint forces can be found as a solution to an optimization problem. Different optimization criteria can be used to solve this problem like the polynomial criteria, soft saturation criteria and min/max criteria [15].

A mathematical interpretation can be obtained by setting up the total system of equilibrium equations:

$$\begin{bmatrix} \mathbf{C}^{(M)} \\ \mathbf{C}^{(R)} \end{bmatrix} \cdot \begin{bmatrix} \mathbf{f}^{(M)T} \\ \mathbf{f}^{(R)} \end{bmatrix} = \mathbf{r} \quad (6)$$

where \mathbf{C} is a matrix of coefficients depending on the current position of the body segments, \mathbf{f} is a vector of unknown forces and \mathbf{r} is a right hand side consisting of external forces and inertia forces. Superscripts M and R designate terms related to muscles and joints reactions respectively. Now it is assumed the vector \mathbf{f} can be obtained by a solution on the following optimization problem:

$$\text{Minimize} \quad G(\mathbf{f}^{(M)}) \quad (7)$$

$$\text{subject to} \quad \mathbf{C}\mathbf{f} = \mathbf{r} \quad (8)$$

$$\text{and subject to} \quad f_i \geq 0, \quad i \in \{1, \dots, n^{(M)}\} \quad (9)$$

where $G(\mathbf{f}^{(M)})$ is some scalar measure of the body load, $n^{(M)}$ is the number of muscles in the system and equation (9) gives the constraint that express that muscles can only pull, never push. The objective function G is minimized with respect to all unknown forces in the problem. It is proposed that muscle force should be predicted in such a way that the sum of muscle action is minimized and therefore activity endurance is maximized. Linear muscle recruitment leads to no muscle synergy at all, while quadratic muscle recruitment results in some muscle synergy. In general, the higher the order of the polynomial of the objective function, the more synergy there is. In this report the objective function is chosen to be a polynomial of the power 3 because it appeared to represent the consensus of values reported in the literature [14]. The objective function becomes:

$$G(\mathbf{f}^{(M)}) = \sum_{i=1}^{n^m} \left(\frac{f_i^{(M)}}{N_i} \right)^3, \quad i = 1, \dots, n^{(M)} \quad (10)$$

where N_i are the weight factors, in this case the muscle strength. Other criteria like the soft saturation criterion and the min/max criterion [15] might me more appropriate,

although none of them has been generally proven to be superior from others from a physiological point of view.

3.5 Muscle modeling

AnyBody contains three types of different muscle models. The simplest one assumes constant strength of the muscle regardless of its working conditions. The more complicated ones take conditions as length, contraction velocity, fiber length, pennation angle, tendon elasticity and stiffness into account. Although it seems the last ones are better than the simplest one, this is not always true. The more complicated muscle models require several physiological parameters that may be difficult to obtain or can be difficult to estimate. Besides that, for models with moderate contraction velocities and small joint angle variations even the simplest model will work reasonable well.

3.6 Scaling

Musculoskeletal models must be scalable to sizes of different individuals. Besides scaling the overall geometry, also properties like muscle insertion points, muscle parameters and wrapping surfaces have to be scaled. There are seven different scaling laws available in AnyBody which can be divided into two groups: scaling with and without external measurements. The simplest scaling law scales equally in all directions where the input is joint to joint distances. More extended scaling laws take also mass and fat percentage into account.

3.7 Motion capture data

In biomechanics it is often wanted to make models move as measured in the laboratory. The recording of movements is mostly done by measurement techniques that track optical markers on the body in space by using synchronized cameras. Most of the systems are capable saving data in a standard format called C3D file. It contains data of the spatial trajectory of optical markers fixed to the object that is recorded. The C3D file can also contain data from force platform measurements or EMG measurements. AnyBody can read these files directly.

4 The model

4.1 Introduction

As stated in the previous section, in AnyBody it is possible to use and to adapt already existing models. The musculoskeletal model used in this study is based on a modified version of the “Seated Human” model from the public domain AnyScript Model Repository [16]. The boundary conditions of the model used in this study are different from the ones in the “Seated Human” model; therefore it is adjusted in several ways. This section will give a detailed description of the adjusted model (Figure 3).

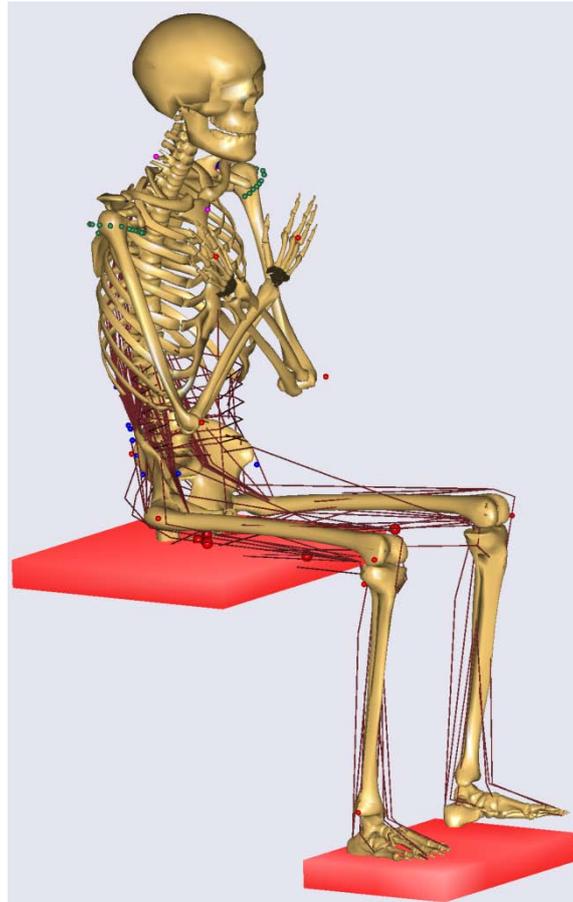


Figure 3: Model view

4.2 Body model

4.2.1 *Body segments and joints*

The body model is built up out of several sub models. It comprises an arm and shoulder complex, with morphology according to Van der Helm [17]. It is assumed the arms are in a fixed position during the experiment, therefore it is chosen to use the arm and shoulder complex without muscles and drive them by external motors. Furthermore it contains a spine model comprising sacrum, all lumbar vertebrae, a rigid thoracic section, and a total of 158 muscles according to de Zee [18], and a pelvis and lower extremity model with a total of 70 muscles. The segments are modeled as rigid bodies which include the mass of the bone and the soft tissue that can reasonably be attributed to the different bones. The joints used in the model are idealized frictionless constraints

between the segments and are modeled as standard joints like for example a hinge joint for the knee and spherical joints for the hip.

4.2.2 Muscle model

The pillow provides relatively small movements to the body, therefore moderate contraction velocities and small joint angle variations of the muscles are assumed. Keeping in mind this it is justified to use a muscle model that contains isometric muscles. The input to the muscle model is only the muscle's presumed isometric strength, the force that the muscle can exert in a static condition at its optimum length. The muscles are modeled as strings spanning the distance between origin and insertion via points and wrapping surfaces.

4.2.3 Scaling

The body model is scaled by a scaling method based on the body height, the body mass and the fat percentage of a person. The scaling method scales the sizes of segments according to their masses in two directions; the third direction is directly controlled by segment length variables. Furthermore it scales the muscle strength with the mass; the heavier the person, the higher the muscle strength. The fat percentage is used to add mass to each segment without increasing the muscle strength. It does not modify the total body mass or the size of the body; it is only used to lower the strength of the muscles. Input parameters of this scaling method are the body mass, body height and segment length variables as defined in Figure 4.

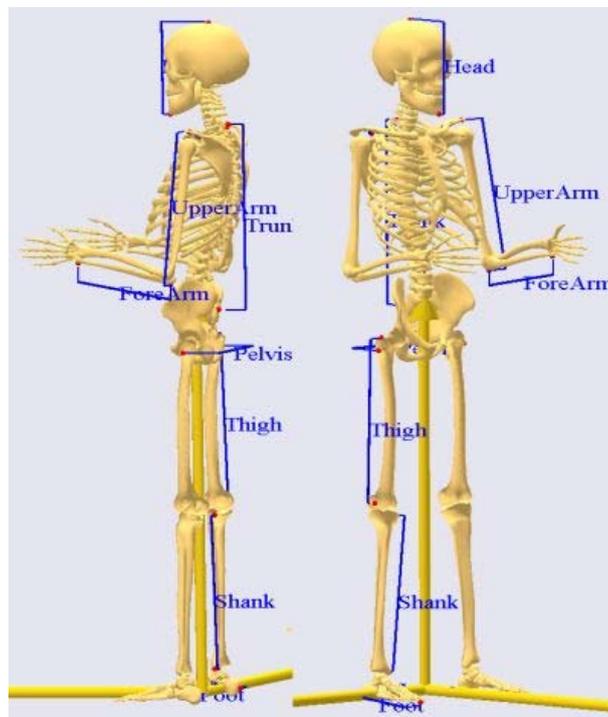


Figure 4: Length variables as input parameters of the scaling method

The fat percentage is controlled by the body height and mass and is calculated by the Body-Mass Index according to Frankenfield [19]. A different relation is used for man and women.

4.3 Application model

4.3.1 Kinematics and kinematic constraints

The kinematics of the body is controlled in three different ways: by predefined joint-angles, motion captured data and by a spine rhythm.

Predefined angles

Some degrees of freedom (DoF) in the body are assumed to be constant during the experiment like for example the elbow pronation or the rotations in the ankle joint. Although they might change little, their contribution to the analysis is assumed to be not significant and therefore they are set by predefined angles.

Spine rhythm

The kinematics of the spine is controlled by a spinal rhythm which sets the three angles between each vertebrae based on the presumption that the passive elastic elements of the spine causes it to behave kinematically as an elastic beam. The input variables are the three angles between the pelvis and the thorax.

Motion captured data

The remaining degrees of freedom are driven by motion captured data (Figure 5). During the experiments the motion of the body is recorded by several cameras. These cameras record the motion of several markers which are placed on predefined locations on the body. This data is imported into AnyBody and the motion of the different markers is

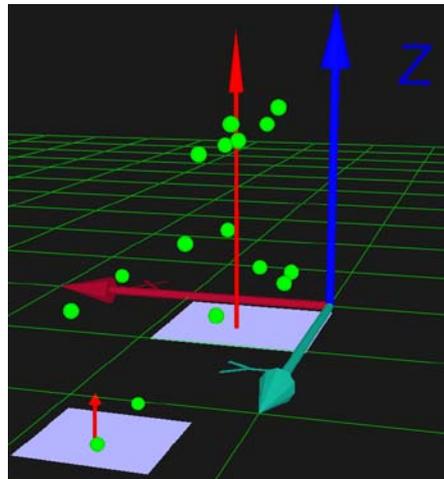


Figure 5: Motion captured data

coupled to landmarks on several segments of the body. In this way the landmarks on these segments follow the trajectory described by the motion captured data. An overview of the DoF driven by marker data and constraint by predefined angles is shown in Table 1. The model is driven by 12 markers, namely on the spine (level C7), sternum (between the two clavicles), elbows, hands, spina iliaca posterior superior (SIPS), knees and ankles (Figure 6). The precise placement of the markers will be shown in the section "Method". Each marker drives one landmark, defined on a particular segment in the model. One marker constraints the position of a landmark in three directions. This would give 36 constraints. As can be seen in Table 1 the markers only drive 14 DoF's; the

Table 1: Joints with their DoF and constraints

| Joint | Degree of freedom | Constraint |
|---|-------------------|------------------|
| Neck | Extension | Predefined angle |
| Rotation thorax with respect to the pelvis | Lateral bending | Markers |
| | Rotation | Markers |
| | Extension | Markers |
| Clavicular joint | Protraction | Markers |
| | Elevation | Markers |
| | Axial Rotation | Markers |
| Glenohumeral joint | Abduction | Markers |
| | Flexion | Markers |
| | External Rotation | Markers |
| Elbow | Flexion | Markers |
| | Pronation | Predefined angle |
| Wrist | Flexion | Predefined angle |
| | Abduction | Predefined angle |
| Hip | Flexion | Markers |
| | External Rotation | Markers |
| | Abduction | Markers |
| Knee | Flexion | Markers |
| Ankle | Plantar Flexion | Predefined angle |
| | Eversion | Predefined angle |

model would be kinematically over-constraint. To cope with this AnyBody uses a “soft” kinematic solver, a solver that accepts more constraints than the system has DoF. The data of the markers become so called “soft” constraints as for example joints stay “hard” constraints. Although the landmarks on the body do not follow the markers completely in all three directions, there are several reasons to do it like this. In a complicated model with many markers it can be very time-consuming to disentwine which Dof is driven by which marker and to select a subset of directions to drive. Besides that the marker data contains errors like tissue artifacts, also called skin artifacts; the markers never move exactly like the landmarks on the bone do. Also the placement of the markers can infest the model with error; it is impossible to place the marker at exact the same location as it is defined in the model. It is therefore natural in the model to presume that the connection between the marker and the bone is not rigid. Factors that influence the reliability of the tracking data are:

- the amount of soft tissue between the bone and the marker
- how close the marker is to a joint (more close to a joint means more movement of soft tissue means less reliable)
- how easy it is to palpate the landmarks on the bone

It is chosen to use the minimum number of markers possible to constraint the degrees of freedom in the model. The theory behind this choice is to place the markers in such a way the tracking data becomes most reliable. To add more markers than needed on places which give less reliable tracking data will infest the model with more error.

As stated before the 12 markers give more constraints than strictly needed, some DoF are double constrained. If it is known one of these constraints is more reliable than the other, weighting factors could be added to minimize error in tracking data.

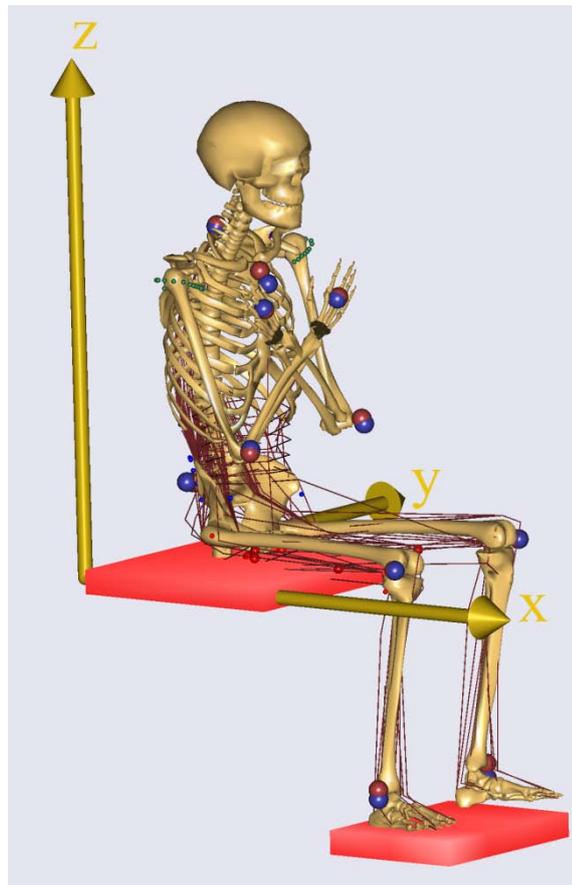


Figure 6: Marker setup

In the model most of the markers are weighted equally, but there are some exceptions. The sternum marker is assumed to be less reliable compared to the marker on the vertebrae. It is more difficult to palpate the exact position of the landmark on the sternum. Therefore the position of the thorax with respect to the pelvis is mainly determined by the marker on the vertebrae. The sternum-marker is only used to constraint the rotation of the thorax around the z-axis (Figure 6). Note that the position of the landmark in y-direction is the middle of the body and therefore can be determined accurately. Furthermore the markers on the elbows and the hands are only used to position the arms in the right position; therefore they are weighted 0.1 (on a scale of 1) in all directions. In this case they do not infest the position of the thorax with error.

4.3.2 Environment

The environment is based on the chair model according to Rasmussen [20]. This chair model is adapted to a model which exists out of a footrest and a seat as depicted in Figure 6. Both footrest and seat are fixed to the “real world”.

Real seated persons are supported by surfaces and pressure distribution. In rigid multibody models support is modeled by supporting points and supporting forces. The

model has been equipped with points which can transfer reaction forces to the supporting seat and footrest. Eight points are defined on the pelvis, two on both of the thighs and two on each foot. Several points are used to emulate the distribution of pressure over the seat and footrest.

The contact elements can only provide compressive reactions. They implement Coulomb friction where the possible friction force is proportional to the reaction force in the element:

$$|F_f| = \mu \cdot R \quad (11)$$

where F_f is the friction force, μ is the friction coefficient and R the reaction force. The direction of R is perpendicular to the supporting surface and F_f can be in any direction perpendicular to R . However the size of these forces is unknown and cannot be determined by equilibrium because it is statically indeterminate. This problem is solved by adding these normal reaction forces to the redundant muscle recruitment problem and letting the normalization vectors N_i corresponding to R_i be large compared to the normalization factors for the anatomical muscles [32]. In this way the modeling approach presumes that the human body uses the available support points on the seat and footrest in such an extent that it minimizes the necessary muscle activity to retain the posture. Mention that the optimization algorithm minimizes the muscle activity.

It is assumed there is always contact between the body and the seat, or actually between the body and the pillow. Therefore no matter what the position of the pelvis is; there will be compressive reactions and friction forces in the predefined points in such an extent that it minimizes the muscle activity.

5 Methods

5.1 Subjects

Six healthy subjects participated in the experiments, five of them are male and one is female. They had an average age of 28 years (22-39 yrs), an average length of 182 cm (157-197 cm) and an average weight of 72 kg (54.1-81.0 kg). The physical characteristics of each subject are shown in Table 2. The segment length variables as defined in Figure 4 are listed in Table 3.

Table 2: Physical characteristics for the different subjects

| Subject | 1 | 2 | 3 | 4 | 5 | 6 |
|-------------------------|------|------|--------|------|------|------|
| Gender | Male | Male | Female | Male | Male | Male |
| Age | 39 | 22 | 26 | 29 | 28 | 26 |
| Body Weight (kg) | 70.0 | 77.6 | 54.0 | 80.5 | 69.2 | 75.0 |
| Body Height (cm) | 186 | 197 | 154 | 181 | 190 | 185 |

Table 3: Segment length variables

| Subject | 1 | 2 | 3 | 4 | 5 | 6 |
|-----------------------|------|------|------|------|------|------|
| Head (cm) | 14 | 14 | 12 | 14 | 13.5 | 13 |
| Upper arm (cm) | 34.0 | 36.5 | 25.5 | 32.5 | 32.0 | 35.0 |
| Lower arm (cm) | 27.0 | 29.5 | 22.0 | 26.5 | 28.0 | 26.0 |
| Trunk (cm) | 66.0 | 69.0 | 52.0 | 65.0 | 70.0 | 69.0 |
| Pelvis (cm) | 18.5 | 26.0 | 24.0 | 26.5 | 24.5 | 26.0 |
| Thigh (cm) | 42.5 | 49.0 | 37.0 | 46.0 | 44.5 | 41.5 |
| Shank (cm) | 39.5 | 43.0 | 35.0 | 44.5 | 40.0 | 41.5 |
| Foot (cm) | 21.0 | 24.0 | 18.0 | 21.0 | 21.5 | 21.0 |

5.2 Setup

The experiments were performed in the gait lab of Aalborg University (Figure 7).

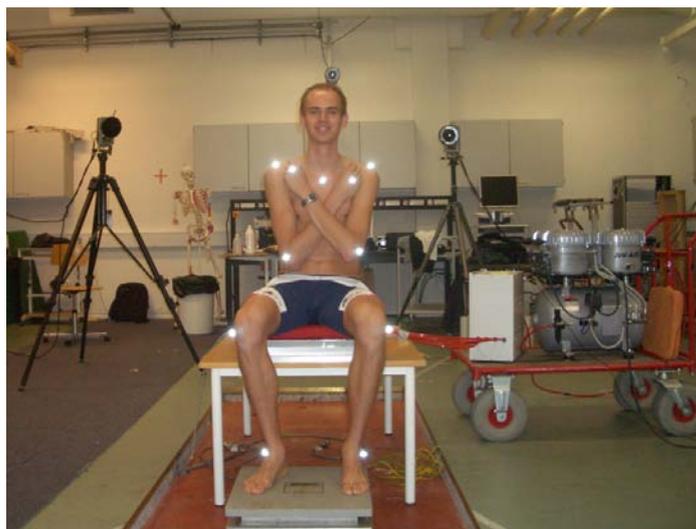


Figure 7: Setup of the experiment

A rigid table was placed in the middle of the lab with a force platform (AMTI) on top of it. The pillow was taped onto this to be sure it was not possible to move the pillow with respect to the force platform. Another force platform was placed on the floor in front of the table. The height of the pillow with respect to the top of the force platform on the ground was chosen to be equal to the height of a normal chair.

The pillow used in the experiments is designed by Aalborg University (Figure 8). It contains four cells that can be inflated and deflated using a pneumatic system (Jun Air). Each cell produces a vertical lift of approximately 5 cm. Different pillow setups are used during the experiments; movement pattern, pulsation frequency and inflation speed are varied.

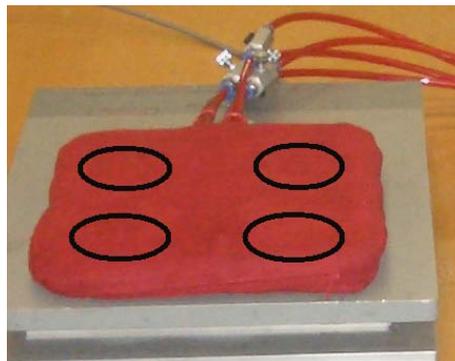


Figure 8: Pillow with the location of the four cells

Marker setup

Sixteen passive markers with a diameter of 14 cm were used to track the movement of the body. Two were placed on the malleolus lateralis, two on the caput fibulae, two on the trochanter major, two on the spina iliaca posterior superior, two on the epicondylus lateralis, two on the caput metacarpale of the middle fingers, two on the acromion, one on the sternum between the two clavica and one on the processus spinosus of one of the vertebrae (level C7). Most of the markers can be seen in Figure 7 and Figure 9. Eight cameras (Oqus, Qualisys) were placed around the subject to record the movement of the markers.

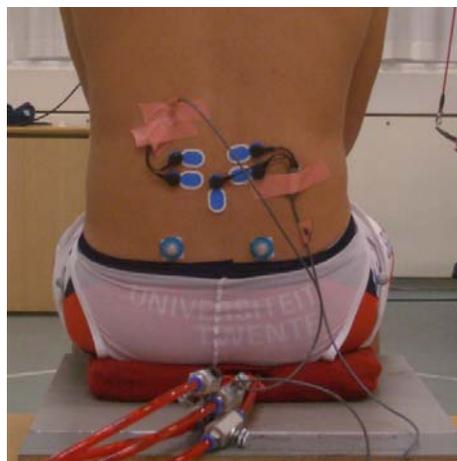


Figure 9: EMG electrodes on the erector spinae

EMG electrodes setup

EMG of the lower superficial part of the erector spinae is recorded with a wireless EMG system (Noraxon). Two electrode pairs are placed on the erector spinae at the height where the lordose of the lower spine is maximal (Figure 9).

5.3 Task

The task for all experiments was the same; the subject was asked to sit on the pillow with both feet on the force platform on the ground. The position of the bottom should be in such a way that the sit bones are above the two cells in the back of the pillow. The arms should be in a crossed position against the chest. The task of the subject was to sit as relaxed as possible on the pillow and remain balance of the upper body. He/she was asked to look at a fixed point in front of him/her.

5.4 Experiments

5.4.1 Validation experiment

The validation experiment is done with the pillow setup as shown in Table 4.

Table 4: Pillow setup for the validation experiment

| Properties | Setup |
|-----------------------------------|--------------|
| <i>Pulsation frequency</i> | 0.67 Hz |
| <i>Movement pattern</i> | Crossed |
| <i>Inflation speed</i> | Maximal |

Pulsation frequency is the frequency at which the next cell or cells are inflated. At the same time the inflated cell(s) is/are deflated. If the movement pattern is circular, the different cells will be inflated consecutively in a counterclockwise direction. If the movement pattern is crossed, the right-front and the left-back cell are inflated and deflated at the same time and vice versa. Inflation speed is the speed of the air that inflates the cell.

The duration of each measurement was 30 seconds and the subjects were given one minute to acclimatize to the different pillow setups.

5.4.2 Experiments with different setups for the pillow

To investigate the effect of different pillow setups on the muscle activity of several muscles in the body, 10 more experiments were done. The pillow setup used in these experiments is shown in Table 5. Unfortunately the results of these experiments are not reported in this study. It was not possible to analyze, report and discuss these results in the given timeframe.

5.5 Data processing

The data which has to be processed are the measurements by the force platforms, the EMG measurements and the kinematic measurements. All data is processed by the software package Qualisys Track Manager (Qualisys AB, Gothenburg, Sweden). Kinematic measurements are sampled with a sample frequency of 100 Hz, kinetic and EMG measurements with a sample frequency of 2000 Hz.

Table 5: Pillow setup for the experiments done to investigate muscle activity

| Nr | Pulsation frequency (Hz) | Movement pattern | Inflation speed front valves | Inflation speed back valves |
|----|--------------------------|------------------|------------------------------|-----------------------------|
| 1 | 0.5 | Crossed | Maximal | Maximal |
| 2 | 2 | Crossed | Maximal | Maximal |
| 3 | 0.5 | Circle | Maximal | Maximal |
| 4 | 0.67 | Circle | Maximal | Maximal |
| 5 | 2 | Circle | Maximal | Maximal |
| 6 | 0.5 | Crossed | Sub-maximal | Sub-maximal |
| 7 | 0.67 | Crossed | Sub-maximal | Sub-maximal |
| 8 | 2 | Crossed | Sub-maximal | Sub-maximal |
| 9 | 0.67 | Crossed | Zero | Sub-maximal |
| 10 | 0.67 | Crossed | Maximal | Sub-maximal |

5.5.1 Filter settings

The EMG signal is hardware filtered with low pass filter with a cut off frequency of 1000 Hz. In Matlab (The MathWorks Inc., Natick, USA) the signal is high pass filtered with a cut off frequency of 10 Hz with a fourth order Butterworth filter. In order to get an envelope to describe muscle activity, the raw EMG data is full wave rectified and subsequently smoothed by a low-pass filter with a cut-off frequency of 4 Hz with a fourth order Butterworth filter using MatLab.

Kinematic data is low-pass filtered with a cut-off frequency of 5 Hz with a fourth order Butterworth filter using AnyBody and the data from the force platform by the same filter settings using MatLab.

5.5.2 Validation

Validation is done by two methods, both validate the model but not in the same extent. The ultimate goal is to validate the muscle activity of the erector spinae estimated by AnyBody. This validation is done by comparing the measured EMG signal of the erector spinae with the muscle activity estimated by AnyBody. One could ask himself why other validation methods are needed then. Two reasons are:

- A validation based on EMG-signals is not too accurate as will be explained in the section "Discussion" and therefore will not be sufficient
- A negative result of validation by EMG-signals can have several causes; the model does not have to be completely wrong. It can be desirable to validate at least a part of the model; therefore other validation methods are used

Therefore a validation is done by measurements of the two force platforms. The measured reactions by the force platforms are compared with the reactions estimated by AnyBody. This validation says something about if the accelerations of the body are correct, if the body is placed in the right position with respect to the seat and footrest and if the model predicts the reactions in the right way.

Validation by the force platforms

The reaction forces and moments are calculated by AnyBody for the seat and the footrest by applying Newton's law on the estimated reactions in the contact elements.

These are compared with data from the two force platforms. The validation is quantified in two ways, first by calculating the correlation coefficient with its corresponding confidence interval. The correlation coefficient shows how close the relationship between two variables is. It is based on the sum of the products about the mean of the two variables. The confidence interval tells something about how close the data is to this correlation coefficient. A 95%-confidence interval is used. Furthermore the validation is quantified by calculating the mean absolute error (MAE):

$$MAE = \frac{1}{n} \sum |R_{force\ platform} - R_{estimated}| \quad (12)$$

where n is the number of time steps, $R_{force\ platform}$ the measured reactions of the force platforms and $R_{estimated}$ the estimated reactions by AnyBody. The correlation coefficient can be high, but it does not say anything about an offset between data, therefore the MAE is used.

The validation is done for the reaction forces of the seat and footrest together in all three directions (total reaction force), for the reaction forces in the seat itself in all three directions and for the reaction moments of the seat around both horizontal axes. The first 12 seconds of the experiment are used for validation. This means at least 4 cycles are included in the validation.

Validation by EMG

The erector spinae in the AnyBody model is divided into a total of 29 fascicles on each side of the spine according to the descriptions by the group of Bogduk [21]. Surface EMG measures only myoelectric signals of the superficial muscles, therefore only the fascicles of the iliocostalis lumborum pars thoracis (ILpT) are used in the validation. These are the most superficial fascicles in the lumbar part of the erector spinae. The validation is quantified by calculating the correlation coefficient and its corresponding confidence interval between the EMG envelope and the average estimated muscle activity of the fascicles of the ILpT. It is assumed the fascicles of the ILpT together provide a certain force and they are active at the same time. Therefore the average muscle activity of these fascicles should give the best representation of activity of this muscle group. The muscle activity in the AnyBody model is defined as the actual force divided by the maximal force this particular muscle could provide.

The mean absolute error (MAE) is not used to quantify this validation method because nothing can be said about the difference in amplitude. Also for this validation method the first 12 seconds of the experiment are used.

6 Results

The results of the experiments described in the previous paragraph are shown below. One must notice that the forces and moments are not measured and reported in the global reference frame, but in the reference frame of the force platforms (Figure 10).

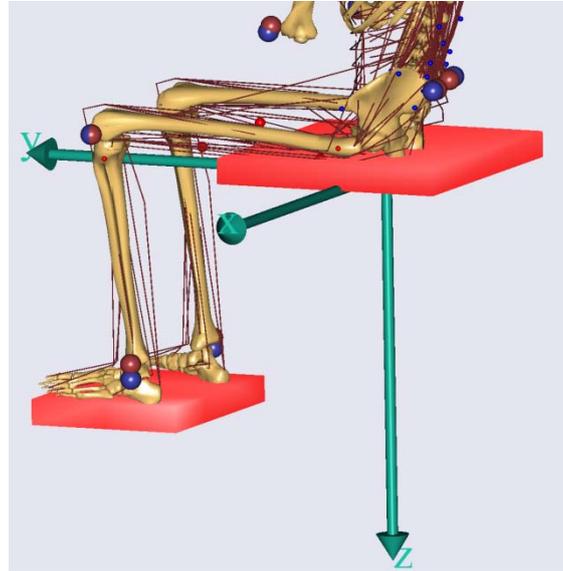


Figure 10: Reference frame of the force platforms

6.1 Validation by force platforms

The results of validation of the total reaction forces in all three directions for subject 2 are shown in Figure 11. The estimated reactions by AnyBody follow the measured reactions well, the mean correlation coefficients (MCC) for Fx, Fy and Fz for the six

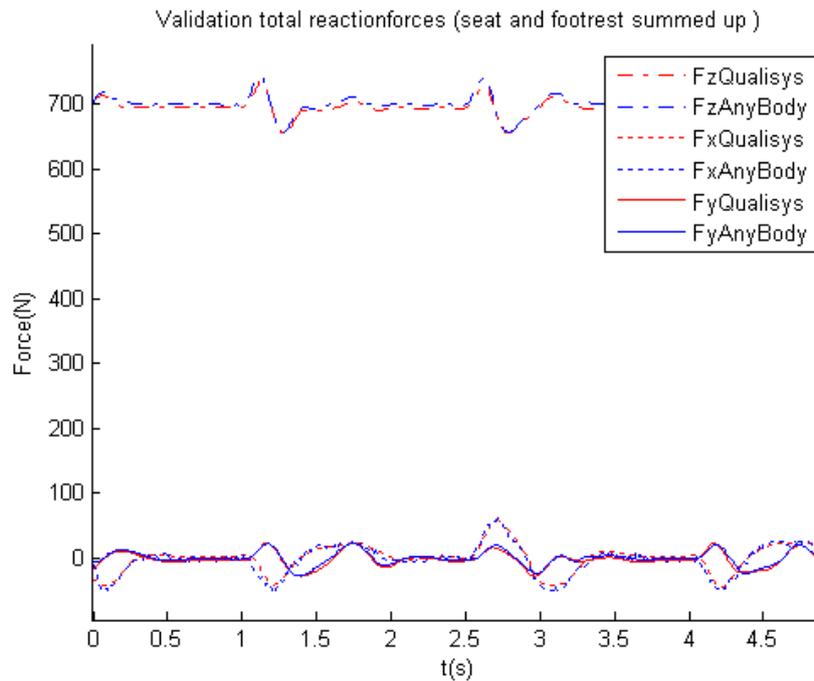


Figure 11: Validation total reaction forces for subject 2

subjects are 0.98, 0.91 and 0.95 respectively with a small mean confidence interval (CI) (Table 6).

Table 6: The mean CC of all six subjects with corresponding mean CI

| | Fx total | Fy total | Fz total | Fx seat | Fy seat | Fz seat | Mx seat | My seat |
|---------------------|-----------------|-----------------|-----------------|----------------|----------------|----------------|----------------|----------------|
| Mean CC | 0.98 | 0.91 | 0.95 | 0.86 | 0.78 | 0.61 | 0.46 | 0.96 |
| Mean CI-low | 0.97 | 0.91 | 0.95 | 0.85 | 0.76 | 0.58 | 0.42 | 0.96 |
| Mean CI-high | 0.98 | 0.92 | 0.96 | 0.88 | 0.80 | 0.65 | 0.51 | 0.97 |

The mean absolute errors (MAE) for all six subjects are listed in Table 7. The mean value of this MAE for all six subjects is 3.6 N, 2.6 N and 9.0 N for Fx, Fy and Fz respectively.

Table 7: Mean absolute errors (MAE) for all six subjects

| Subject | Fx total | Fy total | Fz total | Fx seat | Fy seat | Fz seat | Mx seat | My seat |
|----------------|-----------------|-----------------|-----------------|----------------|----------------|----------------|----------------|----------------|
| 1 | 4.0 | 2.7 | 5.8 | 7.5 | 7.7 | 34.5 | 5.8 | 7.0 |
| 2 | 5.4 | 2.3 | 8.2 | 4.0 | 6.4 | 81.2 | 7.8 | 8.2 |
| 3 | 1.5 | 2.7 | 9.5 | 4.6 | 12.3 | 14.1 | 19.1 | 4.7 |
| 4 | 4.2 | 2.5 | 9.0 | 13.5 | 8.9 | 20.2 | 10.0 | 5.5 |
| 5 | 3.2 | 2.4 | 10.0 | 9.6 | 6.7 | 58.6 | 8.9 | 5.0 |
| 6 | 3.1 | 2.8 | 11.3 | 7.5 | 5.9 | 32.9 | 5.0 | 6.2 |
| Mean | 3.6 | 2.6 | 9.0 | 7.8 | 8.0 | 40.2 | 9.4 | 6.1 |

The validation of the reaction forces of the seat for subject 2 is shown in Figure 12. An offset of the force can be seen for the estimated reactions, especially for the normal force. The mean MAE also shows this: 7.8 N, 8.0 N and 40.2 N for Fx, Fy and Fz respectively. The mean CC are 0.86, 0.78 and 0.61 with again a small mean CI (Table 6).

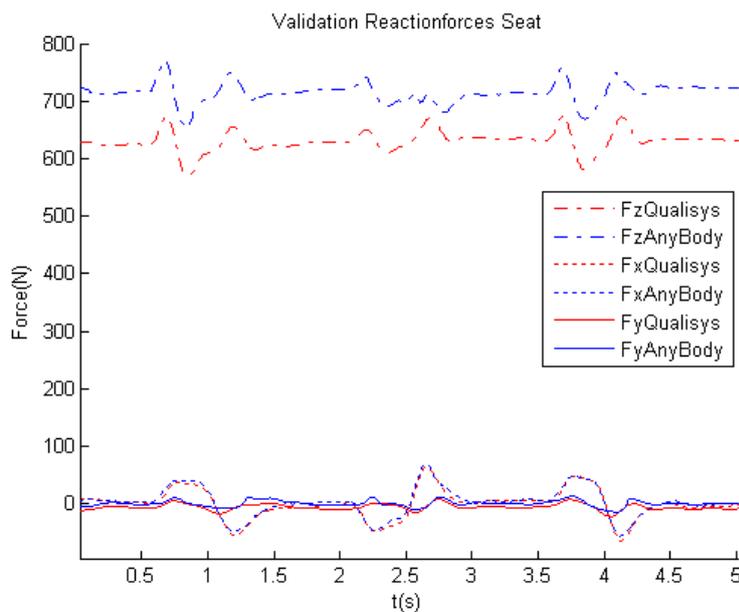


Figure 12: Validation of the reaction forces of the seat for subject 2

The validation of the reaction moments of the seat for subject 5 is shown in Figure 13. The mean CC are 0.46 for Mx and 0.96 for My. The MAE's are 9.1 Nm and 6.1 Nm.

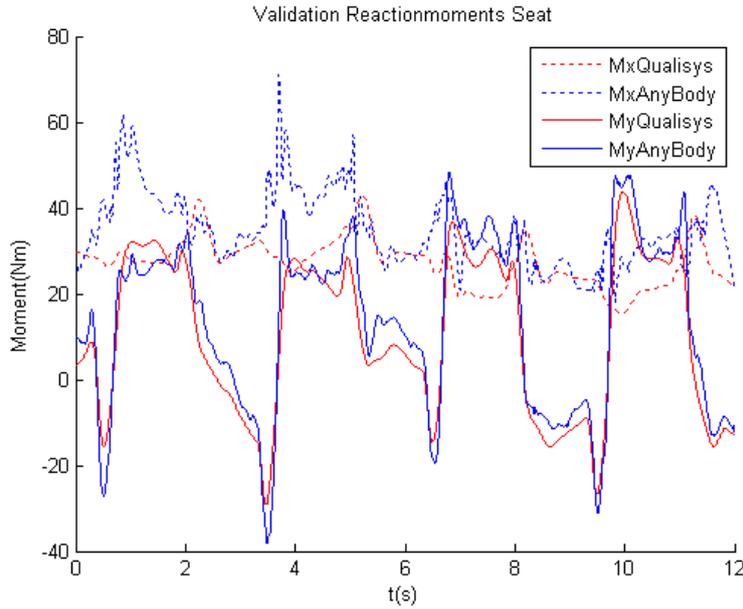


Figure 13: Validation reaction moments Mx and My of the seat for subject 5

6.2 Validation by EMG

The CC and the CI for the left and right erector spinae for all six subjects are listed in Table 8. It can be seen their range is wide, from a CC of 0.70 to even negative CC. For some there is not a correlation at all. The mean CC for left and right erector spinae show on average there is no correlation. The mean CI is small for all subjects.

Table 8: CC and mean CC of the erector spinae with corresponding CI and MCI

| Subject | Erector Spinae Right | | | Erector Spinae Left | | |
|-------------|----------------------|--------|---------|---------------------|--------|---------|
| | CC | CI_low | CI_high | CC | CI_low | CI_high |
| 1 | -0.58 | -0.62 | -0.54 | 0.57 | 0.54 | 0.61 |
| 2 | 0.70 | 0.67 | 0.76 | 0.01 | -0.05 | 0.07 |
| 3 | 0.28 | 0.23 | 0.33 | -0.07 | -0.13 | -0.01 |
| 4 | -0.10 | -0.12 | 0.00 | -0.58 | -0.62 | -0.54 |
| 5 | 0.55 | 0.51 | 0.59 | 0.20 | 0.14 | 0.25 |
| 6 | -0.31 | -0.36 | -0.25 | 0.02 | -0.04 | 0.08 |
| Mean | 0.09 | 0.05 | 0.15 | 0.03 | -0.03 | 0.08 |

A closer look to the EMG results and the estimated muscle forces gives some more relevant information. In general the maximum measured EMG signal is low, except for subject 5 they do not exceed $80 \mu V$. For subject 3 and 6 there is no EMG measured at all. If these two subjects are kept out of the results, the CC for left and right erector spinae is still low: 0.10 and 0.03 respectively.

In the EMG measurements it can be clearly seen that noise infests the EMG signal. The peaks in the graphs of subject 3 and 6 have a constant frequency around 1 Hz. Most

likely these are ECG artifacts. Finally one must notice that the maximal estimated muscle activity is low, on average around 10 percent of the maximal muscle activity.

To be sure the correlation coefficient does not give a wrong interpretation of the results, it is chosen to go deeper into the results of one of the subjects. The EMG signals and muscle activity for both right and left erector spinae are shown in Figure 14 and Figure 15.

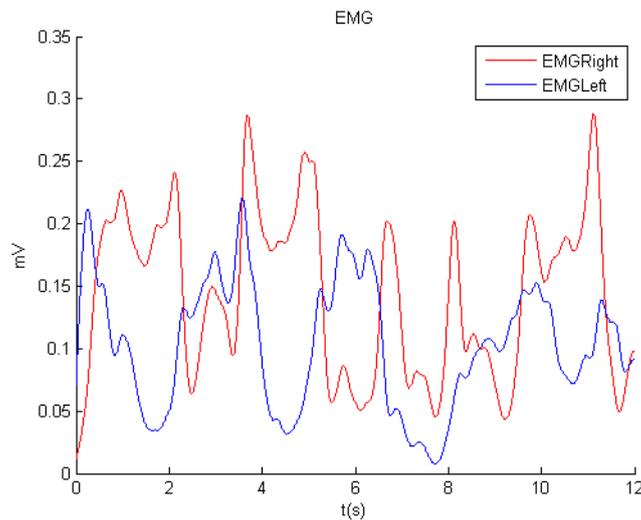


Figure 14: EMG signals subject 5

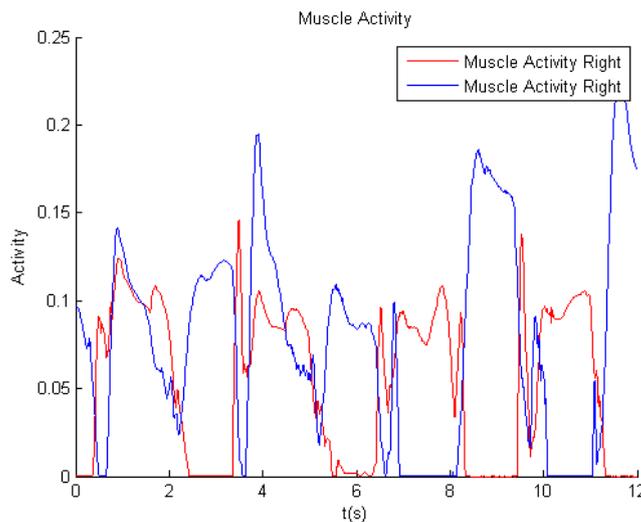


Figure 15: Estimated muscle activity of right and left erector spinae of subject 5

In both graphs bursts of enlarged muscle activity can be seen. In general the left erector spinae is most active when the upper body is leaning over to the right and vice versa. Although, especially in the EMG-signal it can be seen this trend is not always that constant.

The main difference between the graphs is that the burst are clearly seen in the estimated muscle activity where they are not that clear in the EMG-signal. Furthermore

muscle activity does not become zero in the EMG-signal where it is zero between every burst of the estimated muscle activity.

6.3 Unexpected high muscle activity

Unexpected high muscle activities are estimated by AnyBody in the Psoas Major in all subjects (Figure 16). Estimated muscle forces up to 100 percent of the maximal muscle force are experienced.

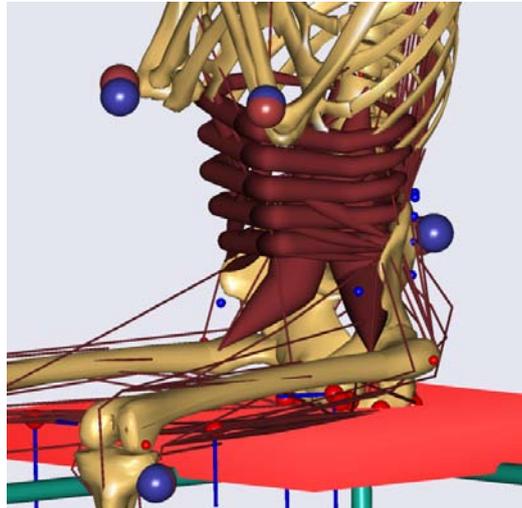


Figure 16: Model view with high estimated muscle activities of the Psoas Major

7 Model modification and additional validation

7.1 Model modification

As shown in the results of the previous section unexpected and unrealistic high muscle activity of the Psoas Major muscle is estimated by AnyBody. Further research is done to find the error in the model which causes this high muscle activity. An inadequate passive stiffness of the lumbar spine of 0.5 Nm/degree is found.

The spine is built up out of seven segments, a thorax segment, five lumbar vertebrae and a sacrum segment which is connected to the pelvis. These segments are connected to each other by spherical joints. The degrees of freedom are controlled by a spine rhythm as mentioned in the section "Method". The passive stiffness provided by the ligaments and the inter-vertebral disc is modeled as a linear torsion spring. This passive stiffness is only modeled for flexion and extension of the spine. The springs are added to all seven spherical joints and it is assumed their stiffness is equal. These torsion springs together provide an equivalent stiffness which is equal to the total passive stiffness of the lumbar spine.

The biomechanical characteristics of different regions of the human spine are studied by Busscher et al [22]. The stiffness of different sections of the spine is determined in neutral position. The stiffness of the most caudal section of the spine (L1-L4) is determined to be 0.6 Nm/degree. This section of the spine contained 4 vertebrae and 3 inter-vertebral discs. This section is modeled in AnyBody as 4 lumbar vertebrae connected by 3 spherical joints with their torsion springs. The stiffness of these torsion springs can be calculated by:

$$\frac{1}{k_{eq}} = \frac{1}{k_1} + \frac{1}{k_2} + \frac{1}{k_3}$$

where k_{eq} is the stiffness of the lumbar section (L1-L4). It is assumed the stiffness of the three springs is equal; therefore the stiffness of each of the springs has to be 1.8 Nm/degree regarding to the equivalent stiffness determined by Busscher et al. As mentioned before it is assumed all seven torsion springs are equal, therefore the equivalent stiffness of the lumbar spine is 0.25 Nm/degree. One has to mention this is half of the passive stiffness previous set in the AnyBody model.

A change of the passive stiffness of the spine changed the muscle activity in a large extent. The muscle activity of the Psoas Major, which was estimated before up to 100%, was reduced to a maximum of 35%. Furthermore it also significantly changed the estimated muscle activity of the spine muscles.

7.2 Additional validation

An additional validation is done for the modified model. The passive stiffness of the spine is adjusted to the value calculated in the previous paragraph. The same setup is used as the one of the validation experiment described in the section "Method".

It is expected the change in passive stiffness of the spine will not change the results of the validation by the force platforms significantly. Therefore the additional validation is only done by validating the estimated muscle activity with the measured EMG envelope. The validation is again quantified by calculating the correlation coefficient with its confidence interval.

The results of the additional validation are listed in Table 9 and depicted in Appendix A. A CC of 0.25 and 0.19 can be seen for the right and left erector spinae respectively. If the results of subject 3 and 6 are kept out again, the validation will lead to CC's of 0.24 and 0.19 which is higher than the CC's of the validation done before the model modification (0.10 and 0.03).

Table 9: CC and MCC of the erector spinae with corresponding CI and MCI of the additional validation

| Subject | Erector Spinae Right | | | Erector Spinae Left | | |
|-------------|----------------------|--------|---------|---------------------|--------|---------|
| | CC | CI_low | CI_high | CC | CI_low | CI_high |
| 1 | -0.12 | -0.18 | -0.07 | 0.29 | 0.24 | 0.34 |
| 2 | 0.68 | 0.64 | 0.71 | 0.52 | 0.47 | 0.56 |
| 3 | 0.16 | 0.10 | 0.21 | -0.05 | -0.11 | 0.00 |
| 4 | 0.33 | 0.28 | 0.38 | -0.18 | -0.28 | -0.13 |
| 5 | 0.56 | 0.52 | 0.60 | 0.47 | 0.42 | 0.51 |
| 6 | -0.13 | -0.17 | -0.07 | 0.08 | 0.02 | 0.13 |
| Mean | 0.25 | 0.20 | 0.29 | 0.19 | 0.13 | 0.24 |

Except these, results do not differ significantly from the results of the validation reported before. Before going into the discussion it is necessary to keep the following conclusions about the results of the validation of the estimated muscle activity by the EMG measurements in mind:

- Signal-to-noise ratio of the EMG signals is low
- Estimated muscle activity does not exceed 10% of the maximum muscle activity except for subject 5
- EMG signals are relatively low, except for subject 5 they don't exceed 80 μV
- EMG signals for subject 3 and 6 are constant and nearly zero
- In most subjects a trend of bursts of enlarged activity is seen in both right and left erector spinae, although especially in the EMG-signal they are not that clear and constant
- The EMG signal is infest with noise probably caused by the ECG-signal

8 Discussion

This section reports the discussion which is divided in two parts. First the validation method and results will be discussed. Secondly the model, experimental setup and the relation of this study to cerebral palsy will be discussed.

8.1 Validation

In general it can be stated the validation of the model by the force platforms shows good results. On average correlation coefficients are high, confidence intervals are small and MAE's are small. Although some of the results and errors need to be clarified.

Furthermore it can be stated it is difficult to validate the model by EMG measurements doing the experiment and use the setup as described in this report. The EMG-signals of the muscles in the back measured during the experiment is of that small extent that it is hard to draw conclusions out of the validation by the EMG-signals. Nevertheless the results and problems will be discussed to show which aspects have to be taken into account doing validation by EMG measurements.

8.1.1 Validation by force platforms

Validation of the total reaction forces

The total normal force and friction forces are estimated well by the model. High correlation coefficients show the shape of the different graphs are alike and a relatively small MAE shows the amplitude of the force is estimated well. One should mention that the MEA in normal direction is less than 1 kg, so only 1-2% of the total body weight. Therefore it can be stated the accelerations of the subject with respect to the force platforms is modeled well in all directions.

Validation of the reaction forces of the seat

The validation of the reaction forces of the seat gives good results for the friction forces but the normal force is not estimated that well. A mean correlation coefficient of 0.65 shows the pattern of the force is some different and a mean MAE of 40.2 N is around 5-10% of the average normal force in the seat. One can conclude that the forces are not divided in the right way between the seat and the footrest. The largest errors can be found in the tallest subjects. This can be explained by the fact that in these subjects the lower part of their upper-leg is lifted off the seat. Logically the footrest will take account for a larger part of the normal force than when the lower part of the upper-leg is supported by the seat. The contact between the seat and the human in the AnyBody model is always present; this lift-off of the upper-leg is not modeled. Therefore the estimated supporting force of the seat is too high for the taller subjects (Figure 12). To cope with this a more detailed contact model is needed.

Validation of the reaction moments of the seat

The validation of the reaction moments of the seat gives good results for the moment around the y-axis, but the moment around the x-axis (M_x) is not estimated that well looking to the correlation coefficient. Although, in Figure 13 can be seen M_x is almost constant during the experiment and therefore the correlation coefficient is not a good measurement to validate as will be explained later in the discussion. The MAE shows the

error is of a low magnitude. An explanation for these small errors can be the fact that the normal force is not divided correctly between the seat and the footrest as concluded before. Furthermore a slightly wrong placement of the upper-body can cause an error of this extent. One has to notice the moment which varies the most, the moment around the y-axis, is estimated well. It can be concluded the position of the center of mass of the upper-body with respect to the force platforms as well as the change of this position is modeled well by the model.

8.1.2 Validation by EMG

The major problem in this validation experiment is the low muscle activity. Except one subject, the maximal measured EMG does not exceed $80 \mu V$. In two subjects the muscle activity is constant and nearly zero (Appendix A). The estimated muscle activity is also low; in most of the subjects it does not exceed 10% of the maximum muscle activity. This is in accordance with the low EMG measurements; the muscles of the erector spinae show low activation.

Noise, most likely ECG signals, is measured with a magnitude of at least $25 \mu V$. This makes it difficult to distinct noise from the myoelectric signals. Solutions to this problem can be:

- to use more adequate measurement techniques, for example to make use of fine wire electrodes
- to measure EMG signals from other muscles which are assumed to be more active during the experiment
- to change the validation experiment in such a way the muscles fascicles of the erector spinae become more active

A validation based on large muscle activities would give better results than a validation based on low muscle activities because the errors in the model have relatively less influence if muscle activities are high.

It is chosen to use the correlation coefficient (CC) to quantify the validation of the muscle activity. One can ask himself if the choice to make use of a correlation coefficient to quantify validation by EMG is a good choice after all. If variables change a lot, it can give a good representation of validation. On the other hand, if variables change not too much, or are even constant during the experiment, the CC can give a wrong interpretation of how good two variables correspond to each other. For example in the results of subject 3 (Appendix A) both EMG-signal as well as muscle activity are nearly zero. So one can conclude they correspond well with each other although the CC is zero. Therefore one has to be careful using the CC as quantification for validation.

Although the problems above are present, one can draw some conclusions out of the results. Looking to the trend of the muscle activity, this means at what time a muscle is active and inactive, some good results can be seen. Especially for the subjects where higher EMG-activity is measured and more muscle activity is estimated (subject 5, Appendix A).

However the results state the correlation coefficient is relatively low, it cannot be concluded that the model estimates the muscle forces in a wrong way. A validation experiment with higher muscle activity and higher signal-to-noise ratio is needed. The analysis shows there are some promising results for a reliable validation by EMG measurements in the future if one takes the comments reported into account.

8.2 Model, experimental setup and relation to cerebral palsy

Scaling

The scaling of the model is based on the measurements of segment length variables. In general it is hard to determine the length of all these segments accurately; some landmarks are difficult to palpate on the body. Therefore it might be more appropriate to scale the model based on an optimization method. Nowadays methods are present to identify biomechanically relevant parameters in kinematically over-determinate systems for a given motion [24]. Although one has to keep in mind in the end experiments have to be done with children who suffer from CP and they might not been able to make every desired movement needed for this scaling method.

Passive stiffness of the spine

The model in this study contains only passive stiffness of the spine for one degree of freedom: flexion and extension. The results show that a change in the value of this passive stiffness has a large influence on the muscle activity. Therefore it is advised to also model the passive stiffness in the other degrees of freedom, at least for the lateral bending of the spine.

Contact model between human and support

As concluded earlier in the discussion a more detailed contact model between the human and the support is needed. In this study the contact is always present and it supports the body in the same extend regardless of the distance between the body and the support. In reality the farther away the body(part) is from the support, the less it will be supported. To solve this problem is to refine the existing contact model or to change the experiment in such a way the body parts which are assumed to be in contact with the support, are fully supported.

Number of markers

It is chosen to use the minimum number of markers necessary to constraint the degrees of freedom in the model. This choice is made to restrict the error in tracking data as explained in the section "The model". Although it is advised to use more markers than the minimum number required. The data of the extra markers can be used to examine how good the model fits the marker data. In some extend it can be used to validate the scaling of the model.

The influence of soft tissue and posture in the experiment

The results show there is a difference in muscle activity between the subjects. There are two reasons for this. First the amount of soft tissue between the bone and the pillow has a large effect on the movement of the body; the soft tissue acts like a damper. To ensure the movement of the pelvis is approximately the same for all subjects, it should

be possible to adjust the pillow. A change in inflation speed of the air which goes into the cells could be a solution for this. Secondly the difference in muscle activity is caused by the difference in posture of the person sitting on the pillow. Although the task was the same for all subjects, to sit as relaxed as possible, some of them were sitting with a straight back while others had a convex curvature of the back. The muscle activity of the subjects who were sitting with a straight back appeared to be higher compared to the others.

Investigation of muscle activity for different movement patterns

One of the goals in this study was to investigate muscle activity in the back for different movement patterns. It was not possible to analyze all the experiments with the different movement patterns and pillow settings in the given timeframe. But one could also ask himself if it is realistic to investigate different movement patterns if the estimated muscle activity by the model is not validated properly. Therefore it is advised to obtain a proper validation method first, where after the effect of variations in movement pattern can be investigated.

Influence of different parameters

It is advised to do a parameter study to examine which parameters have the largest influence on the results. One can put more effort in determine these parameters to get a model that is as accurate as possible. It is advised to study the influence of a change in the angle of the pelvis, the passive stiffness of the spine and the position of landmarks defined on the bone.

The model, CP and other applications

This report shows in some extent what is happening in the muscles in the back when postural movements during sitting are applied. One can conclude out of the results that the erector spinae shows relatively low activity and that this activity is not constant, but is varying with the movement. If the model is validated well, the next step is to do experiments with children who suffer from CP. Research has to be done on in what extend the model has to be adjusted to these children. Furthermore more research has to be done on what kind of muscle activation pattern decreases the spasticity the most. The model then can be used to find the right settings for the pillow to approach this muscle activation pattern the most.

Finally to look beyond the scope of the report, one can think of other applications to use the model. It can be used for development of dynamic seating systems, for example in cars. Furthermore it can support the research of the effect of postural movements in lower-back problems.

9 Conclusions and recommendations

9.1 Conclusions

This report contains a musculoskeletal analysis of light postural movements during sitting of six subjects and a validation of the musculoskeletal model. The results show:

- The model is validated well by the force platforms based on high correlation coefficients and low mean absolute errors.
- The total reaction forces are estimated well by the model. It can be stated accelerations of the body are modeled correct.
- Reactions of the seat are estimated well by the model, although the normal force shows error due to the fact the normal force is not divided correctly between the seat and footrest in the analysis of the taller subjects.
- The reaction moments of the seat in the horizontal plane are estimated well. It can be concluded the position of the center of mass of the upper-body with respect to the seat as well as the change of this position is modeled well by the model.
- Low signal-to-noise ratios of the EMG-signals and low muscle activities make it difficult to provide reliable results of the validation based on EMG-measurements.
- Low muscle activities of the erector spinae are estimated by the model which go up to 10% of the maximum muscle activity.
- One has to be careful using a correlation coefficient to quantify validation of muscle activity. If muscle activities are relatively constant it can give a wrong interpretation of the agreement of the two variables.
- The passive stiffness of the spine has high influence on the muscle activity of the muscles, especially on the activity of the Psoas Major muscle.
- The pillow provides a different movement of the pelvis between subject due to a difference in the amount of soft tissue and posture between the subjects.

9.2 Recommendations

Investigation of the results and literature study show several recommendations for this study:

- The contact modeled between the human and support should be adapted or the experimental setup should be changed to prevent errors in the estimated normal reactions.

- More adequate measurement techniques in EMG measurements should be used to provide a higher signal-to-noise ratio and thereby improve the validation method. Also a change in validation experiment can be done to improve the validation method by EMG-signals.
- The validation of the estimated muscle activity should not only be quantified by the correlation coefficient if muscle activities are relatively constant.
- It is advised to do a validation based on muscles with large muscle activities. This would give better results; small errors in the model have relatively less influence if muscle activities are higher.
- It is advised to scale the model based on an optimization method instead of a scaling method based on measurements of segment length variables.
- It is advised to add passive stiffness of the spine in the model in all directions.
- It is advised to use more markers than the minimum number required. The data of the extra markers can be used to examine how good the model fits the marker data.
- It is advised to obtain a proper validation method for the muscle activity first, where after the effect of variations in movement pattern can be investigated
- It is advised to do a parameter study to examine which parameters have the highest influence on the results to provide data that is as accurate as possible.

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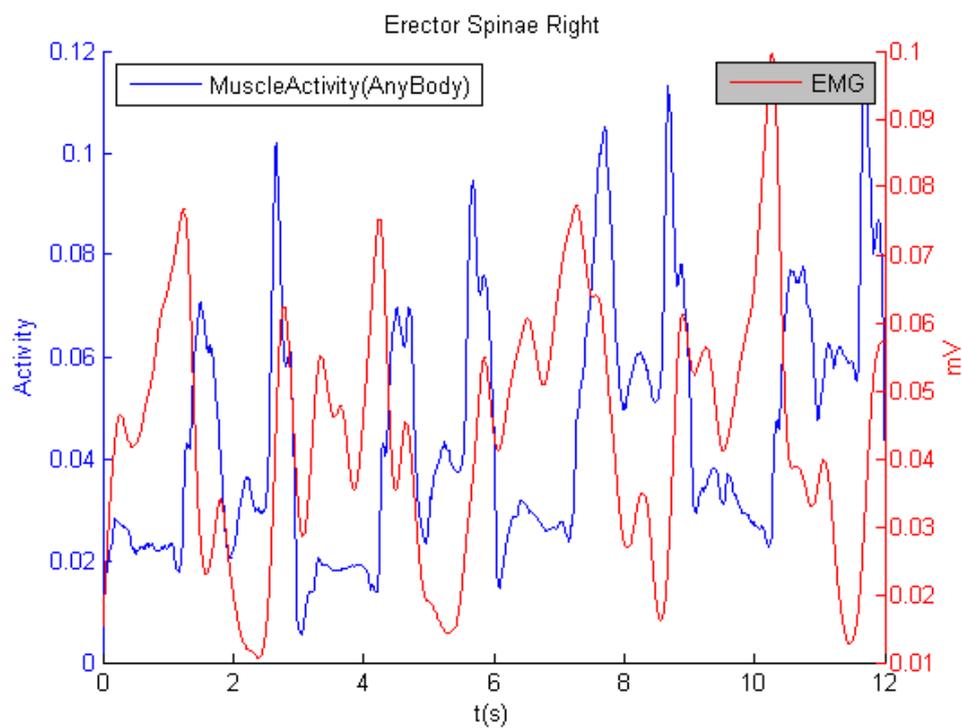
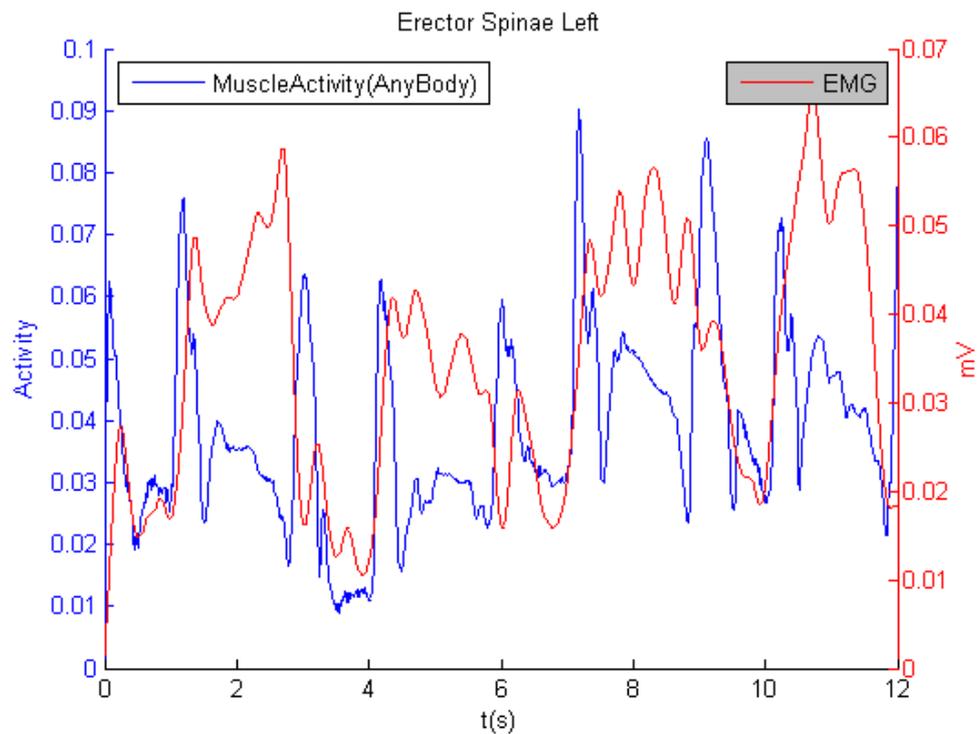
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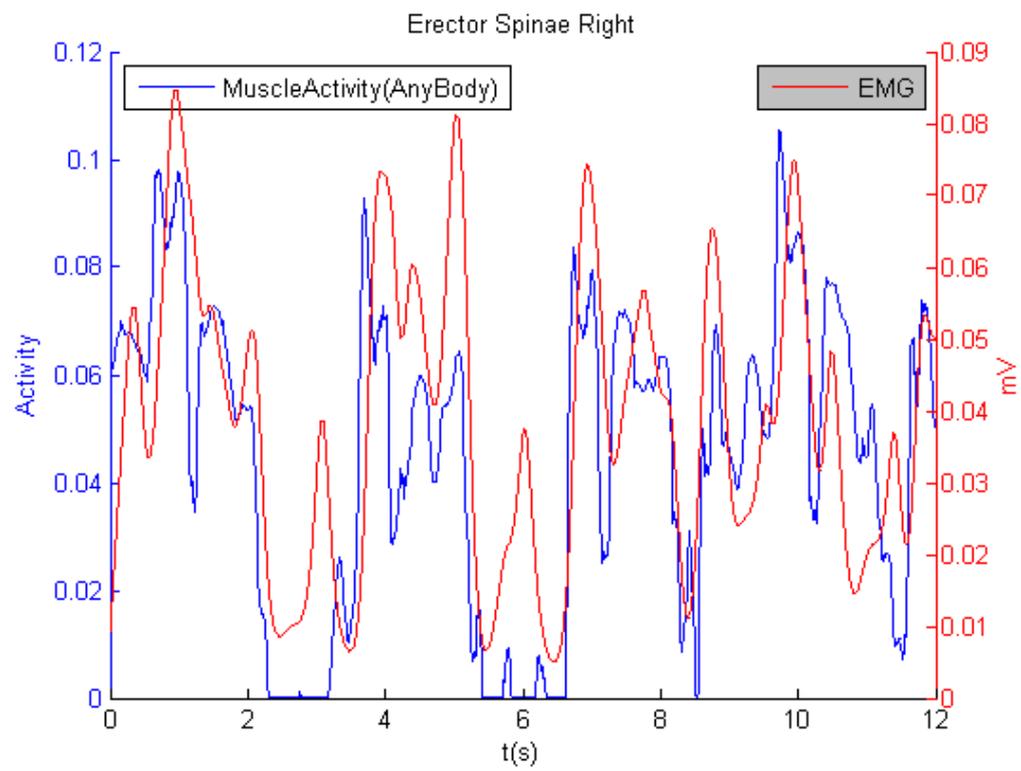
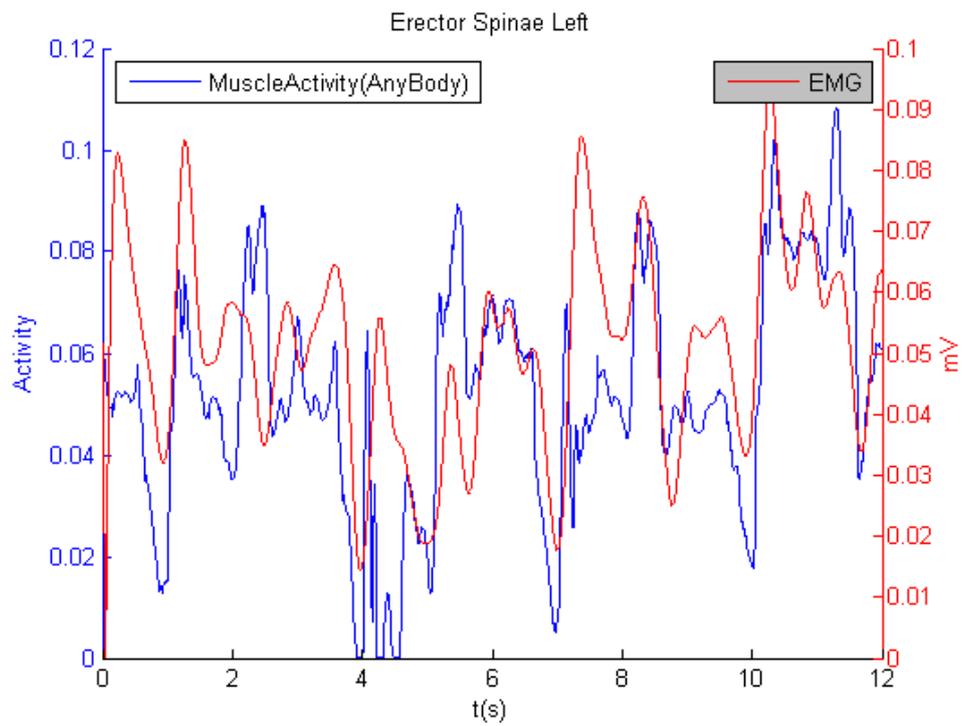
Appendix A

In this appendix the graphs of the muscle activity and the EMG envelope are shown for the additional validation experiment. The graphs are shown for all six subjects.

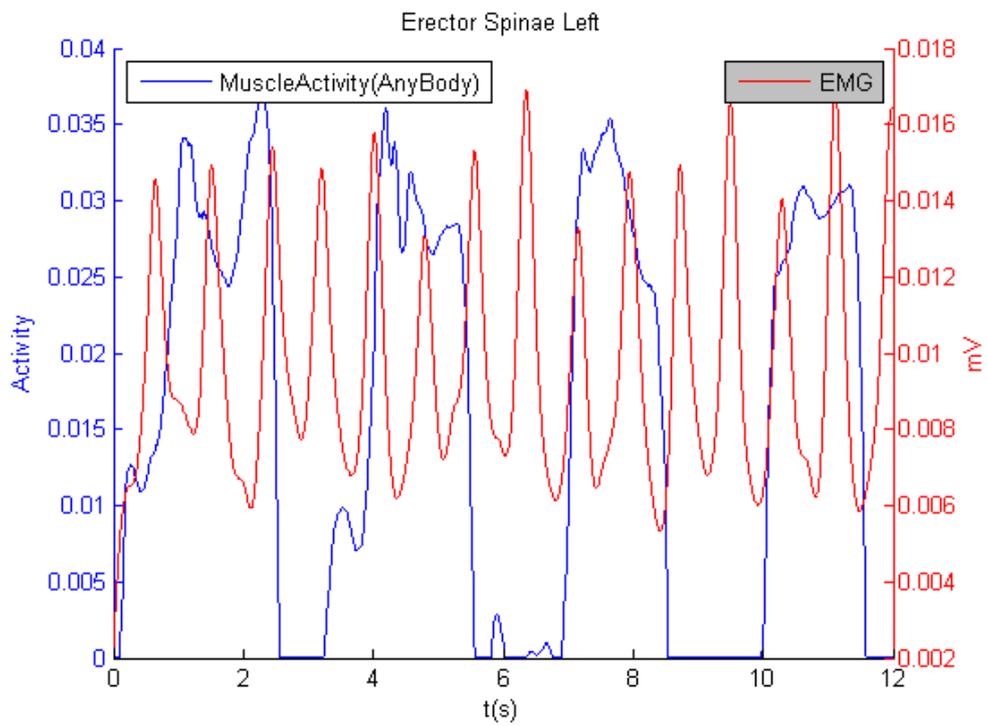
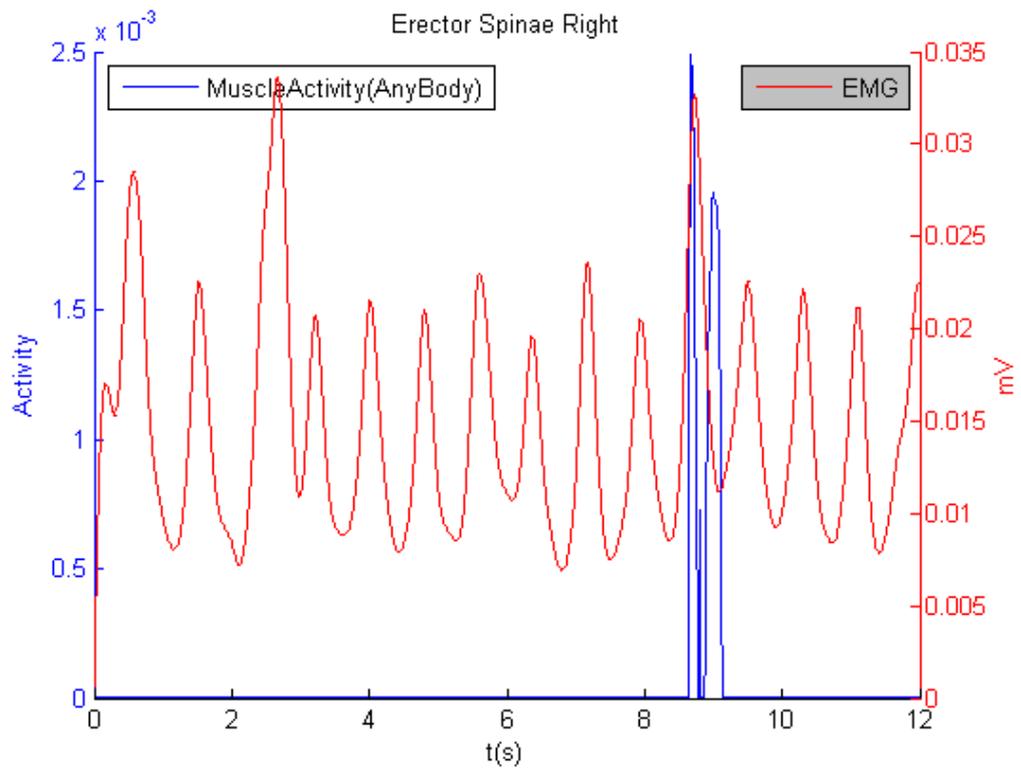
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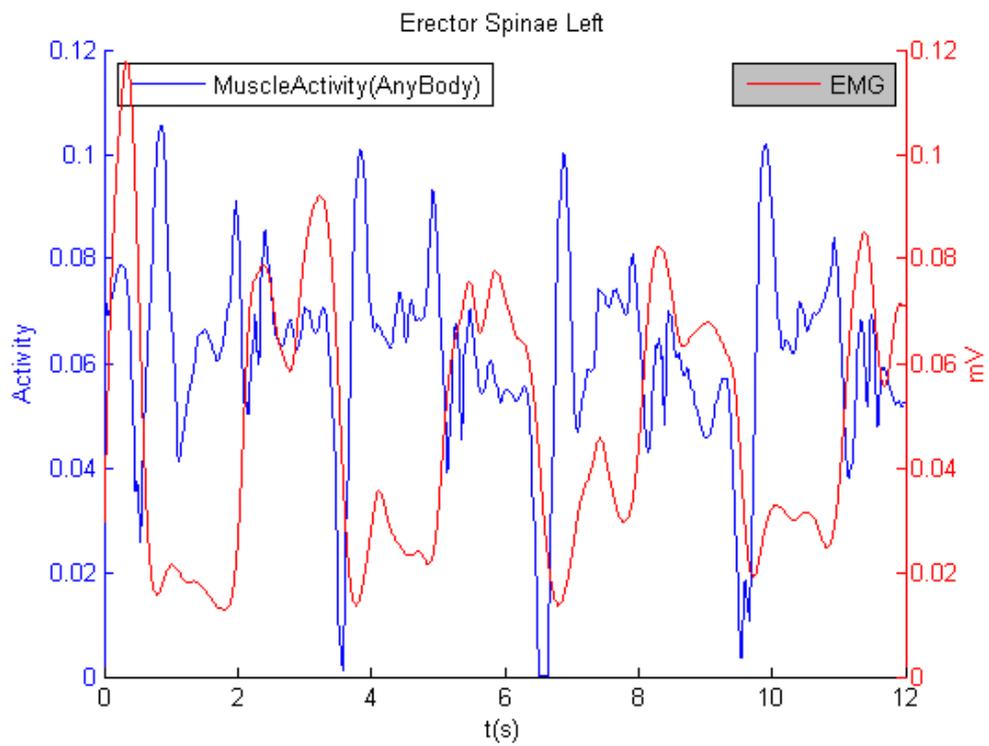
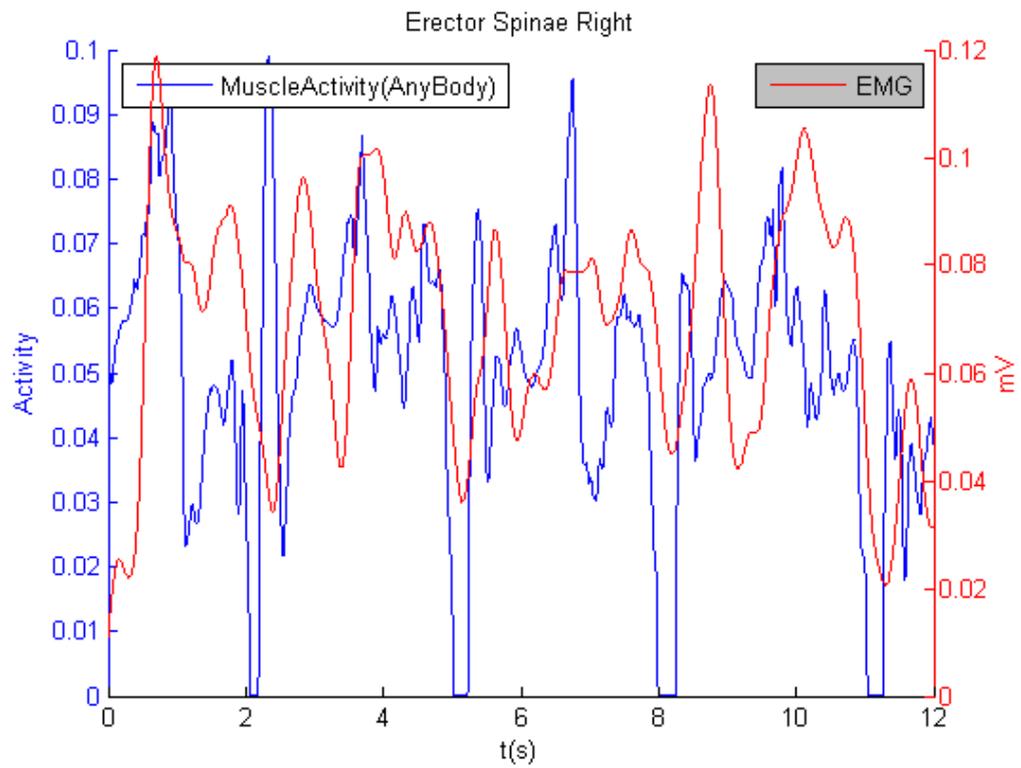
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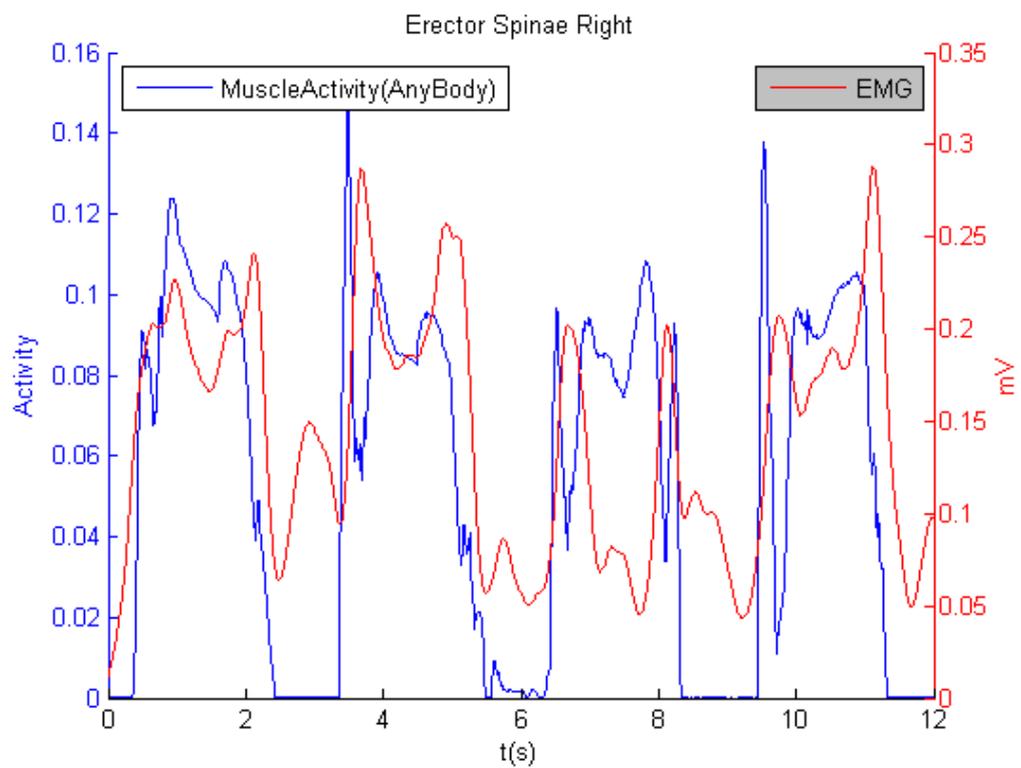
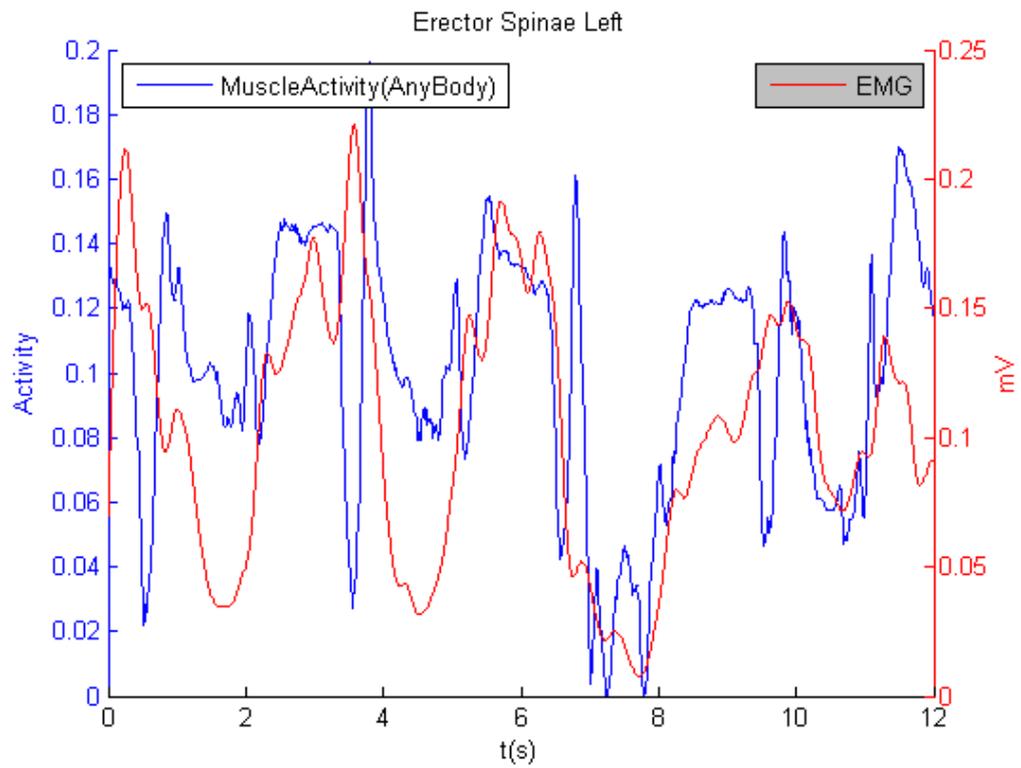
Subject 3



Subject 4



Subject 5



Subject 6

