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ENGINYERIA BIOMÈDICA

ANALYSIS OF ARM MOVEMENTS DURING ACTIVITIES OF DAILY LIVING FOR THE DESIGN OF AN ACTIVE UPPER LIMB EXOSKELETON FOR ADULTS WITH DUCHENNE

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Abstract

Adults with Duchenne Muscular Dystrophy (DMD), due to their severe muscle weakness, have problems performing basic activities of daily living (ADL). The possibility to use an arm support to perform the movements by themselves offers a new paradigm for these patients, giving them more self-autonomy and independence. However, current arm supports do not provide enough assistance for DMD adult patients. Moreover, these patients are sited on a wheelchair, so they have limited mobility, and their quality of life would be greatly improved if they had an inconspicuous and functional assistive arm support.

Flextension A-Gear project emanates with the objective to develop an active upper-limb exoskeleton for DMD patients that can be worn underneath clothing and supports the arm for the independent execution of essential activities of daily living. In this project, researchers aim to obtain an adequate balance between complexity and functionality in design of the device.

This work is carried out in collaboration with that project with the aim to find what are the biomechanical requirements for the design of an active upper-limb exoskeleton for patients that suffer Duchenne Muscular Dystrophy, which should be at the same time simple and aesthetic.

For this purpose, kinematic (joint angles, range of motion, angular velocities, angular accelerations), dynamic (joint torques) and energetic (mechanical joint powers) quantities associated to 4 tasks performed by 6 healthy subjects were analysed. Moreover, Principal Component Analysis (PCA) has been applied to the three shoulder rotations with the aim of minimizing the number of orthosis actuators at that joint.

The results obtained correspond to the biomechanical requirements for the design of an upper-limb exoskeleton for DMD patients. They show how the degrees of freedom and axes can be reduced and readjusted in order to obtain a simpler and functional model. It is concluded that the pro-supination angle can be neglected, and the three axes of rotation of the shoulder can be reduced into two without losing much functionality. Moreover, the results show that having a gravity compensation mechanism, like passive orthoses do, would reduce the required actuator powers, yielding a light and cost-efficient design.

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1. Introduction

Nowadays, thanks to the increment of the life expectancy a higher percentage of Duchenne patients reach the adult life. It is in this stage where the upper extremities present severe dystrophy preventing them to perform basic daily activities. Due to that fact, adult patients need daily assistance that is usually provided by an external person (therapist, volunteer, familiar…).

There is clearly a need to improve their daily life, which can be done through the use of an active arm support. Unfortunately, there are not currently available exoskeletons that truly cover this necessity. It is at this point where Flextension project from the University of Twente arises.

The idea behind Flextension is to develop an active upper-limb exoskeleton that can be worn underneath clothing and supports the arm for the independent execution of essential activities of daily living. In this project, researchers are looking to obtain an adequate balance between complexity and functionality.

Their first prototype of an active arm support, called A-gear (Figure 1), was designed following the suggestions given by the medical community. The prototype developed was able to reproduce approximately the whole movement of a healthy arm, but rather complex at the same time (since three motors are required to drive the shoulder motion). For the next version, there is an interest in maintaining the maximum functionality reducing its complexity. It will be just necessary to assist the essential activities of daily living (ADL), leading to a simplified arm support [1].

This work emanates from collaboration between the Biomechanical Engineering Lab (BIOMEC) from CREB and the University of Twente. It fits in the frame of the Flextension project with the function to help on the design by studying which are the biomechanical requirements needed. Concretely, the main objective of this project is to carry out a quantitative analysis of the biomechanical requirements to obtain a simpler (but still functional) design of an active arm support for people with Duchenne.

For this purpose, kinematic (joint angles, range of motion, angular velocities, angular accelerations), dynamic (joint torques) and energetics (joint powers) quantities of 4 preselected tasks performed by 6 healthy subjects were analysed. Moreover, Principal Component Analysis (PCA) has been applied to the three shoulder angles with the aim of minimizing the number of orthosis actuators at that joint.

The dissertation is structured as follows. It starts with the state of the art, which is first focused on the diseases per se, continues with the current upper limb assistive devices and finishes with the requirements aimed to be found. The next section is the materials and methods, which covers how the study was performed and how the different parameters were calculated. The results are presented in the subsequent section. And finally, there is the discussion part.

Figure 1: Active A-gear upper limb exoskeleton [1]

2. State of the art

2.1. Duchenne muscular dystrophy

Duchenne muscular dystrophy (DMD) is a degenerative disorder characterized by a proximal muscle weakness and calf hypertrophy, which is presented in early childhood. The dystrophy starts on the lower limb but, as the patient grows-up, the upper limb is also affected. It is X-linked, which means that only boys can be affected, and has an incidence of 1/3500 living male births [2].

Lately, the life expectancy of a Duchenne patient has increased considerably being able to reach the adult life [2]. The average is around 30 years old with cases of men living into their 40's or 50's. Because of that, there is an adult group of patients that lives with severe physical impairments (upper and lower limbs) and a strong dependency on care. This group needs assistance to perform daily activities as their arms get weaker and cannot perform daily tasks by themselves. Due to the reduction of the mobility, the muscles get disused causing the joints being tightened little by little and mobility problems increase. It can cause pain in every movement and loses of the arm function. It is at this point where assistive devices for the upper limb are needed [3].

2.1.1 Disorder

The disorder is caused by a mutation in the dystrophin gene, located on the human X-chromosome. This gene encodes a protein called dystrophin, which acts as a mechanical link between actin in the cytoskeleton and the extracellular matrix providing structural stability of the cell membrane. A mutation on it results in loss of function of dystrophin, obtaining a prematurely truncated, unstable dystrophin protein, which causes a degeneration of muscle fibres.

Duchenne muscular dystrophy is inherited in an X-linked recessive pattern, where the females are typically the carriers and males are the ones affected.

Although there is currently no cure for Duchenne, there is a lot that can be done to manage the symptoms and considerably improve quality and length of life. For instance, including different medication, diet, psychosocial management, physiotherapy, orthotics and surgery. In what refers to medication, steroids are proven to slow the progression of Duchenne but a long list of worrying side effects is associated to them [4].

2.1.2. Timeline

On the ages of 2 and 3 is when the first symptoms usually appear. Basically, the timeline of a Duchenne child, as it can be seen in Figure 2, would be:

- Pre-schooler (3-5 years old): seem clumsy and fall often. Have trouble climbing stairs, getting up from the floor or running.
- School age (5-7 years old): walk on their toes or the balls of their feet with a slightly waddling gait, and fall frequently. To keep their balance, they may stick out their bellies and pull back their shoulders. They start having difficulty on raising their arms.
- 7-12 years old: it is at this stage where most of them start to use a wheelchair. This is a gradual process, at the beginning the chair is only required to conserve the child's energy when covering long distances and the arms move it. But soon a fully powered wheelchair substitutes it.
- 10 years old: activities involving arms, legs or trunk may require assistance or mechanical support. Also it is at this age when respiratory problems start, diaphragm and other muscles that operate the lungs may weaken, making the lungs less effective at moving air in and out.
- 14 years old: one-third boys of that age suffer a dilated cardiomyopathy and by 18 years old it is present in all patients. The first evidence of it can be found after 10 years of age. It produces beats rhythm alteration and conduction disturbances.

From this point, all the symptoms get worse having more troubles on movement. In the case of the arms, the weakness starts in the proximal parts and finally reaches the distal ones. There is not an agreement of the exact age at which the upper limb gets drastically affected.

Furthermore, some of the Duchenne boys present some degree of learning disability, where the main areas affected are: attention focusing, verbal learning and memory, and emotional interaction [2,3].

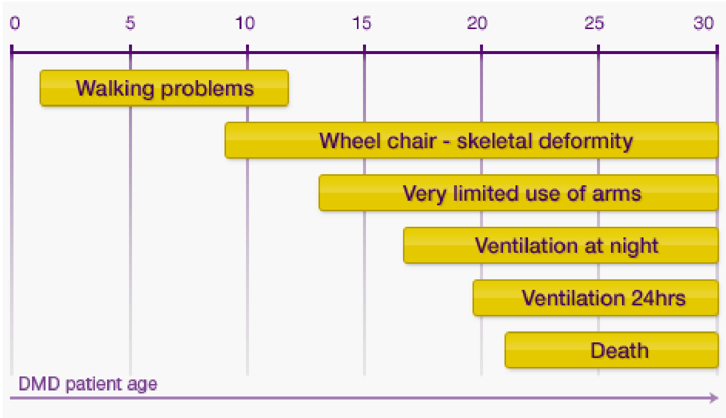


Figure 2: Duchenne symptoms time-line [5]

2.2. Arm orthoses for Duchenne patients

When talking about arm supports for Duchenne's patients, it is important to make a distinction according to the degree of affection of the upper limb. During the first stages of arm dystrophy, usually during the teenager period, patients have still some strength on their arms and just counteracting the weight force is enough to have mobility. In this case, **passive supports** are used. These orthoses just compensate the gravity, meaning that they withdraw the weight of the arm, facilitating the movement. In these exoskeletons, the movement is totally controlled naturally by the arm of the patient and most of them are mounted to a wheelchair.

But when the muscular weakness is higher due to the continuous degeneration of the muscle, it leads to a patient with very reduced strength and inability to move the arm. In which, just with gravity compensation is not enough to allow performing the movements, and then an active arm support is needed. This happens, generally, on the adult stage of life.

Active supports consist on a number of actuators that drive the motion of the exoskeletal robot joints. Depending on the quantity of actuators the device will have more or less degrees of freedom adjusting better or worst to the human natural movement. However, more actuators imply heavier and bigger exoskeletons. As in the passive ones, the devices can be attached or not to the wheelchair. Patients report that wearable devices are more comfortable due to the fact that they allow unconstrained trunk movement. [6]

Another fact to consider when designing these orthoses is the way they are controlled in order to replicate the natural movement accurately. Some controllers include the use of brain computer interfaces (BCI), which use EEG or brain hemodynamic signals; muscle activation interfaces, based on EMG; muscle-contraction interfaces, which evaluates muscle vibration, muscle dimensional change, radial muscle force and stiffness, muscle force and muscle hemodynamic; movement interfaces, which analyse body segment movement and relative joint movement; force interfaces and finally, interfacing with parallel systems such as the eye, the tongue, the head, the speech, the hand, among others. It is also possible to use a combination of these interfaces, having what it is called hybrid control interfaces [7,8,9].

Regarding the current available arm supports, van der Heide et al. [10], Dunning and Herder [7] and Mahoney et al. [11] did an interesting overview of what can be found in the literature. Figure 3 shows some of the current available devices that can be used by Duchenne patients. These current devices have two main limitations: are insufficient at the last stages of the disease, and are often highly stigmatizing due to their large dimensions [12].

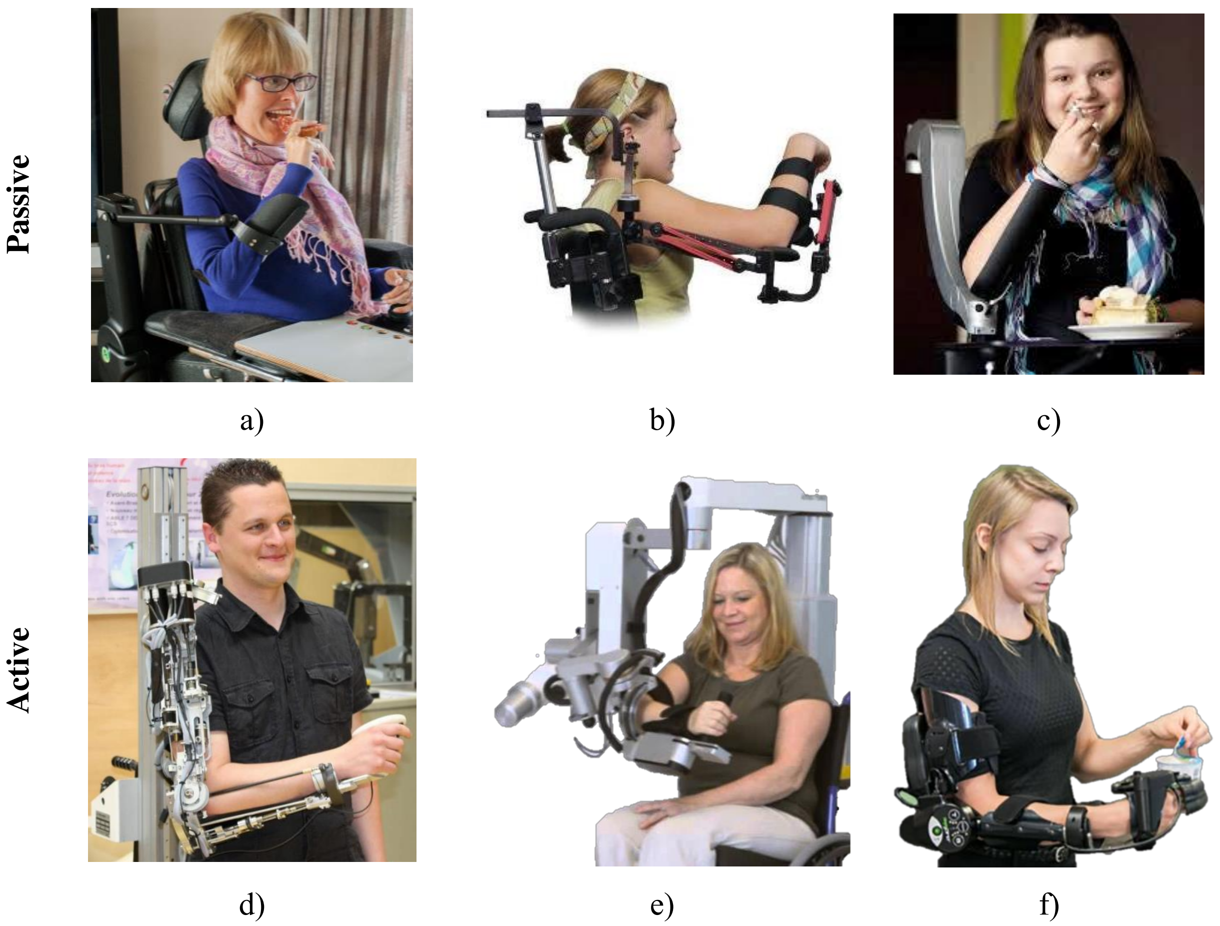


Figure 3: Examples of commercially available arm supports. a) Armon Pura (Microgravity Products, the Netherlands), b) WREX (JAECO Orthopedic, USA), c) Darwing (Focal Meditech, Netherlands), d) Able (Haption, France), e) Armeo Power (Hocoma AG., Switzerland), f) MyoPro (MyomoInc., USA) [13-18].

2.3. Biomechanical requirements

There is very few literature on formal studies carried out to determine the biomechanical requirements of an upper limb assistive device and specifically for Duchenne or patients with muscular dystrophy. Concretely, four studies assessing some of the biomechanical requirements for ADL were found. Three of them study the kinematics and one focuses also on the dynamics.

The first is the one by Romily et al. [19], in which the range of arm joint angles for a variety of common tasks was evaluated in order to determine which of the angular degrees of freedom could be collapsed for the purpose of designing an anthropomorphic orthosis. They evaluated the range of motion (RoM) of 7 joint angles in 22 tasks, including shoulder, elbow and wrist.

The second is from Ramanathan et al. [20], where the trajectories of the elbow were analysed while doing different activities of the daily living in order to determine the elbow-position envelope. 9 tasks were performed and 2 joint angles (elbow flexion and lower arm elevation) were studied.

The next one, by Magermans et al. [21], was aimed to obtain the RoM of the shoulder and elbow for a selection of ADL. It is oriented to help in rehabilation practice to determine the functional capacity of patients according to the RoMs obtained. The authors divided the tasks in range of motion tasks (7 tasks), where the aim was to reach a maximal joint angle, and ADLs tasks (5 tasks).

The last work is the only that studies dynamics and kinematics. It was elaborated by Rosen et al. [22]. The aim of this research was to study the kinematics and the dynamics of the human arm during ADL for the design of a 7-degree of freedom (DOF) powered exoskeleton for the upper limb. Angles, angular velocities and accelerations and finally, total, gravitational and inertial torques were calculated. It is the most complete study, 24 ADL tasks and 9 general motions, where the idea was to reach a maximal joint angle.

The main problem on applying directly these studies for the design of an upper limb exoskeleton for Duchenne patients is that they cover activities of healthy people that differ from the essential ones required for patients with DMD. Even though the one from Ramanathan et al. is focused on patients that suffer neuromuscular diseases, it only studies the trajectories and angles of the elbow without analyzing the other upper limb.

When designing an orthosis, other requirements apart from the biomechanical ones are needed. Some of these include comfort, easy donning and doffing, force transmission to the body, adjustability to the body, functionality, aesthetics, inconspicuousness, etc. [23,24].

3. Materials and methods

Broadly, the experimental part of the project consists on recording 4 different motions, which represent different activities of the daily living, using a motion capture system and extract the cinematic and dynamic parameters through Opensim. Finally, the data obtained is processed and analysed leading to the biomechanical requirements previously specified: range of motion, axis of rotation, angular velocity, angular acceleration, torques and power.
This section covers a deep explanation of the study performed, going step by step.

3.1. Participants

Six healthy subjects, 4 males and 2 females, participated in the study. All of them were right-handed and in ages between 22 and 32 years. Table 1 presents a summary of some physical characteristics of the volunteers.

Participant	Age (years)	Gender	Height (m)	Weight (kg)
1	24	Male	1.84	84
2	25	Female	1.60	54
3	25	Male	1.60	65
4	32	Male	1.70	58
5	27	Male	1.76	66
6	22	Female	1.73	60

Table 1: Participants features.

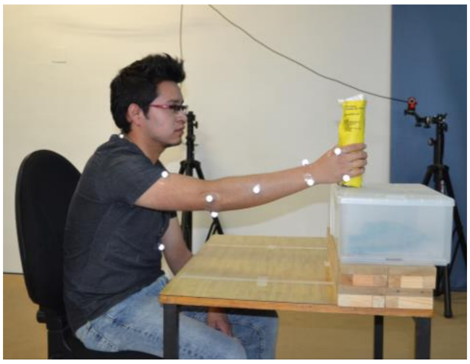
3.2. Experimental task

First, several ADLs of Duchene patients were obtained from the literature, in which in some of the cases the activities were graded according on the relevance and difficulty [25]. The most important movements were eat, drink, scratch, pick or slide an object, dress, wash hands, among others listed in Table 2.
In order to simplify the captures, the acquisition and processing, all the activities were grouped in 4 different tasks: taking an object to the mouth (labelled as Mouth), picking an object from a shelf that has the same height as the shoulder (labelled as Shoulder), simulating to zip a jacket (labelled as Body) and pick an object from the other side of the table (labelled as Table). The object used was a water bottle. In Table 2 there are all the interesting ADLs classified in these 4 groups and in Figure 4 there is a picture of the movements studied. An initial position was set for all the subjects and tasks. The two possible initial positions, with and without bottle are shown in Figure 4 e) and f). The pro-supination angle was not set up identically for all the subjects and all the activities.

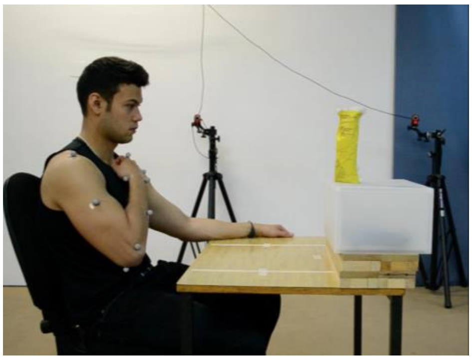
MOUTH	SHOULDER	BODY	TABLE
Eat	Pick /keep objects	Scratch	Eat
Drink	Give a hug	Dressing	Drink
Phone		Use a computer	Pick objects
Scratch		Play videogames	Slide objects
Wash teeth			Wash hands
Brush hair			Dry hands
			Write
			Use a computer
			Play table games
			Open a door

Table 2: Classification of ADLS in the 4 tasks studied

a) Mouth

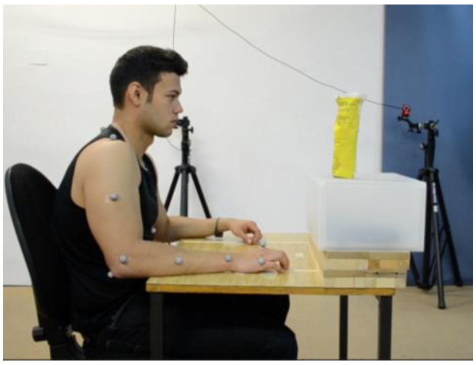


b) Shoulder



c) Body

d) Table



e) Initial position

f) Initial position

Figure 4: Example of each motion studied: a) taking an object to the mouth (Mouth), b) picking an object from a shelf that has the same height as the shoulder (Shoulder), c) simulating to zip a jacket (Body), d) pick an object from the other side of the table (Table), e) initial position without object and f) initial position with object.

3.3. Biomechanical model

The model used consists on 9 solids: thorax, humerus right / left, ulna right / left, radius right / left and hand right / left, and has 16 degrees of freedom. The thorax is the reference body, which is linked to the ground (inertial frame). So, it has 6 degrees of freedom with respect to the ground, 3 rotations and 3 translations. The other degrees of freedom of the model are the relative rotations between the different bodies that constitute the model. The model is presented in Figure 5 and the specifications are explained in Table 3.

The movement of the fingers and the wrist is not studied due to two main reasons: the considered arm support will not articulate the fingers, and those are the last part affected by the disease. Although just one arm is analysed, both arms are modelled equally.

Joint	Parent solid	Child solid	DOF	Coordinates	
Ground - thorax	Ground	Thorax	6	Back tx Back ty Back tz	Back tilt Back list Back rotation
Shoulder	Thorax	Humerus	3	Shoulder flexion Shoulder abduction Shoulder rotation	
Elbow	Humerus	Ulna	1	Elbow flexion	
Radio-ulnar	Ulna	Radius	1	Pro-supination	

Table 3: Model specifications. Just one limb is presented but in the model there are two identical limbs, right and left.

The model was developed after an adaptation and modification of one of the available models on the bibliography. Concretely, from the *Upper and Lower body* model, a full body model that was created from two previous ones: the *Gait2354* developed by Delp et al. [26] for the lower limb and the *UpperExtremityModel* done by Holzbaur et al. [27] for the upper limb. For this study the lower limb was subtracted.

For the dynamics the model used contains the inertial parameters of each body. In this specific case, as most of the motions studied imply a weight, the dynamic part parameters of the hand segment had to be modified. The option chosen to include the effect of the object in the analysis was to have a hand segment that consisted on the hand plus the object, in this case the bottle used. So, some parameters of the body hand such as the mass, the centre of mass location and the moments of inertia were recalculated and modified. The calculus performed to obtain the new parameters is explained in the following subsection.

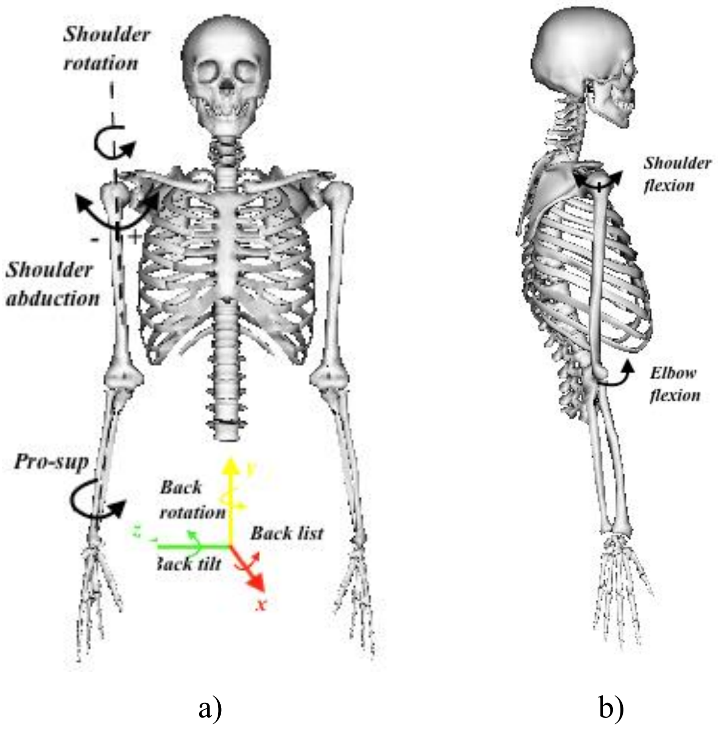


Figure 5: Skeletal model used in OpenSim showing the joint angles. a) Frontal view, b) Lateral view

3.3.1 Inertia of the new solid hand calculation

The new weight is just the current weight of the hand plus the weight of the bottle. The calculation of the centre of mass and the inertial parameters is more complex.

For the centre of mass equation 1 is applied, which from the mass of the two solids and their respective positions of their centre of mass computes the new common one. In this case, it is assumed that the position of the new centre of mass only change in the z direction, maintaining the x and y coordinates. So, applying equation 1 in this particular case where $Z_{hand}$is zero and Z_{bottle} is the radius of the bottle but has negative direction, the final formula used is equation 2.

About the new moments of inertia of the new solid, Steiner theorem is applied. Steiner theorem determines the moment of inertia of a rigid body about any given axis, given that moment of inertia about the parallel axis through the centre of mass of an object and the perpendicular distance between the axes. The expression that relates both moments of inertia is showed in equation 3.

In the model, the inertia tensor is a diagonal matrix, so only has I_{xx}, I_{yy} and I_{zz} components. Being these the ones to be modified. In 3D the perpendicular distance is calculated as the sum of squares of the two perpendicular coordinates to the axis studied. Finally, to obtain the different inertia components of the new solid, inertia of the bottle and the hand in the new centre of mass has to sum up. In summary, equation 4 is applied. Remember that only Z_{CM} is different than zero.

OpenSim gives the values of the tensor of inertia of the hand, so the only parameters missing in equation 4 are the tensors of inertia of the bottle. The bottle is modelled as a cylinder which inertia parameters are described in equation 5.

For a better understanding of the calculations and assumptions made, see figure 6.

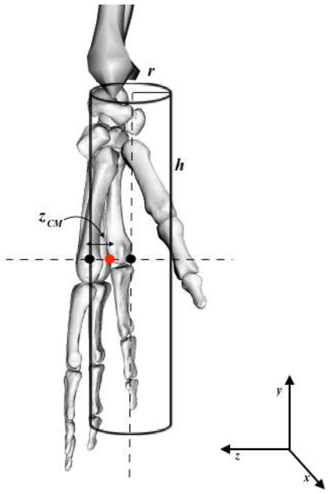


Figure 6: Schematics of the new solid: hand + bottle. The black dots are the centre of mass of the hand and the cylinder, respectively and the red one is the new centre of mass of the whole solid.

$$Z_{CM} = \frac{m_{hand} \cdot Z_{hand} + m_{bottle} \cdot Z_{bottle}}{m_{hand} + m_{bottle}} \quad \text{(Equation 1)}$$

$$Z_{CM} = \frac{-m_{bottle} \cdot r_{bottle}}{m_{hand} + m_{bottle}} \quad \text{(Equation 2)}$$

$$I_d = I_G + md^2 \quad \text{(Equation 3)}$$

$$\begin{gathered} I_{xx,total} = \left[I_{xx,h} + m_h \cdot z_{CM}{}^2\right] + \left[I_{xx,b} + m_b \cdot (r_b - |z_{CM}|)^2\right] \\ I_{yy,total} = \left[I_{yy,h} + m_h \cdot z_{CM}{}^2\right] + \left[I_{yy,b} + m_b \cdot (r_b - |z_{CM}|)^2\right] \\ I_{zz,total} = \left[I_{zz,h}\right] + \left[I_{zz,b}\right] \end{gathered} \quad \text{(Equation 4)}$$

$$\begin{gathered} I_{xx,b} = I_{zz,b} = \frac{1}{4} m_b r_b{}^2 + \frac{1}{12} m_b h_b{}^2 \\ I_{yy,b} = \frac{1}{2} m_b r_b{}^2 \end{gathered} \quad \text{(Equation 5)}$$

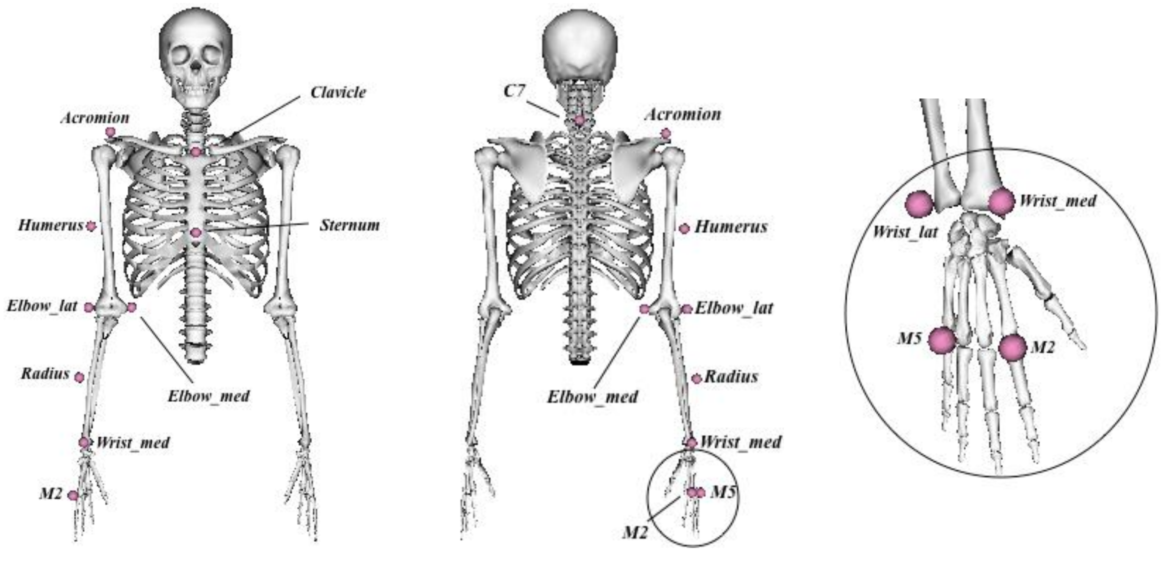
Finally, the model was scaled and adjusted per subject and so, the dynamical parameters of the hand were calculated independently for each one.

3.4. Markers protocol

Twelve markers are placed in the subject in order to capture the different motions analysed. The number and locations of the markers were selected following the International Society of Biomechanics (ISB) recommendations that are based on the use of body landmarks to place the markers [28]. Body landmarks are points easy to find and close to the bones. So, these points do not have mobility associated to soft tissues, or it is very reduced.

The markers on the scapula were not applied in this model because its motion is not analysed in detail, and the fact that the subject was sit on a chair might have blocked the view of these markers for the cameras. Moreover, two additional markers were placed in the middle of the segments as in the Helen Hayes model [29]. These two markers were added because three markers per segment are normally used in order to minimize the errors of the motion capture.

Finally, as just the motion of the right arm is studied, the markers are only placed in the right part of the body. Markers setup is shown in Figure 6 with the corresponding names.



a) b) c)

Figure 7: Markers setup. a) Frontal view, b) Back view, c) Zoom on the hand

3.5. Data acquisition

Data was acquired in the Biomechanics Laboratory of the Polytechnic University of Catalonia (UPC) using the *OptiTrack*^{TM} system of NaturalPoint. This equipment consists on a set of infrared cameras, 17 in this work, which are able to capture the 3D position of the different markers over time. The model of the cameras is the V100:R2 (Figure 7a) that has a sample frequency of 100Hz, that is one measurement every 0.01 seconds.

The markers are small reflective spheres (Figure 7b) that reflect the infrared light emitted by the cameras. This light is captured by an optical system of the cameras that determines the position of the markers on the perpendicular plane to the optical axis of the camera. From the information of the 17 cameras, the system computes the position of the markers at each instant of time.

To have accurate results it is important to calibrate the equipment before doing the captures. In this study, each motion was captured several times until a clear one was obtained. Finally, one trial per person was analysed.

a) b)

Figure 8: Equipment used to capture the data. a) Camera, b) Marker.

3.6. Data processing

From the motion captures, the positions in the coordinates X, Y and Z over the time were known and used to calculate the kinematics and dynamics of each motion.

The model described in Section 3.3 was first scaled and then adjusted in OpenSim according to each subject, in order to have the exact body measurements and markers placed in the same location as in the capture. Once the model per person was obtained, the motion data were imported to OpenSim in order to compute the kinematics and dynamics associated to each motion. This process was repeated for all the subjects.

In terms of kinematics, relative or joint angles were extracted. The ones of interest were: shoulder flexion, shoulder abduction, shoulder rotation, elbow flexion and pro-supination. In what concerns to the dynamics, the different joint torques were obtained. The considered torques were again the ones associated to shoulder flexion, shoulder abduction, shoulder rotation, elbow flexion and pro-supination. Regarding energetics, mechanical joint power, which is necessary to select the actuator needed for an assistive orthosis, was computed.

All this process was done using OpenSim. It is a powerful free-software for modelling and simulating human movement used to uncover the biomechanical causes of movement abnormalities and to design improved treatments. Since its development in 2006, researchers have used OpenSim to address fundamental issues in movement science focusing on critical areas of rehabilitation medicine, including stroke, spinal cord injury, cerebral palsy, prosthetics, orthotics, and osteoarthritis.

3.7. Data analysis

Once the data are processed, two sets of values are obtained. The first one contains the relative angles of the motion along time and the other the joint torques along time. Using Matlab these values were translated into the desired biomechanical requirements. Table 4 summarizes all the aimed biomechanical requirements.

One additional essential parameter is the number of degrees of freedom, which will determine the number of actuators needed. This parameter is not directly obtained after the analysis, just an idea of which are the most important ones to be assisted will be given. In such a way that the least important may not need to be assisted for this particular case, leading to a simplified exoskeleton. In this case, where shoulder and elbow are the assisted joints, the maximum number of degrees of freedom is 5.

<table>
<tr><th></th><th colspan="2">Requirement</th><th>Definition</th></tr>
<tr><td rowspan="4">Kinematics</td><td colspan="2">Range of motion (RoM)</td><td>Total angle variation in absolute value during a certain motion or task. Also minimum and maximum angles are given.</td></tr>
<tr><td colspan="2">Instantaneous axis of rotation (IAR)</td><td>Straight line defining the direction of the relative angular velocity between segments at each time.</td></tr>
<tr><td colspan="2">Angular velocity</td><td>Change in the angle per unit time.</td></tr>
<tr><td colspan="2">Angular acceleration</td><td>Change in the angular velocity per unit time.</td></tr>
<tr><td rowspan="3">Dynamics</td><td rowspan="3">Torques</td><td>Total</td><td>Total torque needed at the joint level to perform a certain task.</td></tr>
<tr><td>Inertial</td><td>Torque needed at the joint level to perform a certain task subtracting the effect of gravity (simulating the needed torque if a gravity compensation mechanism was used).</td></tr>
<tr><td>Gravitational</td><td>Torque produced by the gravity force.</td></tr>
<tr><td>Energetics</td><td colspan="2">Powers</td><td>Amount of energy consumed per unit time. It is the scalar product of joint torque and relative (or joint) angular velocity. Total, inertial and gravitational powers are computed.</td></tr>
</table>

Table 4: Summary of all the biomechanical requirements calculated with their correspondent definition.

3.7.1. Kinematics

Primarily, let's focus on the kinematics part in which the aim is to know the range of motion of joint angles and evaluate the IAR of the shoulder in order to know if less than 3 actuators are enough to reproduce the movements with certain precision.

3.7.1.1. Range of motion

Before analysing the overall range of motion it is necessary to know if all the subjects did all the motions similarly and detect and eliminate any possible outlier. In order to do this, the time has to be normalized from 0 to 100% of motion cycle in each case. When all the motions are normalized in time it is possible to compare them. So, a plot including the evolution of each joint angle along time per subject and motion was done.

The range of motion per coordinate was examined in two ways: per motion and the total, taking into account all the subjects and tasks. The one calculated per subject, includes the overall range of motion done per each motion considering all the participants together. Similarly, the total range of motion encompasses all the motions and volunteers.

To visualize the results in an easy way, the minimum and maximum ranges in each case were found and illustrated using box plots.

3.7.1.2. Shoulder instantaneous axis of rotation

For every time-step there is an instantaneous axis of rotation, which is a normalized vector that has the same direction as the joint angular velocity in that instant. It is calculated as:

$$\{\bar{e}_{IAR}\} = \frac{\{\bar{\Omega}^S_{Ref}\}}{\left\|\bar{\Omega}^S_{Ref}\right\|} \text{ where } \{\bar{\Omega}^S_{Ref}\} = \dot{\psi}\{\bar{e_1}\} + \dot{\theta}\{\bar{e_2}\} + \dot{\varphi}\{\bar{e_3}\} \qquad \text{(Equation 6)}$$

$\dot{\psi}, \dot{\theta}, \dot{\varphi}$ are derivatives of the relative angles , which are defined as a set of three Euler angles, and $\{\bar{e_1}, \bar{e_2}, \bar{e_3}\}$ are the axes associated with each Euler rotation. ψ is the flexion – extension angle of the shoulder, θ is the abduction – adduction angle of the shoulder, and φ is the internal – external rotation angle of the shoulder. $\{\bar{e_1}\}$ is the first Euler axis, fixed to the reference, in this case the trunk, and it is the one associated with flexion; $\{\bar{e_2}\}$ is the second Euler axis, perpendicular to $\{\bar{e_1}\}$ and $\{\bar{e_3}\}$, and it is the one associated with abduction; and finally $\{\bar{e_3}\}$ is the third Euler axis, fixed to the segment, and it is the one associated with internal rotation of the solid. Equations (7) to (9) provide the direction of the previous axes and Figure 8 shows the direction of these axes when consecutive Euler rotations are applied.

In the initial configuration, flexion it is caused by a rotation along the z-axis, abduction along the x-axis (x’-axis in a general configuration), and internal rotation along the y-axis (y’’-axis in a general configuration).

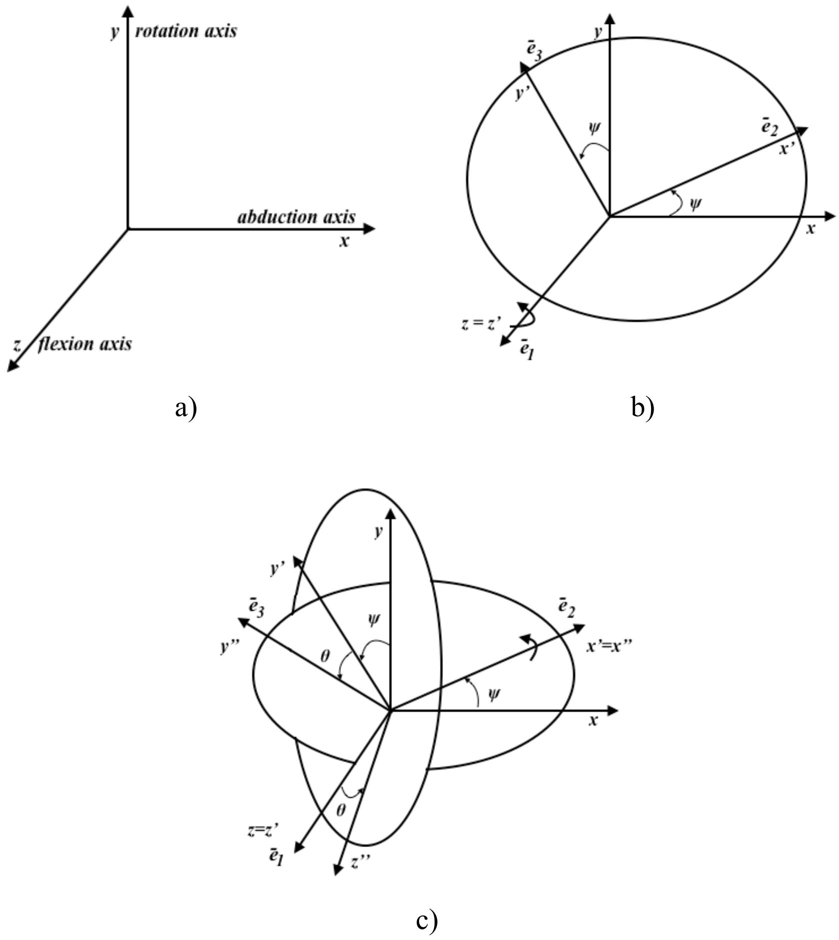


Figure 9: Euler angles. a) Axes before any rotation –initial configuration–, b) Rotation along z-axis, flexion, c) Rotation along x' axis, abduction. A rotation along y'' axis, does not affect the other Euler axes.

$$\{\bar{e_1}\} = \begin{Bmatrix} 0 \\ 0 \\ 1 \end{Bmatrix} \qquad \text{(Equation 7)}$$

$$\{\bar{e_2}\} = \{\bar{e_3}\} \wedge \{\bar{e_1}\} \qquad \text{(Equation 8)}$$

$$\{\bar{e_3}\} = \begin{Bmatrix} -cos\theta \cdot sin\psi \\ cos\theta \cdot cos\psi \\ sin\theta \end{Bmatrix} \qquad \text{(Equation 9)}$$

Once all the instantaneous axes of rotation for all the motions and participants are known, a Principal Component Analysis (PCA) is performed in order to find which are the principal axes. PCA gives as a result the principal components of the data analysed and how much of the variation in the data is explained by each of them. They are sorted according to the importance. In this particular case, as the motions are mainly symmetric, the backward axes are flipped into forward in order to reduce the variability. As a result, three principal components with their corresponding percentage of variation are obtained. Each principal component corresponds to a new axis. The idea of applying this method is to find how many axes are needed to explain the majority of the variation, and see if it is possible to use less than three actuators to drive the shoulder joint.

3.7.1.3. Angular velocity and angular acceleration

The angular velocity is obtained through numerical derivation of the angles with respect to time. Similarly, angular acceleration is the time derivative of the angular velocity.
For these two magnitudes, the interesting values are the mean and the maximum value per joint coordinate.

3.7.2. Dynamics and energetics

Regarding dynamics and energetics, torques and powers are computed in order to help in the selection of the actuators and batteries for the active arm support.

3.7.2.1. Joint torque

OpenSim also allows the inverse dynamic analysis, which allows to obtain directly the corresponding joint torques. As in the case of the angles, the torques of the different subjects and motions are evaluated. Apart from the torque evolution, the maximum and mean torques are computed. Moreover, it is interesting to distinguish between forward and backward motion of the arm, which means to differentiate between flexion and extension, abduction and adduction and finally, internal and external rotation.
The torques calculated by the inverse dynamics tool of OpenSim are the sum of the torques associated with the gravitational terms plus the torque associated with the motion, called inertial torque. It is interesting to know what is the torque caused by the motion, called inertial torque, and the one caused by the gravity, named gravitational torque. If the exoskeleton includes gravity compensators, as some passive supports do, just the inertial torque would be needed to actuate.
In order to compute the inertial torque a force equivalent to the gravity (segment mass times gravity), but with opposite direction, was added to each segment during all the motion. Figure 9, shows how these forces are added to the model. To know the gravitational torque, inertial torque is subtracted from the total joint torque.

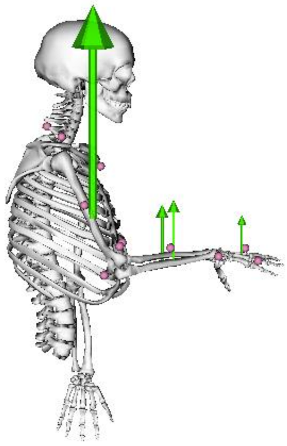


Figure 10: External forces added to the model to compensate the gravity effect. The force is the gravity multiplied by the mass of each segment.

3.7.2.1. Joint mechanical power

Other important feature is the power needed in each case, which is obtained using equation 5. Knowing the joint mechanical power is important for selecting the different actuators.

$$\dot{W} = \overline{M} \cdot \overline{\Omega} \qquad \text{(Equation 5)}$$

where $\overline{M}$ is the joint moment or torque and $\overline{\Omega}$ the angular velocity.

The evolution of the power versus time, as well as, the mean and maximum powers, are calculated.

Similarly to the torques, the power calculated as it is obtained multiplying the overall torque and velocity, is the total power. Then, from the new torques calculated, the inertial ones, inertial power is obtained, and from the gravitational torque, the gravitational power is computed.

It is necessary to mention that the most critical case, when more force is needed, was the one analysed. That implies that for the analysis, the object is present during all the motion, instead of just half of the cycle, in Mouth, Shoulder and Table tasks. Moreover, the object is not considered to be sliding on the table in Table task. It is studied as it was moving on the free space. These assumptions also simplify the analysis.

4. Results

Following the scheme of the methodology, this section is split in two basic parts: kinematics, which includes range of motion, axes of rotation and angular velocity and acceleration; and dynamics and energetics, which encompasses torques and powers.

4.1. Kinematics

4.1.1. Range of motion

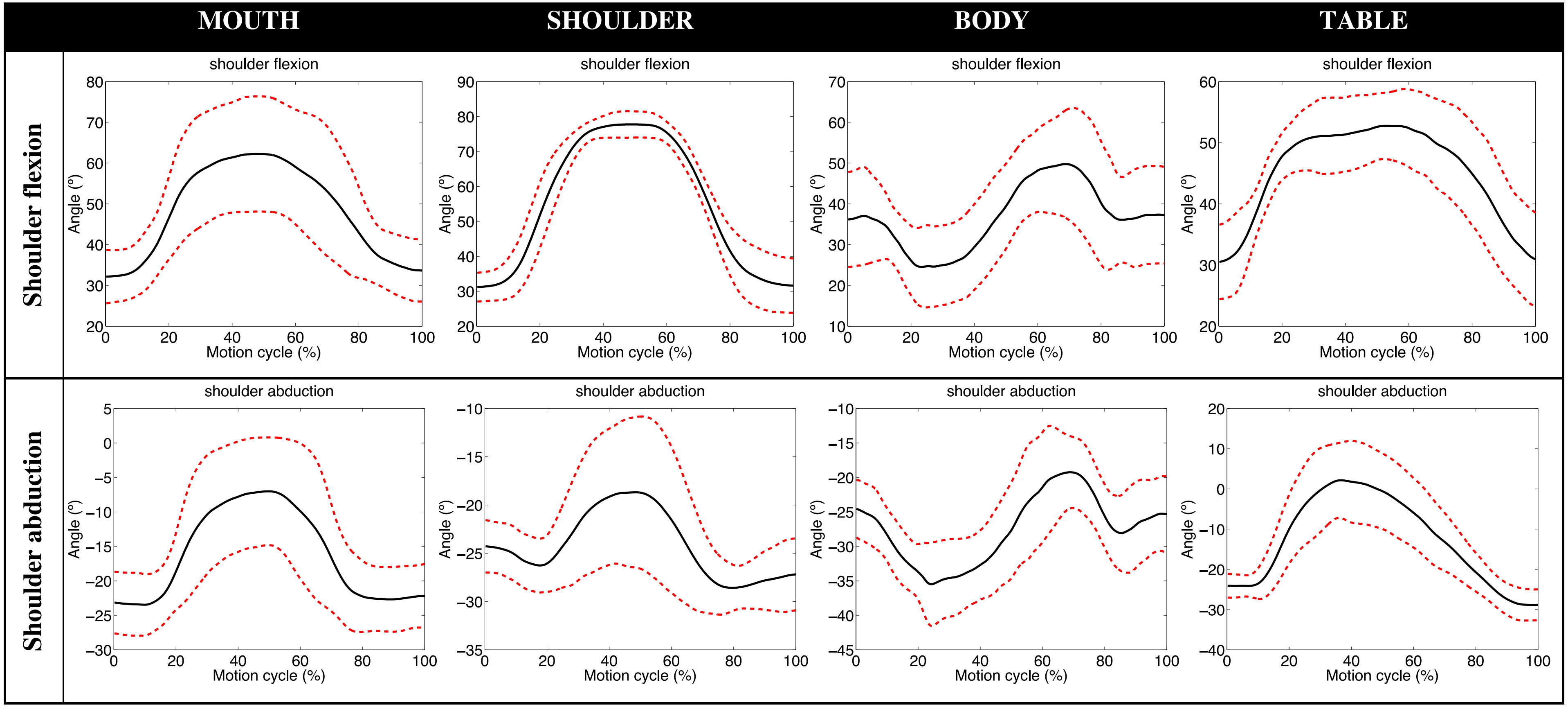
The mean of the joint angles during all the motion and the correspondent standard deviations is showed in figure 10. It can be seen that the patterns of the motions are mostly symmetric and similar between subjects. Standard deviations are small and follow the same profile as the mean curve.

The range of motion was analysed in two ways, first comparing the overall ROM between motions (Figure 11) and finally, computing the total (Figure 12), including all participants and motions. From all these data, the required minimum and maximum angles were found and are showed in Figure 13.

The range of motion is not exactly the same for all subjects due to the anatomic differences and variations on the initial position. When different motions are compared, there are huge differences on the ROM between them. It is reasonable because each motion occupies a specific place in the space; so, the angles needed to reach these positions are different.

For example, shoulder abduction is mainly needed in Table motion where is wanted to reach a far position on that axis. Similarly, shoulder rotation has a huge impact on Body and Table tasks, which are the two lateral motions, lower impact on Mouth task and almost no effect when going forward. Shoulder and elbow flexion are the opposite, they are more needed in Mouth and Shoulder motions than in the two others, where the rotations are lower but existent. Pro-supination is a weird case. Even though it seems to affect in all the motions, when visualize the motion over the cycle as in Figure 10 it can be observed that it changes depending only on the initial position. So, if the subject starts at 0, -10 degrees the angle remains more or less constant over all the motion. If not, it has to first reach this aimed angle.

Finally, when comparing all the joints, the one that has less impact on the motions studied is pro-supination followed by shoulder abduction.


MOUTH
SHOULDER
BODY
TABLE
Shoulder flexion
Shoulder abduction
shoulder flexion
shoulder abduction
Angle (°)
Motion cycle (%)

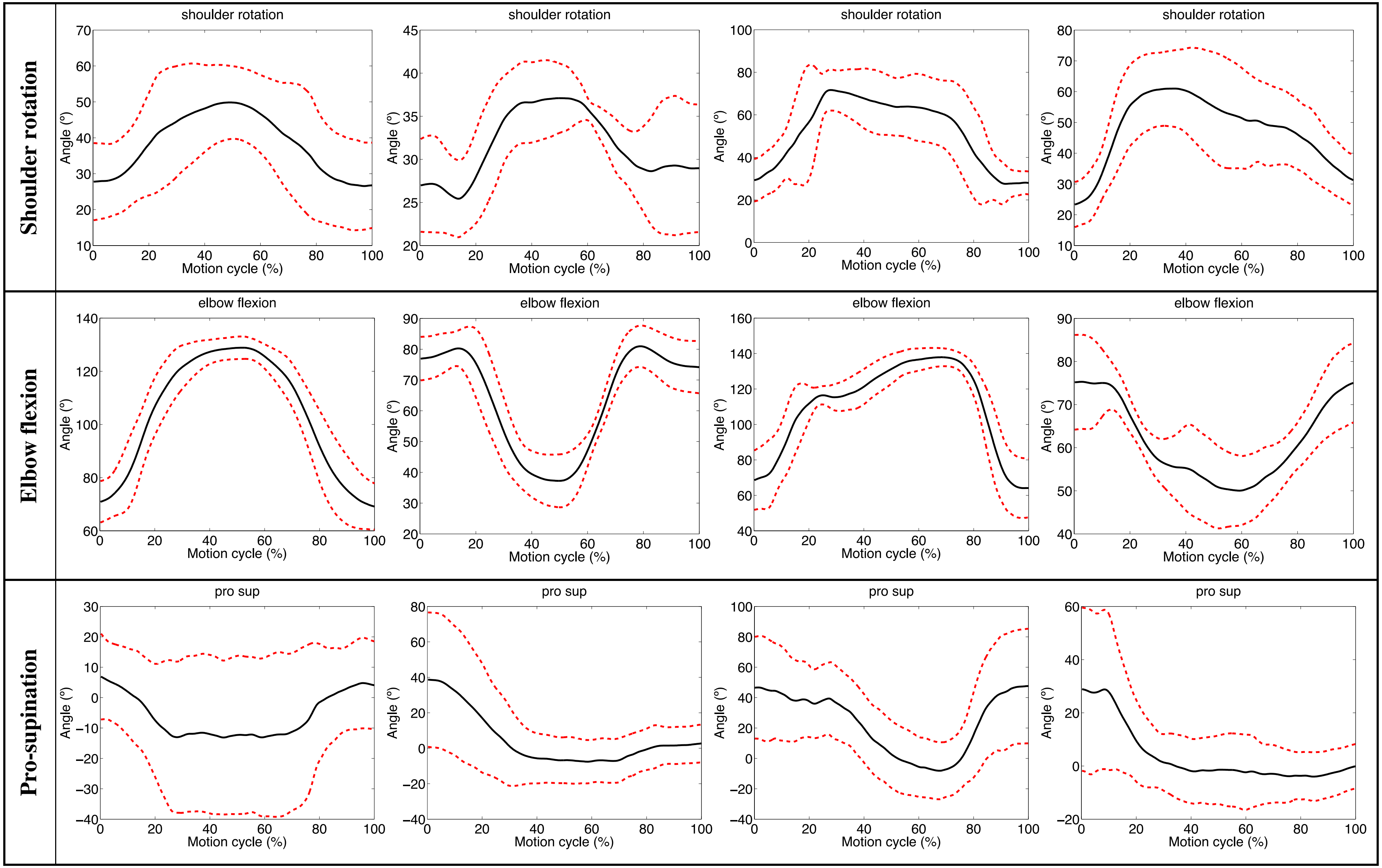
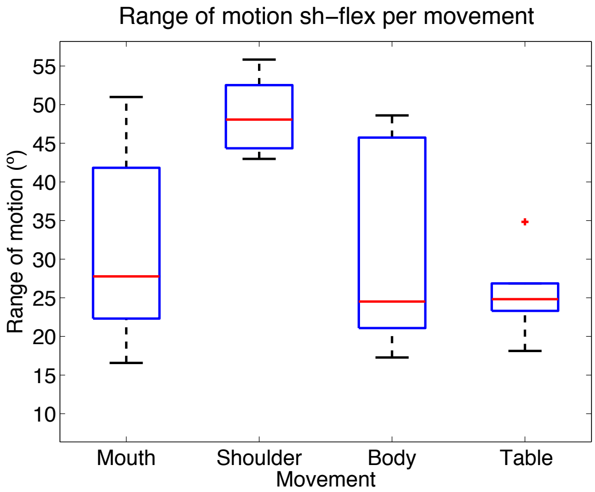
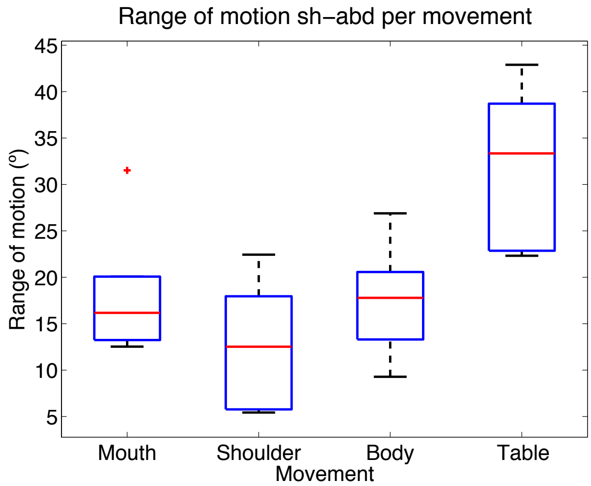


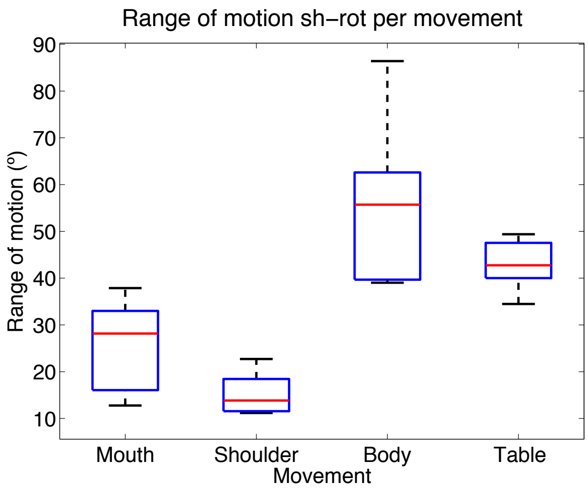
Figure 11: Mean joint angle per motion in black and the corresponding standard deviations in red.



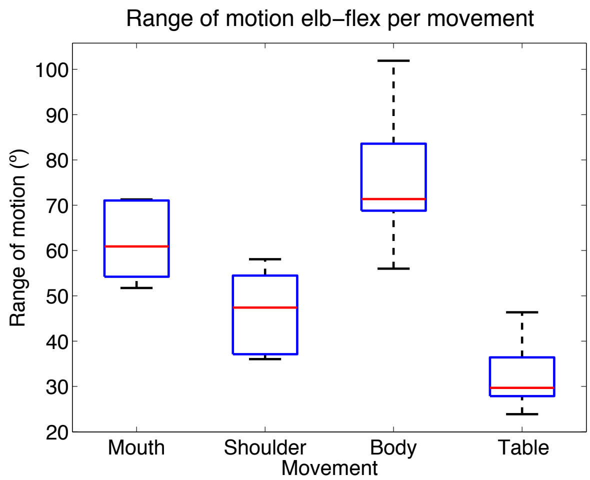
a)



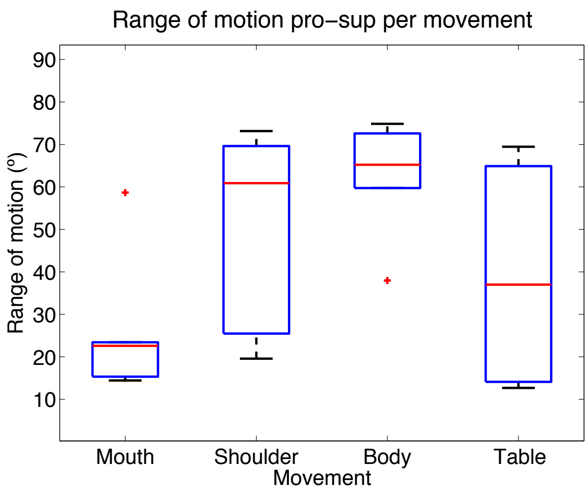
b)



c)



d)



e)

Figure 12: Range of motion per motion for all the joints angles. a) Shoulder flexion, b) Shoulder abduction, c) Shoulder rotation, d) Elbow flexion, e) Pro-supination.

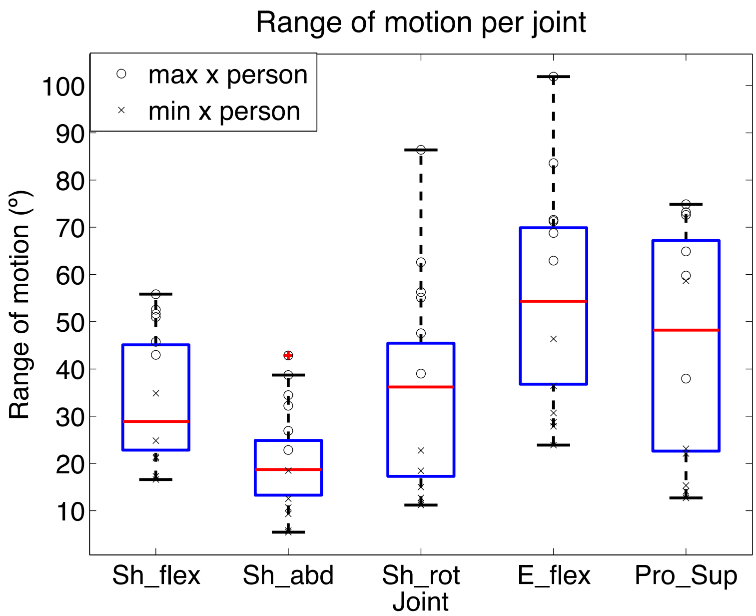
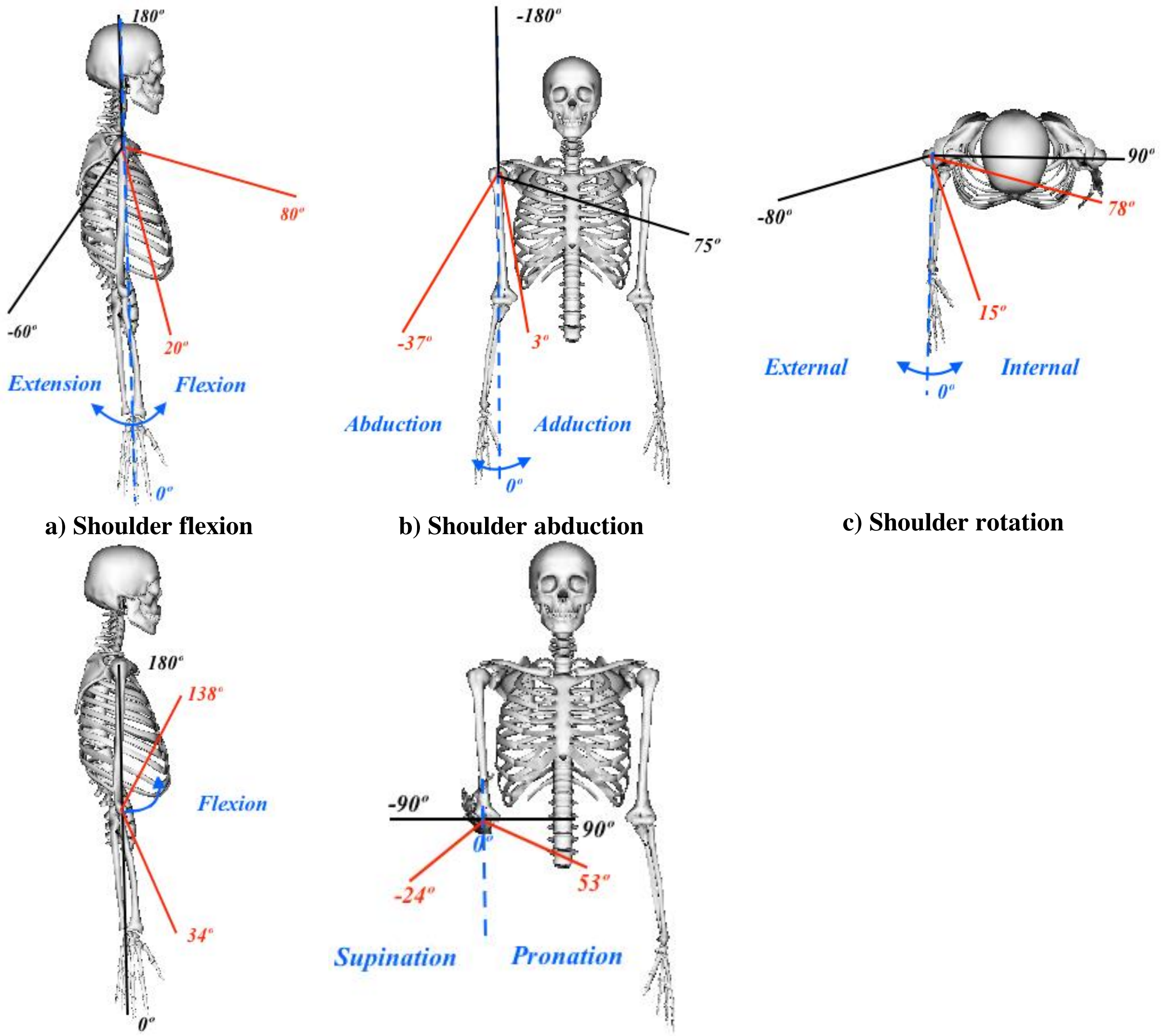


Figure 13: Total range of motion per joint. Includes all subjects and motions studied.



a) Shoulder flexion

b) Shoulder abduction

c) Shoulder rotation

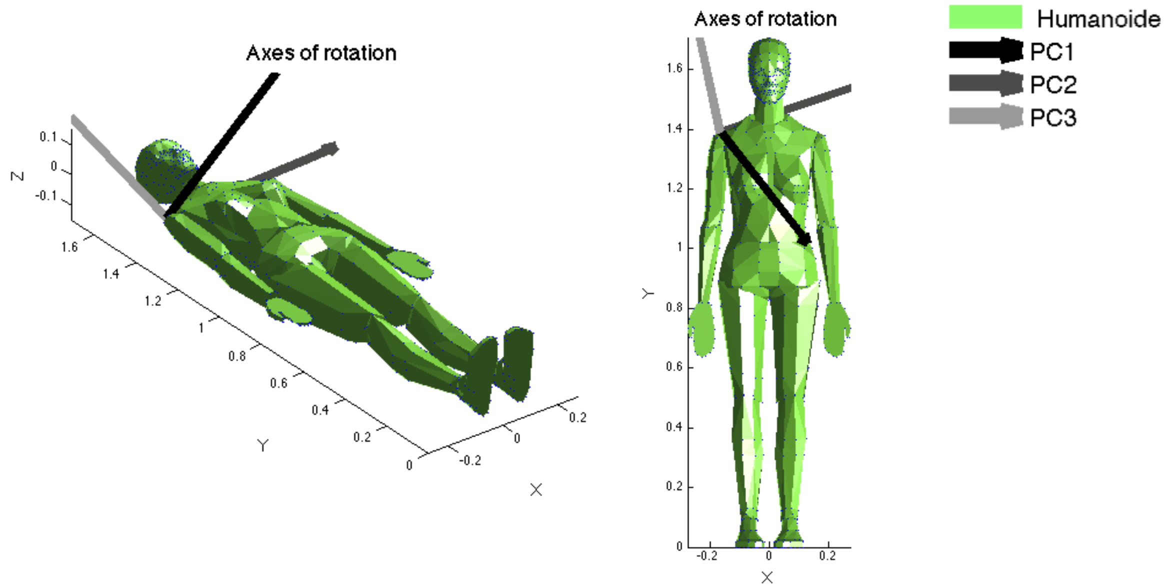
d) Elbow flexion

e) Pro-supination

Figure 14: Maximum and minimum angle required per joint.

4.1.2. Shoulder instantaneous axis of rotation

The new axes obtained after computing a PCA analysis on the IAR found per all motions and subjects are presented in Figure 14 and in Figure 15 there is the percentage of variation explained by each one. Two axes explain 93,26% of the total variability, which roughly reproduce the studied motions. The coefficients of the principal components obtained are defined in Table 5.



a) b)

Figure 15: Representation of all the instantaneous axis of rotation and the three new axes found through the PCA. a) 3D view, b) Frontal view.

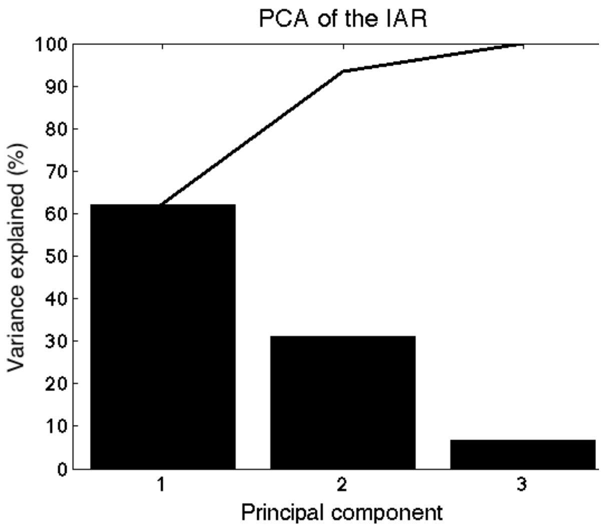


Figure 16: Variance explained per principal component

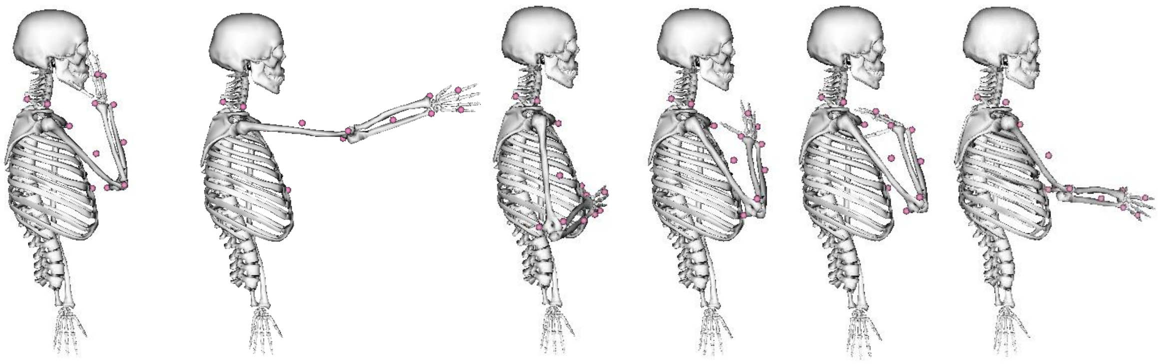
	PC1	PC2	PC3
X - abduction	0.32	0.93	-0.18
Y - rotation	-0.41	0.31	0.86
Z - flexion	0.85	-0.20	0.48

Table 5: Principal component coefficients

From the coefficients, first axis is mainly flexion with part of rotation, the second mainly abduction and finally, rotation combined with flexion.

As a first validation, these 3 new axes, found through the PCA analysis, were applied to the

OpenSim model substituting the old ones. The third axis, corresponding to the PC3, was blocked at 60 degrees. Then, from just two axes, kinematic were calculated again in order to test if just with these two axes was enough to perform the movements. Figure 16, shows the furthest frames aimed to reach in each task. The whole motions obtained can be seen in movies 1-4 of the project supplementary material (CD-ROM).



a) b) c) d) e) f)

Figure 16: Furthest positions per motion. a) Reach the mouth, b) Extend the arm at shoulder height, c),d),and e) Refer to body motion, c) Touch the belly, d) Attemp to touch the chest, e)Addition of wrist flexion to touch the chest and finally, f) Arrive to the other side of the table.

4.1.3. Angular velocity and acceleration

To conclude with the kinematics section, angular velocities and accelerations per joint were computed. Angular velocity gives an idea of how change the angle in function of time and angular velocity of which is the variability on the angular velocity over the time. The mean values with the correspondent standard deviation and maximums are summarized in Table 6.

	Angular velocity (rad/s)						Angular acceleration (rad/s^2)					
	Mean		Std		Max		Mean		Std		Max	
	+	|-|	+	|-|	+	|-|	+	|-|	+	|-|	+	|-|
S flex	0.25	0.23	0.31	0.28	1.40	1.28	1.03	0.97	1.29	1.08	7.30	5.21
S abd	0.15	0.15	0.22	0.18	0.98	0.70	0.67	0.66	0.91	0.82	4.78	4.11
S rot	0.28	0.23	0.40	0.27	1.71	1.30	1.40	1.39	1.77	1.82	10.07	11.90
E flex	0.37	0.41	0.47	0.52	1.97	2.25	1.49	1.52	2.12	2.04	10.76	9.50
Pro-sup	0.26	0.33	0.44	0.43	2.43	1.88	2.58	2.58	3.31	3.49	23.19*	22.62*

Table 6: Mean, standard deviation and maximum of angular velocity and acceleration. (*) High values partially produced due to the presence of noise in this particular joint.

4.2. Dynamics and energetics

Dynamics gives information about the force needed to perform the motions. For a design the most useful parameters are joint torques and powers. The results obtained of this analysis are presented in the following subsections.

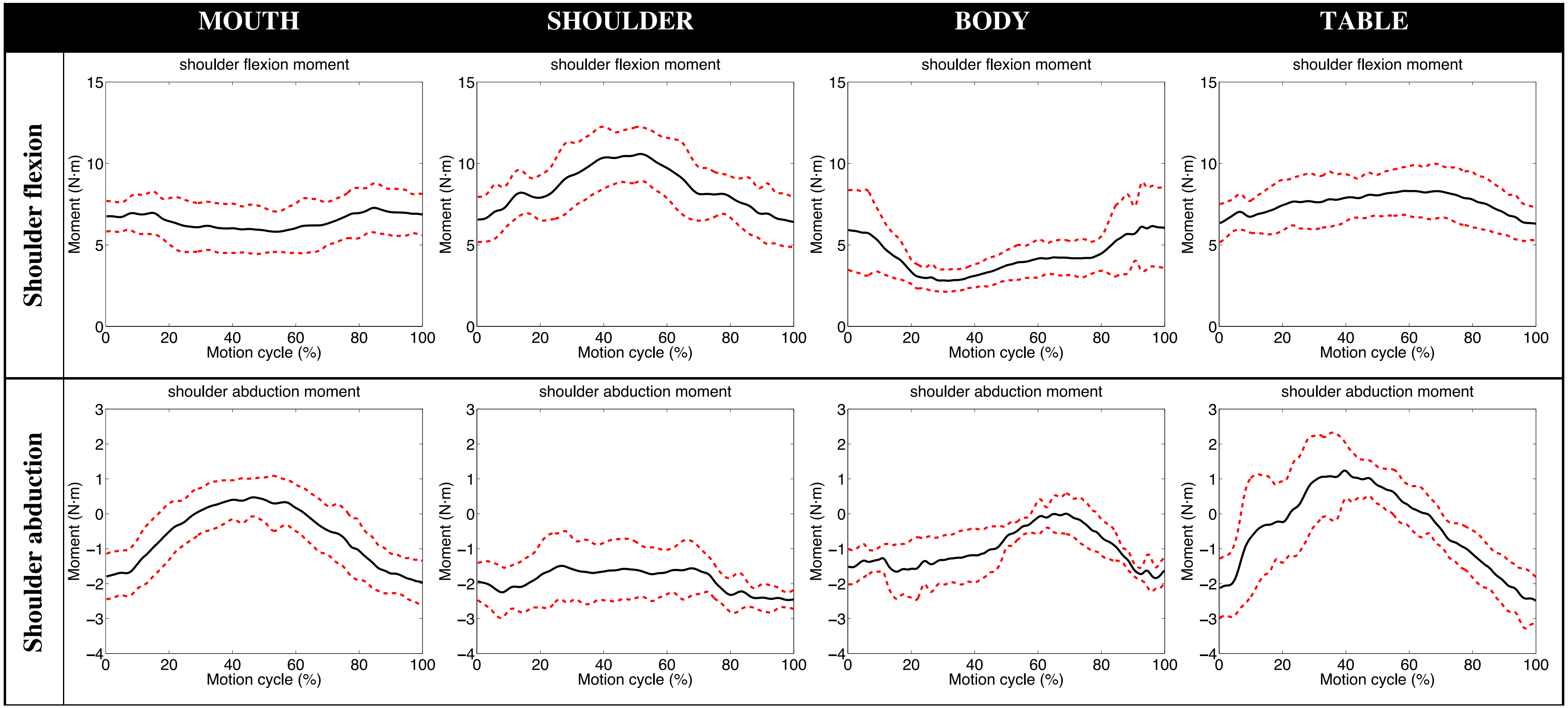
4.2.1. Torques

Following the same idea of the kinematics part, first an exploratory study is done and then the maximum torques are extracted. All the results are shown in Figures 17-19. All the motions include an object with the exception of Body task.

What it is interesting when analysing the torques is the maximum forward torque and the maximum backward one. It was compared between motions and finally, the overall was calculated.

In the results it can be observed that the values obtained are similar for all the subjects but highly different between from one rotation to another. It becomes important on some of them like shoulder flexion, where the highest torque is generated. Contrary, in pro-supination rotation, torque is almost zero for all the cases. Another curiosity is that torques needed for forward and backward motions are practically the same, since the motion is symmetric.

Additionally, gravitational and inertial torque were computed and contrasted. In Figure 20 the mean of all the subjects of total, inertial and gravitational torque is presented. In all the cases the cases gravity torque has a high influence on the total torque whereas the inertial is considerably low, closer to zero in most of the situations.


MOUTH
SHOULDER
BODY
TABLE
Shoulder flexion
Shoulder abduction
shoulder flexion moment
shoulder abduction moment
Moment (N·m)
Motion cycle (%)

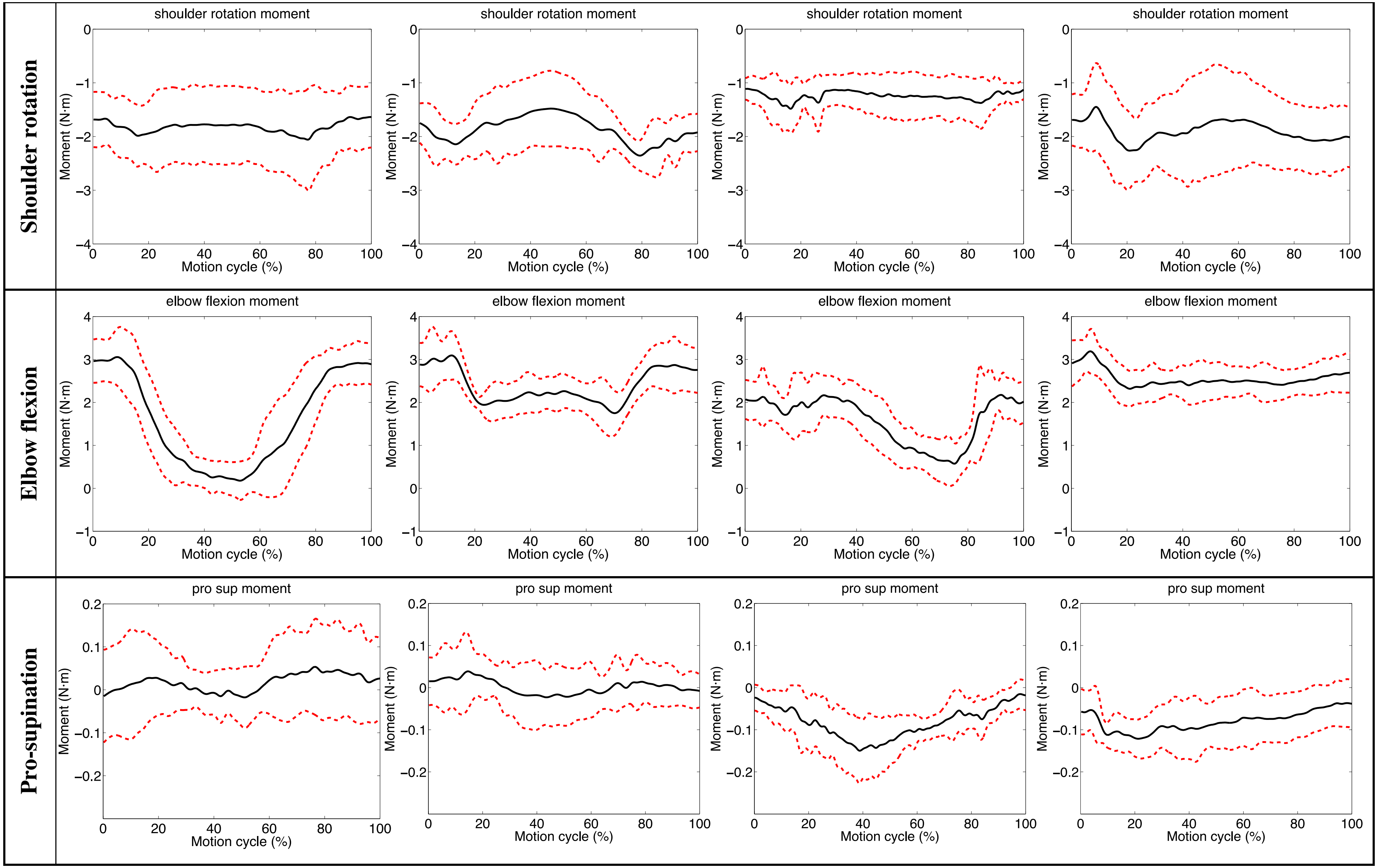


Figure 17: Mean total torque per motion in black and the corresponding standard deviations in red.

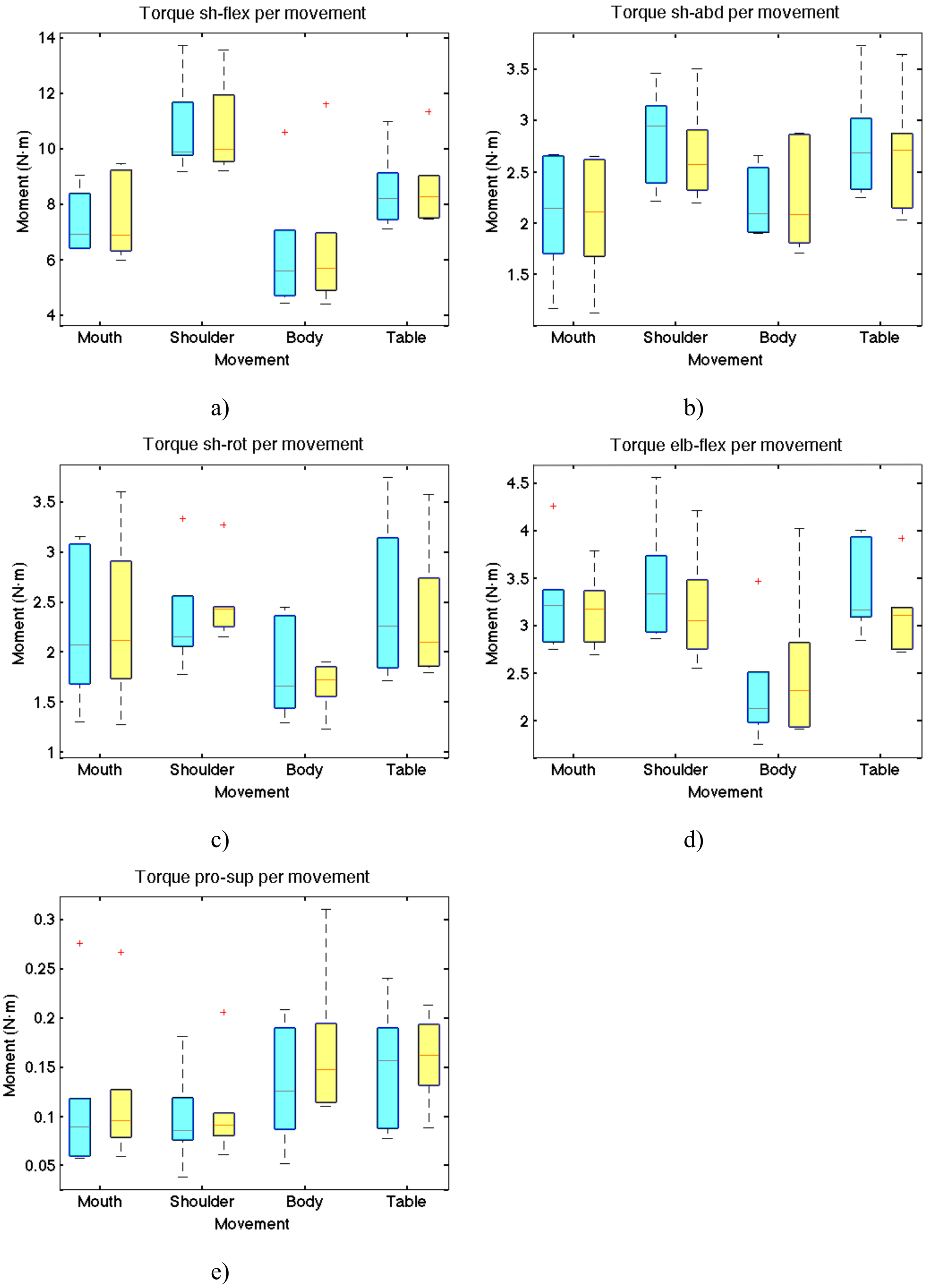


Figure 18: Torque per motion for all the joints angles. In blue: maximum forward torque and in yellow: maximum backward torque. It is calculated in absolute values. Includes all subjects and motions studied. a) Shoulder flexion, b) Shoulder abduction, c) Shoulder rotation, d) Elbow flexion, e) Pro-supination

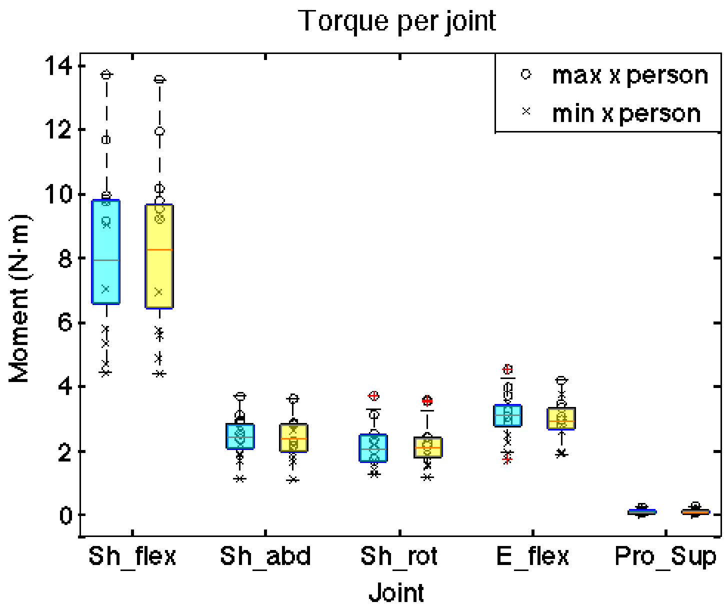
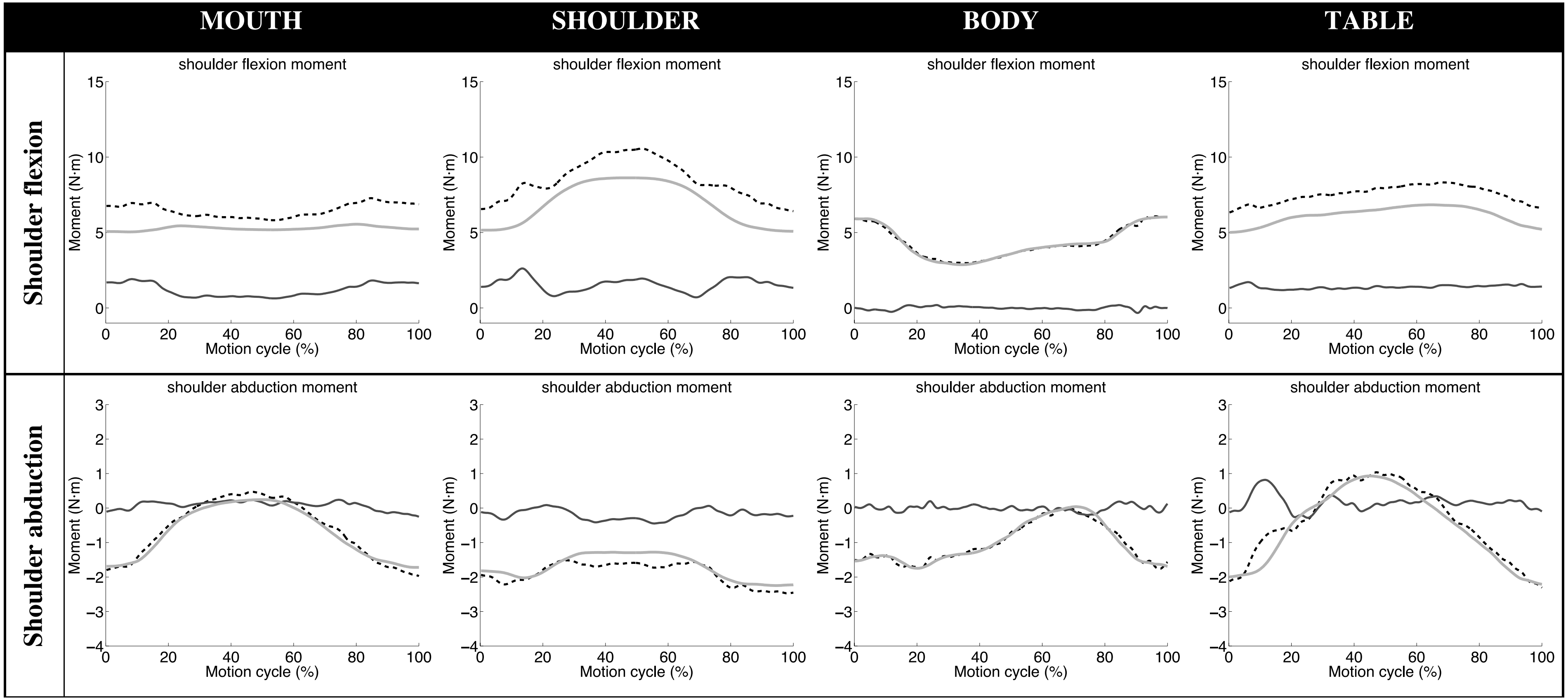


Figure 19: Total torque per joint. In blue: maximum forward torque and in yellow: maximum backward torque. It is calculated in absolute values. Includes all subjects and motions studied. The circle is the maximum per person among the maximums and the cross the minimum per person among the maximums.


MOUTH
SHOULDER
BODY
TABLE
Shoulder flexion
Shoulder abduction
shoulder flexion moment
shoulder abduction moment
Moment (N·m)
Motion cycle (%)

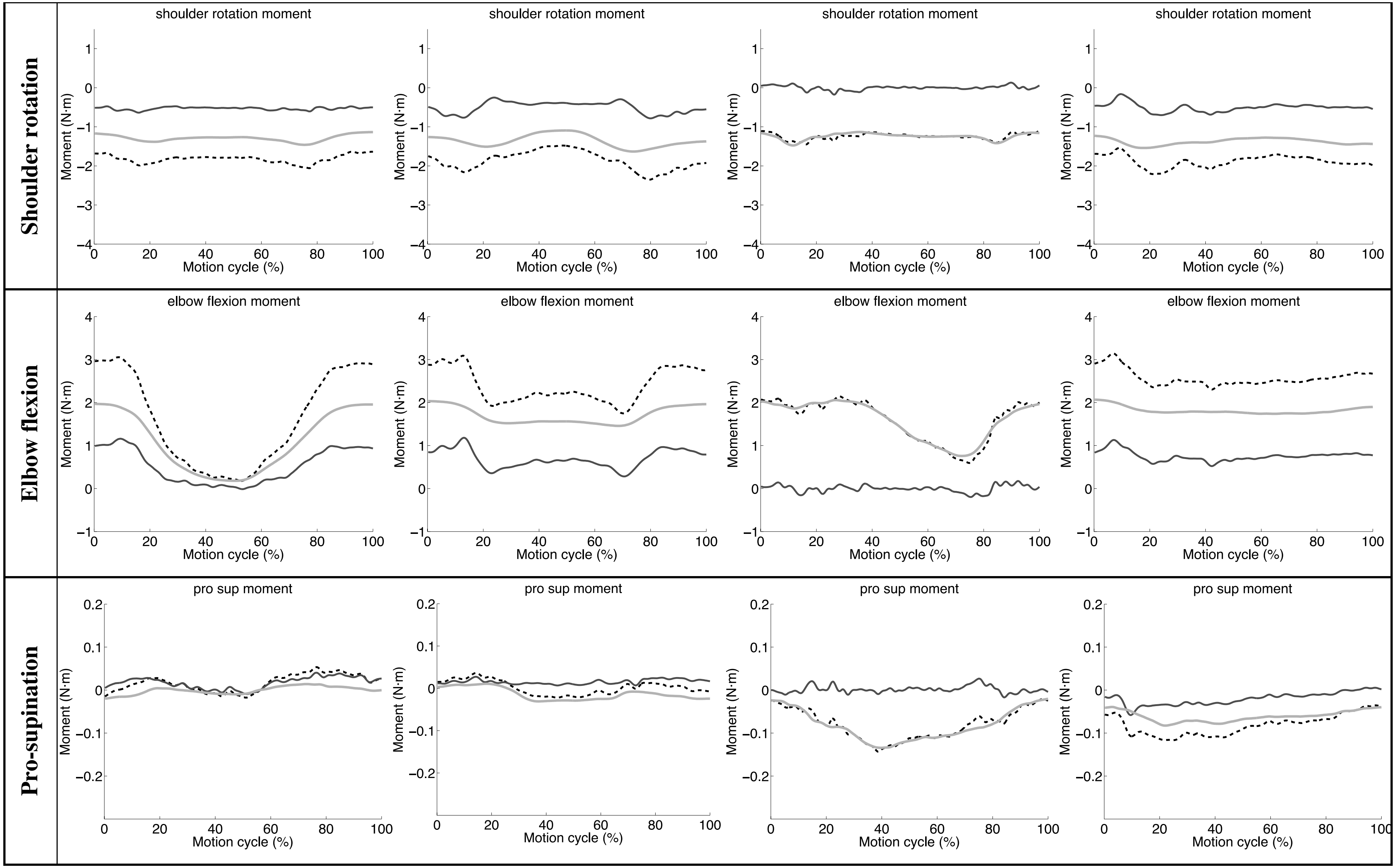


Figure 20: Mean torque per motion is presented. In dashed black total torque, in dark grey inertial torque and in light grey gravitational torque.

4.2.2. Mechanical joint powers

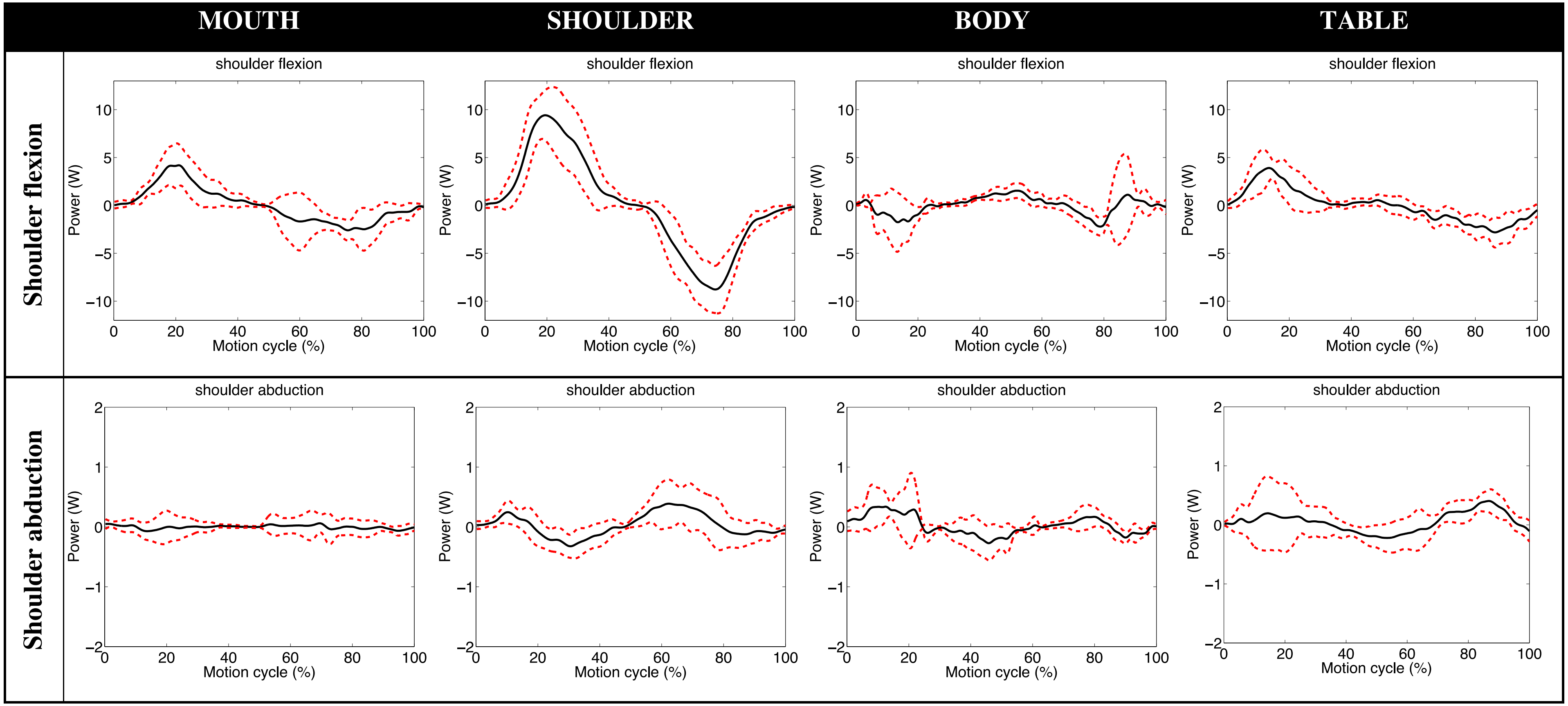
Again, there is first an exploratory part where the mean of the powers of the subjects during the motions together with the standard deviation is computed and plotted. This part is followed by an analytical part where the maximum forward and backward power need per motion and the total is calculated. All these results are presented in Figures 21 to 23.

From all the graphics obtained is observed that the power needed per joint is similar for all the activities but there are some exceptions like shoulder flexion when doing Shoulder task. The power is higher than in the other motions.

It can be observed that the value of power differs considerably between joints. There are joints where the power used is close to zero during the entire task like pro-supination and shoulder abduction. While in others as shoulder and elbow flexion is quite high.

Power and torques are somehow related, as power is calculated from torques. For the same velocity and higher torque, also power is higher. Additionally, power takes into account the velocity of the motion, obtaining more power when the motion is performed faster.

Moreover, gravity and inertial powers were computed and contrasted. In Figure 24 the mean of all the subjects of total, inertial and gravity power is presented. In all the cases the cases gravitational power has a high influence on the total torque whereas the inertial is significantly low.


MOUTH
SHOULDER
BODY
TABLE
Shoulder flexion
Shoulder abduction
shoulder flexion
shoulder abduction
Power (W)
Motion cycle (%)

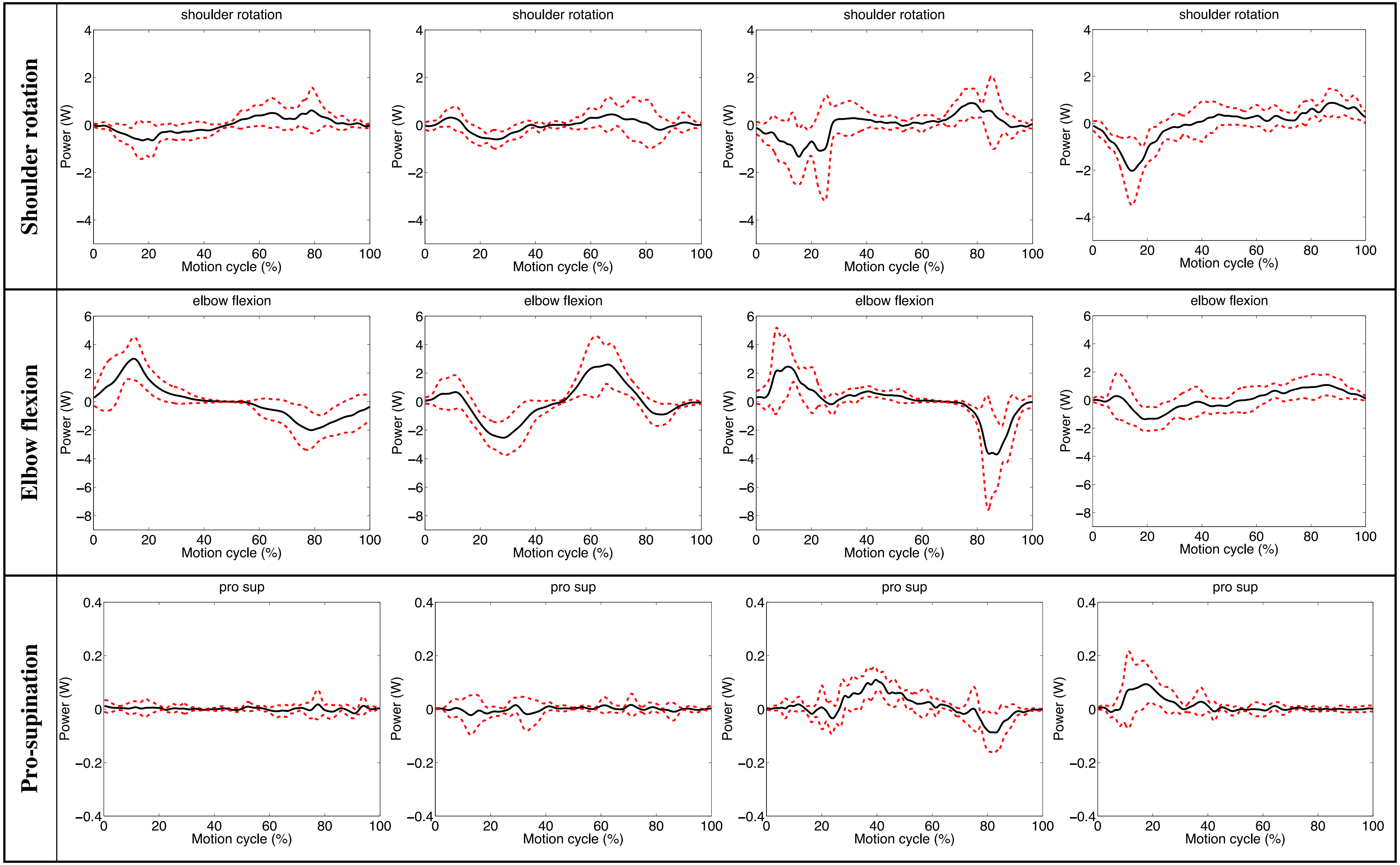
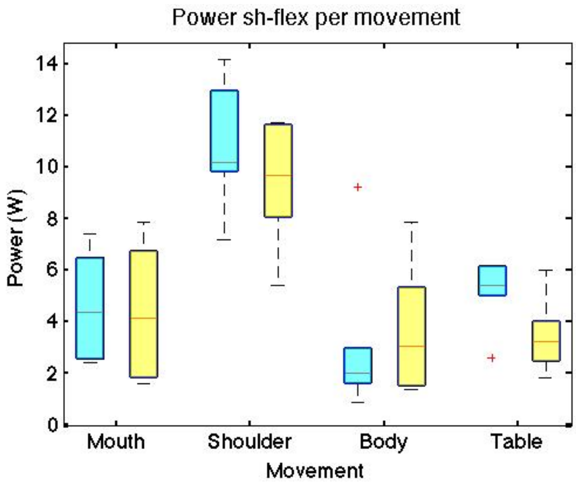
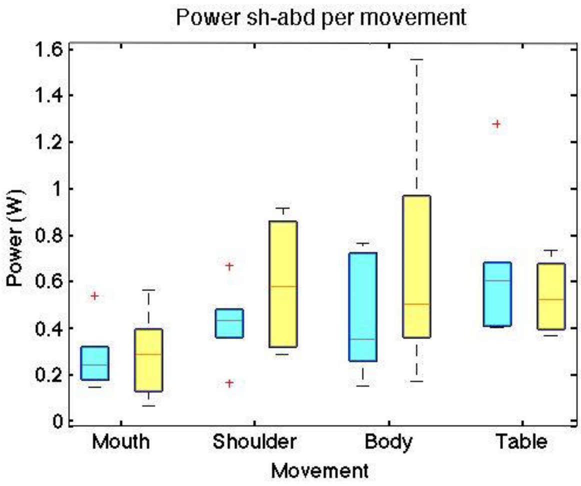


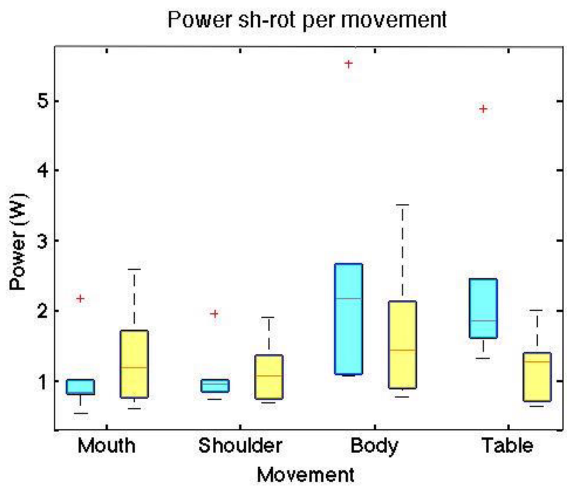
Figure 21: Mean power per motion in black and the corresponding standard deviations in red.



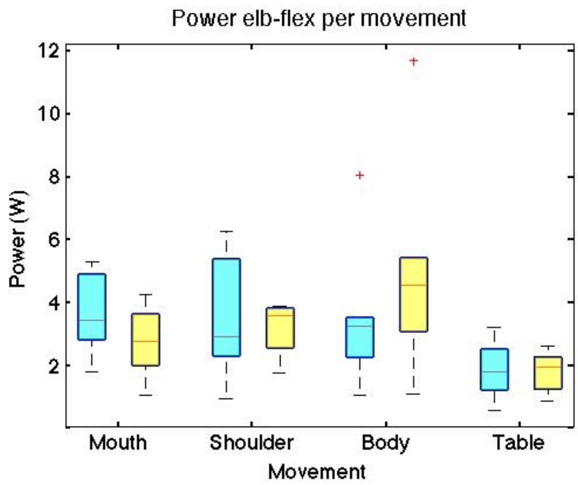
a)



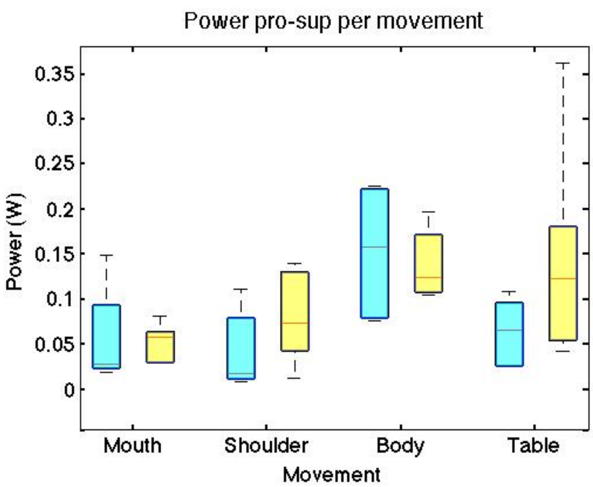
b)



c)



d)



e)

Figure 22: Power per motion for all the joints angles. In blue: maximum forward power and in yellow: maximum backward power. It is calculated in absolute values. a) Shoulder flexion, b) Shoulder abduction, c) Shoulder rotation, d) Elbow flexion, e) Pro-supination.

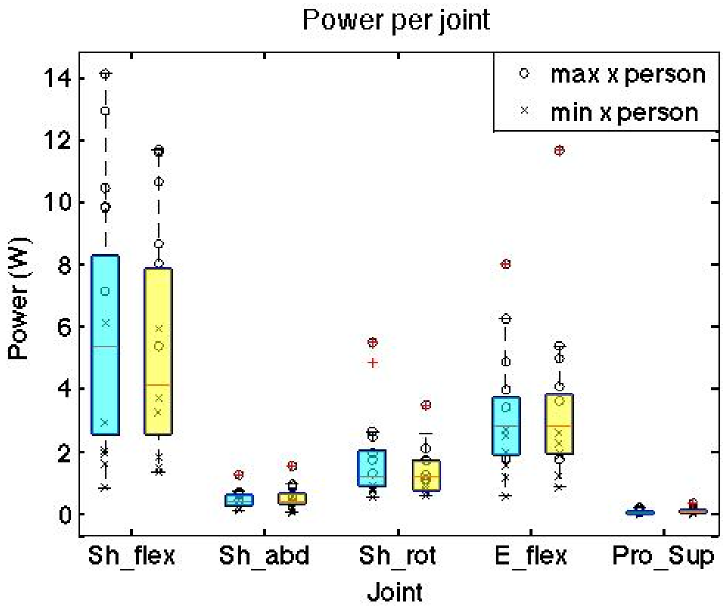
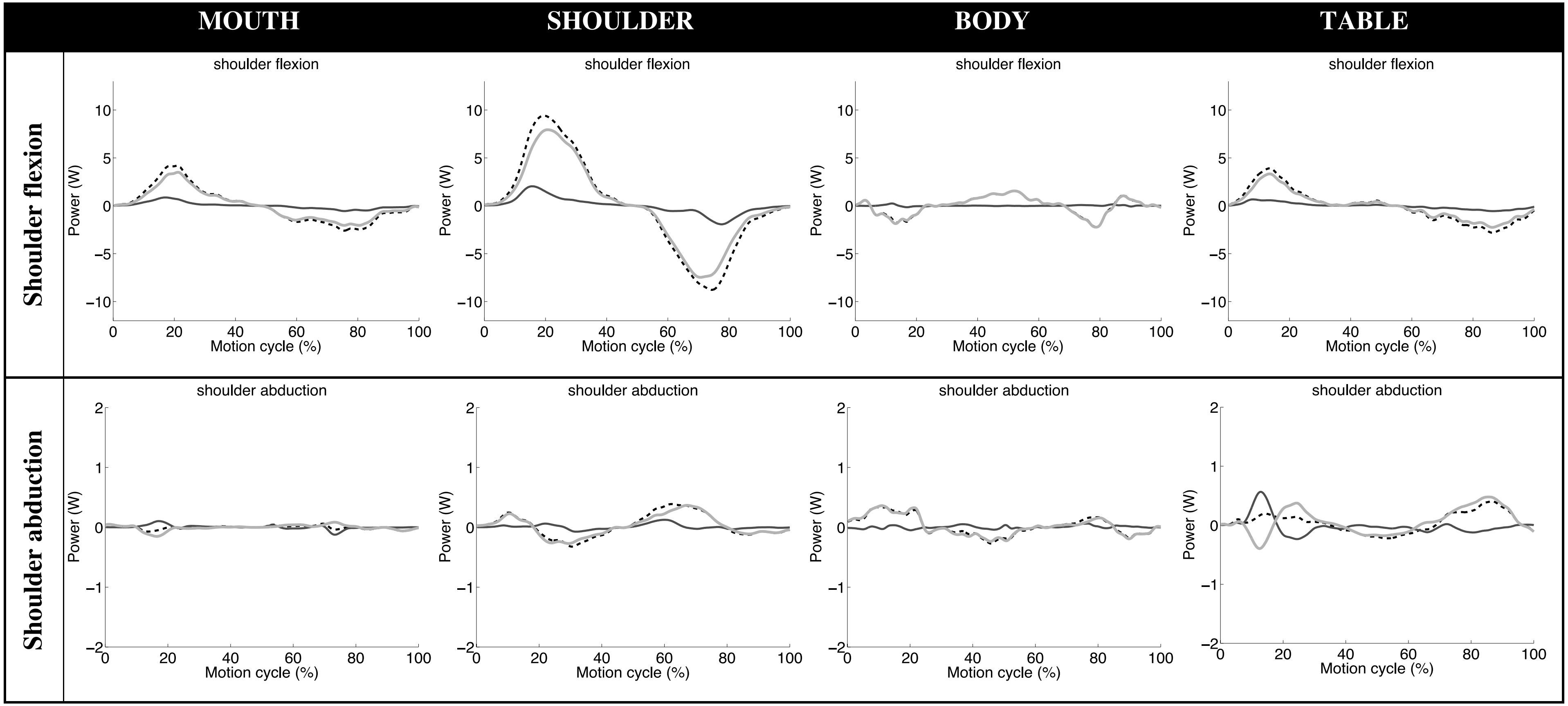


Figure 23: Total power per joint. In blue: maximum forward power and in yellow: maximum backward power. It is calculated in absolute values. Includes all subjects and motions studied. The circle is the maximum per person among the maximums and the cross the minimum per person among the maximums.


MOUTH
SHOULDER
BODY
TABLE
Shoulder flexion
Shoulder abduction
shoulder flexion
shoulder abduction
Power (W)
Motion cycle (%)

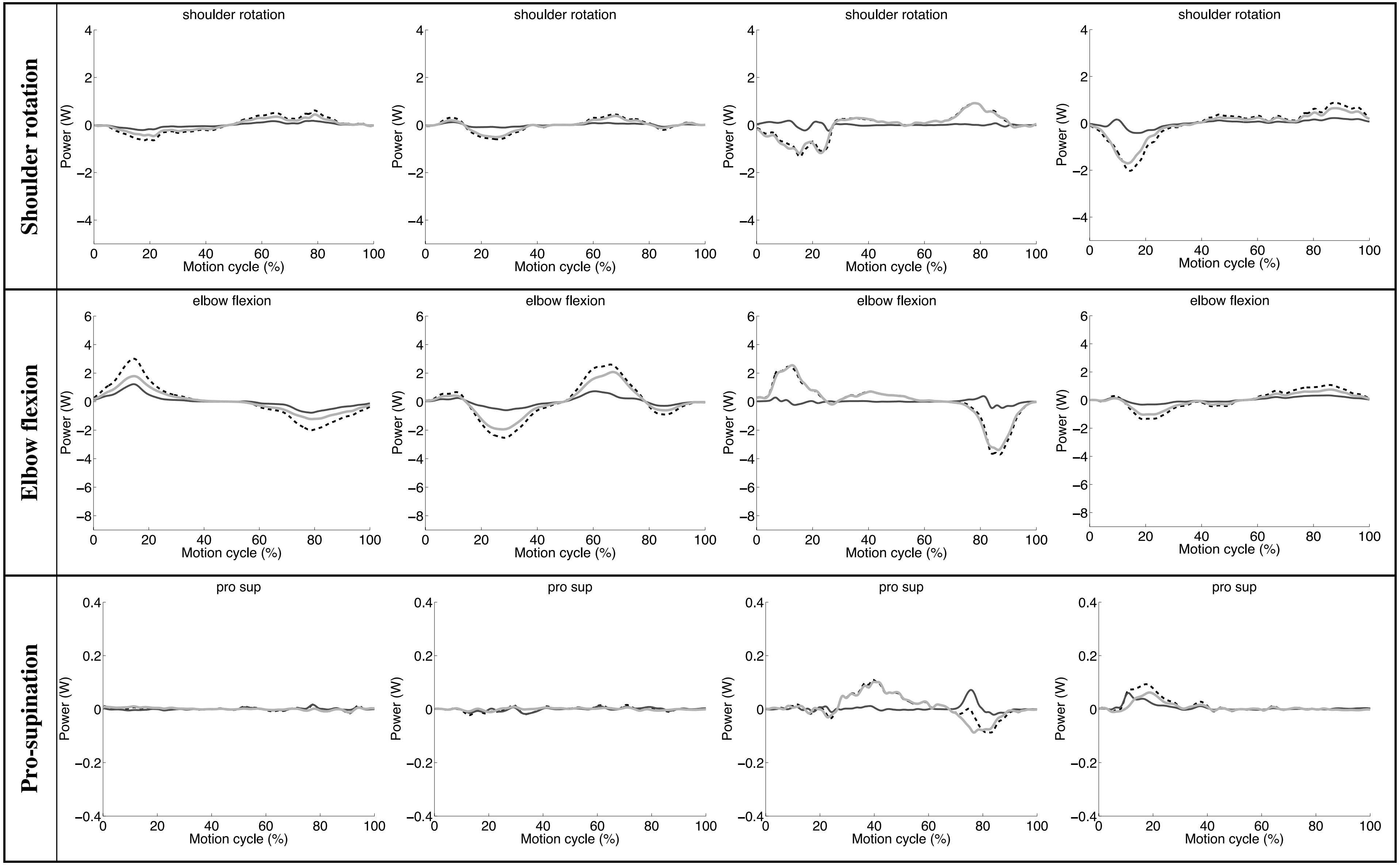


Figure 24: Mean power per motion is presented. In dashed black total power, in dark grey inertial power and in light grey gravitational power.

5. Discussion

The objective of this project was to carry out a quantitative analysis of the biomechanical requirements for the design of an active arm support for people with Duchenne, assessing how the model can be simplified in order to fulfil aesthetic and functional criteria at the same time. The basic functionality is understood as assist on the desired daily activities for these patients: eat, drink, reach the phone, and scratch the body, among others. The question will be answered along this section where the different parameters obtained through studying the kinematics, dynamics and energetics of 6 different subjects while performing 4 pre-determined tasks are defined and discussed.

It seems obvious that before designing an orthosis some biomechanical studies have to be done in order to define the parameters of the model. However, when looking into the literature, just one project analyses the dynamics of the motions and four cover some kinematics. Indeed, just one is focused on Duchenne Muscular Dystrophy [19-22]. So, this project provides a new set of guidelines to take into account when designing arm exoskeletons.

Before extracting any conclusions, it is necessary to validate the data and see if it exists similarity between subjects. Figures 10, 17 and 21 show this resemblance, proven by low standard deviation and same curve shape, in terms of joint angles, torques and power. So, the data captured is useful and the desired parameters of Range of motion, joint angles, angular velocities and accelerations, torques and power needed for the orthosis can be obtained. Mean, maximum value and standard deviation for different joint variables are summarized in Table 7.

When contrasting the kinematical results obtained in this study with previous reported studies [19-22], the angles obtained when analysing the same tasks as in this project are similar. Angular velocities and accelerations are in the same order of magnitude as in Rosen et al. [22]. Dynamics results are not possible to compare due to the difference on the objects used to do the tasks and the different inertial properties of the subjects.

In most of the cases, the mean cannot be used as a direct design value because is not sufficient to perform all the tasks properly. Instead of the mean, the maximum values obtained are the quantities that will be of interest for the requirements, if these values are not reached, issues on the motion can occur.

	Angle (º)				Range of motion		Inertial torque (N·m)				Gravitational torque (N·m)				Total torque (N·m)			
	Min		Max		Req.		Mean	Std	Max		Mean	Std	Max		Mean	Std	Max	
	Mean	Std	Mean	Std	Mean	Std			Mean	Std			Mean	Std			Mean	Std
S flex	21.70	8.55	79.47	5.19	49.95	4.72	0.98	0.72	3.02	0.71	5.73	1.86	8.66	1.60	6.67	2.24	10.77	1.71
S abd	-36.54	5.00	3.15	9.49	33.00	7.39	0.20	0.22	1.59	0.62	1.21	0.82	2.70	0.61	1.26	0.87	2.94	0.47
S rot	15.05	8.70	78.23	7.43	57.83	16.18	0.40	0.28	1.07	0.21	1.39	0.46	1.88	0.50	1.76	0.63	2.82	0.66
E flex	34.15	3.98	138.76	4.95	76.67	14.07	0.48	0.38	1.57	0.34	1.58	0.66	2.16	0.54	2.02	0.90	3.60	0.62
Pro sup	-24.48	20.97	53.48	29.44	63.87	13.95	0.02	0.02	0.10	0.03	0.06	0.05	0.15	0.07	0.08	0.05	0.18	0.07

	Gravitational power (W)				Inertial power (W)				Total power (W)			
	Mean	Std	Max		Mean	Std	Max		Mean	Std	Max	
			Mean	Std			Mean	Std			Mean	Std
S flex	1.41	1.94	9.85	2.20	0.24	0.42	2.76	1.24	1.63	2.23	11.16	2.40
S abd	0.14	0.19	1.15	0.56	0.04	0.11	0.82	0.61	0.14	0.18	0.96	0.41
S rot	0.35	0.49	2.56	1.64	0.09	0.14	0.85	0.51	0.42	0.54	2.64	1.49
E flex	0.63	0.90	4.34	2.51	0.19	0.32	1.93	1.06	0.80	1.10	5.67	3.30
Pro sup	0.02	0.03	0.18	0.05	0.01	0.02	0.16	0.14	0.02	0.03	0.21	0.09

Table 7: Minimum, maximum, mean, standard deviation obtained for angles, range of motion, torque and power.

Focusing on the kinematics part, from Figures 10-13 it can be observed that the variability depends mainly on the initial position. For all the motions, during the main part of the movement, pro-supination angle is around 0º. So, it can be assumed that setting an adequate initial position, around these 0º used to perform the motion, will be translated into a fix pro-supination angle. Then, it is possible to say that pro-supination angle is not needed for these tasks and therefore does not need to be actuated.

The kinematic results make clear that there are three basic joints, which are elbow flexion, shoulder flexion and shoulder rotation, without them the motions could not be executed. In the case of the shoulder, from computing a principal component analysis on the instantaneous axis of rotation it is possible to obtain two more general axes that cover almost all the arm trajectories. As it is shown in Figure 15, with these two new axes it is possible to reproduce 93.26% of the IAR variability during all the tasks.

From the validation of the new axes, done through applying them to the OpenSim model and computing the motions again, it was found that all the motions could be performed with the exception of reaching the chest, in Body task. It is possible to reach the belly and when adding wrist flexion it is also possible to reach the chest. In Figure 16, it is shown how the furthest positions aimed are achieved just with these two axes and, in movies 1-4 of the project supplementary material (CD-ROM) is possible to see the motions. From this point, to continue it would be necessary to find the new range of motions, torques and powers.

About angular velocities and accelerations obtained, it is surprising that pro-supination is the one with highest angular velocity and acceleration. This phenomenon is related with the fact that pro-supination joint is one of the noisiest ones and it rotates really fast in comparison with the others when doing a normal task. However, it only rotates when the initial position is different than 0º. The other joints with higher values are shoulder rotation and elbow flexion. Positive and negative values of velocities and accelerations are quite similar, representing again this symmetry on the motions [see Table 6].

Finalizing with the kinematics part, a reasonable simplification of the model would be to block pro-supination joint and to transform shoulder flexion, abduction and rotation axes into the new ones obtained from the IAR. Applying these adjustments will lead to a simplified model with 3 degrees of freedom instead of the 5 that has the general one. Therefore, the number of actuators could be reduced from 5 to 3.

This simplification can be implemented theoretically on the models but its physical realization is out of the scope of this project. Due to this fact, the new axes found may need to be rearranged when building the model in order to fulfil the design criteria.

Jumping to dynamics, the results enlighten the idea that adding a gravity compensator on the model reduces the torques and powers needed per joint (Figure 20, 24 and Table 7). A structure that compensates the gravity effect is nothing weird. It is what it has been used on passive exoskeletons to provide assistance to these patients that still have some strength on their muscles [7]. Consequently, a way to reduce the size of the orthosis by simplifying the size of the actuators would be to combine a passive and an active exoskeleton. However, including a mechanism to compensate the weight implies more complexity and thus weight.

When going deeper analysing the values obtained, the highest torques and motions are found in the same joints were the highest range of motion was needed. That fact reinforces the theory that pro-supination actuation can be eliminated, torques are around 0 N·m and powers also around 0 W. On the shoulder, flexion is the one with highest power, followed by rotation and abduction.

Power values are highly influenced by angular velocity, which at the same time varies in function of the velocity used to perform the task. Higher velocities are translated into higher powers. Not all the subjects used the same velocity, in such a way that the results have certain variability. As the means are the ones extracted, the values obtained are the necessary to do the motions in an average speed.

Regarding the limitations of the model, it is easily visible that the main one is the small sample of subjects studied. Although only 6 subjects were analysed it is enough in order to have a first idea of the biomechanical requirements. Even more, given the similarity on the performance among them. More subjects could be used in further studies to help in a more accurate quantification of the optimum values.

Other aspect about the sample is that all the subjects were healthy. This feature is explained through the fact that Duchene patients do the movements in a compensatory way and each patient does it differently, which may lead to strange patterns. For this study, it was aimed to have a healthy model and then apply it to these patients.

Another limitation would be the fact that only four tasks were performed. The four main tasks that at the same time represent a set of subtasks were studied. It was done in this way to simplify the analysis and because it is enough to have a first overview. However, analysing task by task individually would lead to a more precise study.

Further studies would require to increment the number of participants, analyse each desired task individually and finally, go deeper in the analysis testing what happens when simplifying the model. A point to discuss would be if it necessary to analyse Duchene patients motions. That would depend on the idea of the exoskeleton, if it is aimed to perform the movements on a natural way or to focus on the way that these patients do the tasks. Studying the way they compensate their shortcomings or in which rotations have more disability and assist them according to that.

Summarizing, it was possible to find the basic biomechanical requirements for the design of an actuated exoskeleton for Duchenne patients, which means that the suggested goal was achieved. From now on main degrees of freedom, axes orientation and actuators features are known, obtained through kinematical and dynamical analysis of different daily living tasks.

The study proves that the arm support can be simplified in such a way that ease the aesthetics and inconspicuousness of the design, two of the desired features of Flextension project, and be still functional. The biomechanical requirements are just a part of the total requirements needed to consider when designing an exoskeleton. Other requirements include comfort, easy donning and doffing, force transmission to the body, adjustability to the body, functionality, aesthetics, inconspicuousness, etc. [23,24].

Finally, as Duchene is a muscular dystrophy, the majority of affections between the patients that suffer other arm dystrophies are similar, meaning that the exoskeleton or the biomechanical requirements defined in this project could be extended to other muscular dystrophy diseases.

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Projecte Final d'Estudis MÀSTER EN ENGINYERIA BIOMÈDICA

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Dr. President

Dr. Vocal 1

Dr. Vocal 2

Dr. Secretari

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