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# **simMACT, a software demonstrator to improve maximum actuation joint torques simulation for ergonomics assessment.**

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## **ABSTRACT**

*The maximum actuation joint torques that operators can perform at the workplace are essential parameters for biomechanical risk assessment. However, workstation designers generally only have at their disposal the imprecise and sparse estimates of these quantities provided with digital manikin (DHM) software. For instance, such tools consider only static postures and ignore important specificities of the human musculoskeletal system such as inter-joints couplings. To alleviate the weaknesses of existing approaches implemented in digital human modelling tools relying on torque databases, this paper describes a methodology based on a class of polytopes called zonotopes and musculoskeletal simulation to assess maximum actuation torques. It has two main advantages, the ability to estimate maximum joint torques for any posture and taking into account musculoskeletal specificities unlike existing digital human modeling tools. As a case study, it also compares simulated maximum actuation torques to those recorded during an experiment described in the literature, focusing on an isometric task of the upper limb. This simulation has led to similar or smaller errors than DHM software tools. Hence, this methodology may help in interpreting inter-joint couplings, choosing appropriate mathematical models or design*

*experimental protocols. It may also be implemented in DHM software to provide designers with more comprehensive and more reliable data.*

## **1 INTRODUCTION**

Many workstations in industry require operators to perform repetitive and strenuous activities. Such high or repetitive biomechanical demands may lead to discomfort, muscle fatigue and eventually pathologies, for instance work-related musculoskeletal disorders (MSD) [1, 2]. To prevent such occupational risks, the *design-integrated risk prevention* approach sets the workstation designer the objective of obtaining the lowest possible level of residual risk for the future operator. In Europe this approach is formalised by the “Machinery Directive”, which applies to all work equipments used in Europe [3].

Maximum actuation joint torques are key data for carrying out this mandatory assessment of biomechanical risk factors. A large amount of such experimental data has been gathered and published over the years [4, 5, 6]. However, these data depend heavily on both the subjects (age, gender) and the task to be performed (prescribed body posture, duration and frequency). Furthermore, the reference values are sparse (recorded for only a few joint angle values) and mainly measured in very specific conditions (isometric or isokinetic exertions) far removed from occupational demands. Since accessing these data and interpreting them is beyond the common skills and duties of work equipment designers, they usually rely on only one source for such information: databases of admissible exertion torques provided by Digital Human Models (DHM) software available in their usual Computer-Aided Design (CAD) tools. For instance, commercial DHM software such as Delmia Human® (Dassault Systèmes), Tecnomatix Jack / Process simulate® (Siemens) or 3DSSPP® (University of Michigan) provide abacuses with maximum reference actuation joint torque values for the upper limb and the trunk.

However, this software usually makes sweeping assumptions about movements (quasi-static approximation) and completely ignores the specificities of the human locomotor system such as the relation between joint torques and joint angles or joint velocities, as well as intra- and inter-joint actuation couplings. Thus maximum exerable joint torques available for workstation designers

may be highly approximated. This may result in potential under-estimation of occupational risk exposure. This is a major limitation of DHM features in the domain of occupational risk prevention.

With such limitations and gaps in mind, the objective of this paper is to present a two-fold methodology to compute more reliable maximum actuation torques. Firstly, it uses musculoskeletal simulation to compute the upper and lower bounds of individual muscle tensions. Then, it uses mathematical objects called *zonotopes* to efficiently represent the corresponding maximal joint actuation torques. This approach is intended to account for the geometrical and physiological characteristics of the musculoskeletal system. The ultimate goal would be to implement this methodology as a “black-box” piece of software that would be embedded in DHM tools to improve the assessment of biomechanical risk factors from the early stages of work equipment design. As a first step towards this ambitious aim, this article describes *simMACT*, a demonstrator framework for the simulation of upper-limb maximum actuation torques.

The next section of this article highlights the limitations of current DHM features for occupational risk prevention. The third section describes the basics of the approach considered, namely *zonotopes* combined with musculoskeletal simulation. In the fourth section, we considered experimental measures found in the literature as a use case and simulated the maximum joint torques during an isometric task of the shoulder described by [7]. This section also considers simulations carried out with two DHM softwares commonly used by workstation designers for comparison purposes. The last two sections discuss this approach and consider the potential future use of this approach in combination with DHM simulation for ergonomic assessment of occupational tasks.

## **2 OCCUPATIONAL RISK PREVENTION: STRENGTHS AND LIMITATIONS OF DHMS**

DHM software is a valuable tool for manufacturers of work equipment, even from the early stages of the design process [8, 9, 10, 11]. Before any physical prototype is ready, designers may test several scenarios and create communication media (video, operational simulations) to discuss technical choices with their project partners (future users, occupational risk prevention experts, policy makers, etc.). They enable the calculation of biomechanical quantities such as interaction forces and wrenches, internal joint torques, as well as various statistical indices (for

instance the percentage of men and women who have the strength to perform the prescribed task). DHMs usually also integrate ergonomics tools such as RULA [12], REBA [13], EAWS [8] or OCRA [14]. Furthermore, advanced DHM features enable the simulation of the onset of muscle fatigue, which reduces the operators' force production capacity and modifies their postures and movements [15, 16, 17].

Commonly used DHM software are, for instance, Delmia Human® (Dassault Systèmes), Tecnomatix Jack / Process simulate® (Siemens) or 3DSSPP® (University of Michigan), which run poly-articulated rigid body models. Academic labs and industrial companies have also developed similar tools [18, 19]. One can also cite Santos® from the Virtual Soldier Research (VSR) program [20], which is a musculoskeletal human model.

Among their features, these tools may provide databases of exertable torques which come from numerous bibliographical sources. For instance, Delmia Human uses maximum joint torques from [21] for the shoulder and [22] for the elbow, and Jack® refers to [23] for the upper limb. Other articles describe similar experiments with complementary or contradictory results [24, 4, 5, 25, 6]. As stated in [4], these data may change drastically depending on both the subject (age, gender) and the experimental task to be performed: prescribed body posture, duration and repetition frequency, type of exertion (isometric, isokinetic or isotonic), etc. Two such experiments are detailed and replicated by simulation in section 4.

Combined with such maximum reference values, DHM simulations are widely used in the design of “ergonomically safer” industrial workstations and work equipment. However, this approach relies on some oversimplifications:

1. the effects of velocities and accelerations are ignored (simulation engines are based on quasi-static approximation, so dynamics and inertia effects are neglected);
2. for joints made of several degrees of freedom (DoF), such as the shoulder, DHM software assumes that joint torques are independent from one another. This assumption is referred to as the “independent axis approach” (IAA) by [7].
3. except for Santos®(which is a musculoskeletal human model), DHM models usually assume “ideal” joint actuation (as a localised hinge motor). Thus, the relationships between joint angle

/ torque and joint velocity / torque are not accounted for (see section 3.2).

Considering the first point, some DHMs have been developed which comply with dynamics [26, 27, 17]. They can provide more reliable instantaneous actuation torques, but the issue of assessing their maximum values remains.

The second point is a more significant one, as highlighted by [28, 29, 7, 30]. It results in strongly over-estimated maximum actuation joint torques. For instance, [7] found that the maximum 2D shoulder joint torque (no internal/external rotation of the upper arm) was over-estimated in 14 out of the 20 exertion directions analysed, with an average error about 5.7 N.m, and a peak difference up to +33% compared to experimental measurements. [30] identified even larger errors in a study focusing on female linear arm strength capabilities: the root mean square (RMS) error on forces was about 56 N, i.e. 40% of experimental observations. These two papers were focused on isometric exertions without the effect of velocity / inertia. If we consider concentric and eccentric efforts, when dynamics and the effects of muscle contraction velocity come in addition, differences between the simulation and measurements may become even greater.

Since IAA relies on strong approximations, [31] compared 2D-elliptic and linear interpolations of actuation torque to predict maximum actuation torques in dual DoF directions (elbow flexion/extension vs fore-arm pronation/supination, see figure 1). Similarly, [7] used elliptic interpolations to predict planar maximum actuation torques of the shoulder (see figure 1). Such an interpolation process is denoted the *Weighted Average Approach* (WAA). Computation of maximum torque is done by means of quadrants, needing at least one measurement in each cardinal direction. To address this point, [31] tried to determine averaged ratios between the measurement directions, but the statistical distribution of such ratios may depend greatly on inter-subject characteristics.

The last point may come down to the issue of simulating the physiological properties of muscles, such as the relationships between joint angle / torque and joint velocity / torque (see section 3.2) which are not accounted for in commercial DHM to date. Hence, we are still faced with the major limitation of assessing the maximum actuation torques during a generic activity remains and this shall be addressed in the next section.

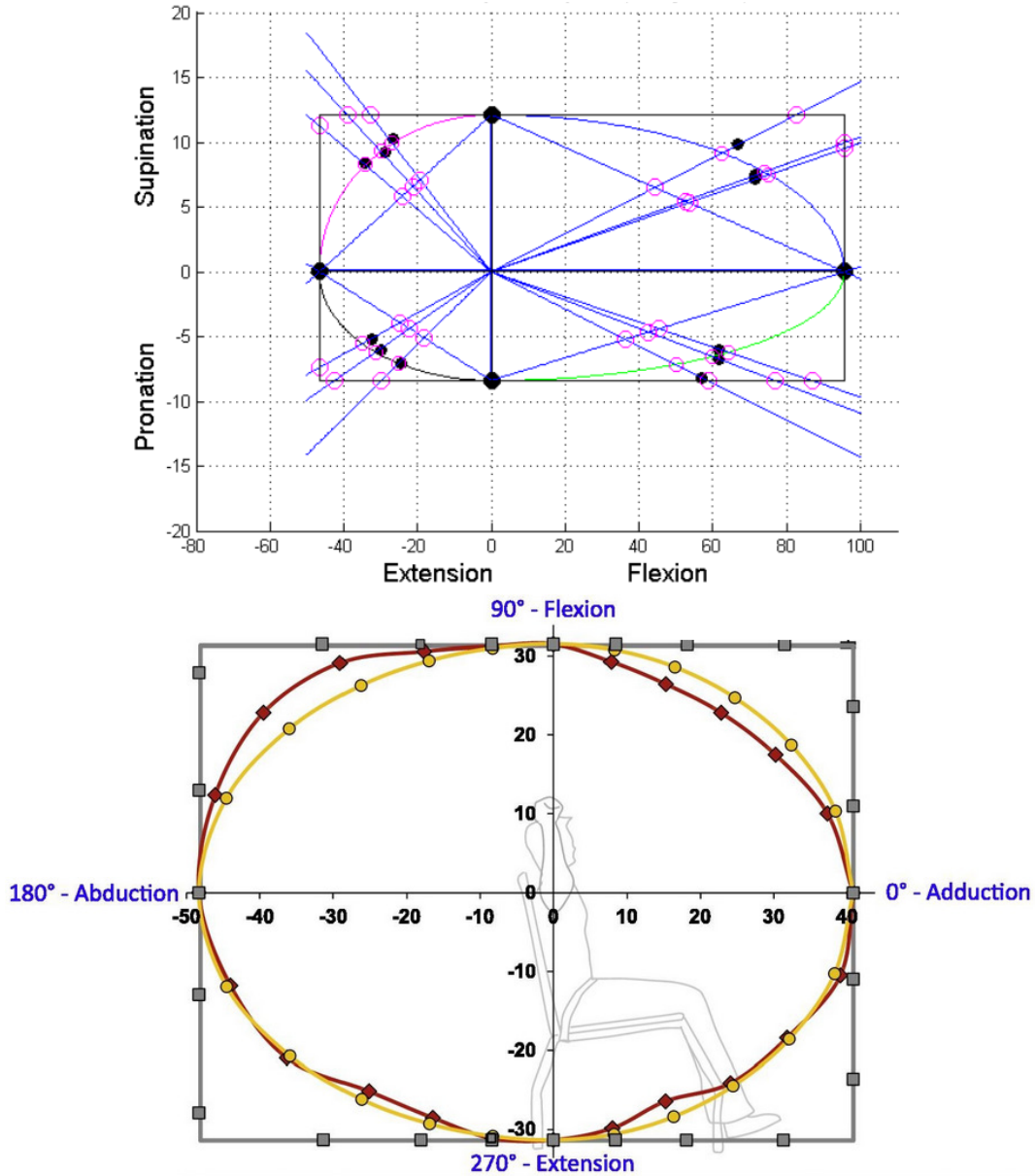


Fig. 1. Various representations of maximum actuation torques in dual 2D directions. Top: experimental measurements (black dots), elliptic and linear approximations (pink circles) for the fore-arm [31]. Bottom: experimental measurements (red) and elliptic approximations (yellow) for a planar shoulder exertion [7].

### 3 ZONOTOPES: A CONVENIENT TOOL TO COMPUTE MAXIMUM ACTUATION TORQUES

#### 3.1 Modelling joint actuation capacities

The upper limb is modelled as a system  $\mathcal{M}$  including  $m$  muscles and  $p$  rigid bodies linked together by  $n$  degrees of freedom (DoFs). As described in [29], muscles may be attached to

several bodies and contribute to the actuation of several DoFs. Hence, many types of coupling exist between joint angles, joint velocities, intra- and inter-joint actuation. This is confirmed by the literature [28, 25, 31, 7, 32], but not accounted for by current DHM simulation software [29, 7, 30].

In the remainder of this paper, we use the following notations:

$\mathbf{q} = [q_i \in [1, n]]^T$  is the vector of joint angles and  $\dot{\mathbf{q}} = [\dot{q}_i \in [1, n]]^T$  the vector of joint angles velocities;

$\mathbf{t} = [t_j \in [1, m]]^T$  is the vector of individual muscle tensions.  $\mathbf{t}^{min} = [t_j^{min} \in [1, m]]^T$  and  $[t_j^{max} \in [1, m]]^T$  are the vector of individual upper and lower bounds;

$\mathbf{N} = [r_{i,j}] \in \mathbb{R}^{n \times m}$  is the matrix of muscle moment arms ( $r_{i,j}$  is the moment arm of muscle  $j$  around DoF  $i$ ).  $\mathbf{N}$  depends on the current configuration (joint angles) of  $\mathcal{M}$ :  $\mathbf{N} = \mathbf{N}(\mathbf{q})$ ;

$\boldsymbol{\tau} \in \mathbb{R}^n$  is the vector of joint actuation torques of  $\mathcal{M}$ .

A basic relation links the vector  $\boldsymbol{\tau}$  of joint actuations and the vector  $\mathbf{t}$  of individual muscle tensions:

$$\boldsymbol{\tau} = \mathbf{N}\mathbf{t} \quad (1)$$

As each individual muscle tension is bounded, the set of all achievable muscle tensions is an *hyper-rectangle* (a box) of dimension  $m$  denoted  $\mathcal{C} = \{\mathbf{t} = [t_i]^T, t_i^{min} \leq t_i \leq t_i^{max}\}$ .

According to equation (1), the set of achievable joint torques is hence the image of  $\mathcal{C}$  by the linear mapping  $\mathbf{N}$ . Let us denote it  $\mathcal{Z}$ . By definition,  $\mathcal{Z} = \{\boldsymbol{\tau} = \mathbf{N}\mathbf{t}, \mathbf{t} \in \mathcal{C}\}$ . The literature shows that  $\mathcal{Z}$  is a convex *polytope* in  $\mathbb{R}^n$ , more precisely a *zonotope*. Because  $\mathbf{N}$  is a linear mapping, the zonotope  $\mathcal{Z}$  has specific geometric properties. For instance, its faces have a symmetry center and are parallels in pairs. Such properties make its computation efficient and easy to implement thanks to the Hyperplane Shifting Method algorithm [33]. The literature shows it is a suitable tool for robotics and DHM modelling and analyses [34, 35, 36, 37]. Each point *inside*  $\mathcal{Z}$  corresponds to an admissible set of actuating joint torques. Each point *on the frontier* of  $\mathcal{Z}$  is an *extremum* actuation, *i.e.* at least one individual muscle tension is extremum.



Classical linear algebra and geometrical computations in  $\mathbb{R}^N$  enable the calculation of the intersection of  $\mathcal{Z}$  with any (hyper)plane or line. These intersections can then be projected in specific 2D- or 3D-subspaces for illustration purposes. For instance, figure 3 in section 4 shows a 3D-zonotope.

### 3.2 Estimating the bounds of individual muscle tensions

The previous section stated that joint actuation capacities can be estimated from the orthotope of individual muscle tensions. A convenient way to estimate this orthotope is to use a musculoskeletal simulation tool. Several musculoskeletal simulation frameworks exist, for instance OpenSim<sup>1</sup>[38, 39], AnyBody™Technology<sup>2</sup> [40], BoB-biomechanics<sup>3</sup> [41], CusToM [42], etc.

In this study, we used the OpenSim software [38, 39]. The force exerted by a muscle is described by the equation

$$t = f_0 \left( a \cdot f^l(q) \cdot f^v(q, \dot{q}) + f_P(q) \right) \quad (2)$$

where  $f_0$  is the maximum isometric tension of the muscle,  $a$  is the activation signal sent by the central nervous system (CNS),  $q$  and  $\dot{q}$  are the joint angle and joint velocity,  $f^l$  and  $f^v$  are positive multipliers respectively describing the force-length and force-velocity relationships, and  $f_P$  stands for the passive force of the tendons and muscle connective tissues.

We have built our demonstrator upon the OpenSim programming interface along with Python scripts to keep the framework, the coding language and additional libraries and modules completely open source. However, any other musculoskeletal simulation engine could be used as well, provided its programming interface implements equation 2. Similarly, the literature describes numerous musculoskeletal models of the upper limb [43, 44]. We have chosen the “MOBL” upper limb model described by [45] “as is”, without any customization. This model corresponds to a male subject with a height of 1.77 m and a mass of 75 kg. It includes 50 muscles and 7 DoFs. However,

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<sup>1</sup><https://simtk.org/projects/opensim>

<sup>2</sup><https://www.anybodytech.com/>

<sup>3</sup><https://www.bob-biomechanics.com/>

other musculoskeletal models of the upper-limb could be used as well.

### 3.3 From any external wrench to associated joint actuation

Ergonomics assessments at the workstation require the calculation of maximum actuation torques during any occupational task. This implies the calculation of the maximum actuation torques for any interaction wrench applied by the musculoskeletal system  $\mathcal{M}$ . Let  $\mathbf{W} = (\vec{F}, \vec{\Gamma})^T$  be the interaction wrench (force and moment) at a point  $P$ . Let the subscript  $j$  or the superscript  $\star$  indicate the frame in which this wrench is expressed:  $\mathbf{W}^\star$  in the world frame  $\mathcal{R}_O^\star$  or  $\mathbf{W}_j$  in the local frame  $\mathcal{R}_j$  attached to the body  $\mathcal{B}_j$  of  $\mathcal{M}$ . This wrench is linked to the joint actuation by the classical relation of mechanics:

$$\boldsymbol{\tau} = \mathcal{J}_P^T \mathbf{W}_j \quad (3)$$

where  $\mathcal{J}_P^T$  is the transpose of the jacobian matrix  $\mathcal{J}_P(\mathbf{q})$  of the system.

Let us denote  $\mathbf{u}_W$  the unit vector bearing the actuation vector  $\boldsymbol{\tau}$ :

$$\mathbf{u}_W = \frac{1}{\|\mathcal{J}_P^T \mathbf{W}_j\|} \mathcal{J}_P^T \mathbf{W}_j \quad (4)$$

The desired max actuation torque set  $\boldsymbol{\tau}^{max}$  which complies with wrench  $\mathbf{W}$  is the intersection of the current zonotope  $\mathcal{Z}$  and the line  $\Delta$  directed by  $\mathbf{u}_W$ :

$$\boldsymbol{\tau}^{max} = \mathcal{Z} \cap \Delta \quad (5)$$

### 3.4 Code architecture and implementation

Our demonstrator framework was developed using an object-oriented programming approach for the sake of code modularity and inter-operability. Several modules have been designed, as

described below:

`MSM_Processor`: this module implements generic classes and functions to access OpenSim's API. This module could be adapted to any other simulation engine;

`ZNT_Processor`: this module implements generic classes and functions to initialize a zonotope (Hyper-plane Shifting Method as described by [33]), process it (compute intersection with an edge, a face, an hyperplane) and plot it. This module uses the `polytope` library<sup>4</sup>;

`ACT_Analyser`: this module describes an abstract class and functions to simulate experimental tasks. It may be derived into in specific classes to mimic specific experimental protocols such as experiments and analyses described by [28, 31, 29, 7, 32], etc.

The corresponding code is available online<sup>5</sup> and can be used under the terms of the 3-clause BSD license.

#### **4 CASE STUDY: SIMULATION OF AN EXPERIMENT FROM THE LITERATURE**

Since our simulation demonstrator is currently a proof of concept, this paper presents a use case based on an experiment described in the literature (we did not carry out any specific experiment). We compared these experimental data to a simulation of maximum actuation torques with our demonstrator `simMACT`, as well as with two DHM software applications denoted DHM-1 and DHM-2 in the next 2 sections. Appendix A gives more details about those simulations.

This use case considers the experiment carried out by [7]. It consisted in assessing the shoulder maximum isometric joint torque for 24 exertion directions, requiring 2 DoFs in a sagittal plane (see figure 2) while the posture remained unchanged. The 24 exertion directions were defined in a 2D-plane formed by the forward  $\vec{x}$  and vertical  $\vec{y}$  axes of the world frame, in intervals of 15°. It should be noted that the authors denoted these 2 main directions “flexion-extension” and “abduction-adduction” respectively. This may be confusing because:

the names and the axes of the DoFs considered may differ from those used by the musculoskeletal model or the DHM. For instance, the MOBL model uses the sequence of axis rec-

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<sup>4</sup><https://pypi.org/project/polytope/>

<sup>5</sup><https://github.com/INRS-France/simMACT>

ommended by the ISB [46], namely *elevation angle*, *shoulder elevation* and *shoulder rotation* but DHMs refer to other references; depending on the subject's posture, these orthopaedic directions may not be aligned with the main axis of the world frame.

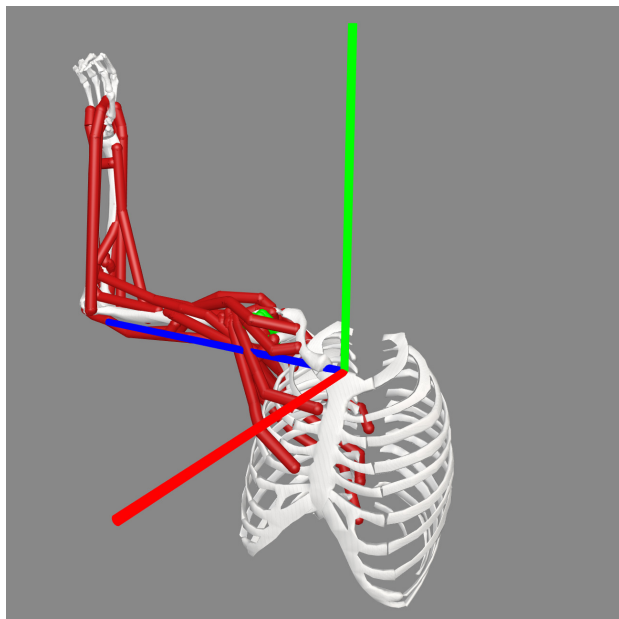


Fig. 2. MoBL musculoskeletal model in the posture of [7]'s experiment. OpenSim convention for the world frame:  $\vec{x}$ -axis (forward, red),  $\vec{y}$ -axis (upward, green) and  $\vec{z}$ -axis (rightward, blue).

We would like to express our gratitude to Professor J. Potvin for granting us access to the experimental data recorded in the article [7].

#### 4.1 **simMACT simulation**

To mimic Hodder's experiment, we have implemented the algorithm below in our framework demonstrator:

1. set the musculoskeletal model's static posture as in figure 2;
2. compute the moment arm matrix  $\mathbf{N}(\mathbf{q})$ ;
3. compute the torque zonotope  $\mathcal{Z}(\mathbf{q}, \dot{\mathbf{q}})$ ;
4. compute the jacobian matrix  $\mathcal{J}_P(\mathbf{q}, \dot{\mathbf{q}})$ ;

5. FOR  $\theta \in [0 : 15 : 360]$  ( $^\circ$ ):

- (a) compute the direction  $\mathbf{u}_{\mathbf{W}^{des}}$  of the desired exertion in the local frame of the upper-arm (see equation 4),
- (b) compute the intersection point  $P_\theta$  between  $\mathcal{Z}$  and the line  $\Delta$  directed by  $\mathbf{u}_{\mathbf{W}^{des}}$  (see equation 5). Its coordinates in the space of actuation torques are  $\tau_\theta^{max}$  and  $m_\theta = \|\tau_\theta^{max}\|$  is the magnitude of the maximum actuation torque.

Figure 3 illustrates the zonotope in the 3D-space of shoulder actuation torques for this shoulder task. It includes 304 vertices and 508 edges. Figure 4 illustrates the maximum actuation torques observed experimentally as well as those simulated with our `simMACT` framework and DHM-1 respectively. Unfortunately, the maximum actuation torque database included in the DHM-2 only provides values for one DoF (shoulder flexion-extension). No value is given for the abduction-adduction DoF when its angle is greater than  $0^\circ$  (see section A.2). In this polar representation, experimental maximum actuation torques have a piece-wise elliptical form. Musculoskeletal assessment of maximum torques demonstrates a piece-wise quasi-circular form, with two singularities when the exertion direction is antero-posterior. As described by [7], DHM simulations have a rectangular form due to the IAA.

As the musculoskeletal model embedded in `simMACT` is not parameterised according to the physical performance of the subjects who performed Hodder's experiment, we normalised all simulated data with respect to their maximal and minimal values, according to the relation:

$$m_{norm}^*(\theta) = \frac{m_\theta^* - m_{min}^*}{m_{max}^* - m_{min}^*} \in [0, 1] \quad (6)$$

where  $\star$  denotes either the simulated or measured values,  $m_{min}^* = \min_\theta m_\theta^*$  and  $m_{max}^* = \max_\theta m_\theta^*$ .

Finally, we also computed the difference  $\delta$  :

$$\delta = m_{norm}^{sim} - m_{norm}^{exp} \quad (7)$$

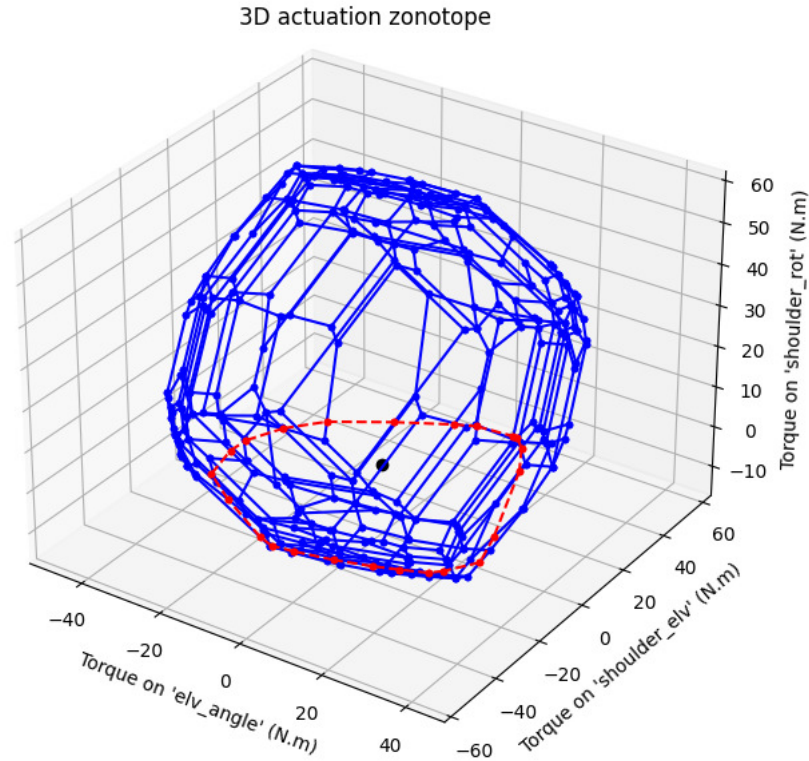


Fig. 3. Zonotope of actuation torques projected in the 3D-space of shoulder actuation torques for Hodder's experiment. The red polygone is the intersection between this 3D polytope and the plane  $\tau_{\text{internal/external arm rotation}} = 0$ .

Figure 5 presents the normalised actuation torques in a cartesian plot. In such a representation, experimental maximum actuation values have a roughly sinusoidal form. The musculoskeletal-based estimate demonstrates a maximal peak value in the same exertion direction as the experimental values, but also shows plateaus (quasi-circular parts of the associated polar representation).

## 5 DISCUSSION AND PERSPECTIVES

### 5.1 Comparison of experimental data and *simMACT* simulations

In section 4, we compared the experimental and simulated maximum actuation torques for 4 “cardinal” and 20 “combined” exertion directions. The range of actuation torques for the 2 DoFs considered are similar (DOF1  $\sim$  90 N.m; DOF2  $\sim$  60N.m). The diagram of experimental maximum torques shows an elliptic shape while *simMACT* shows a quasi-circular shape (see figure 4).

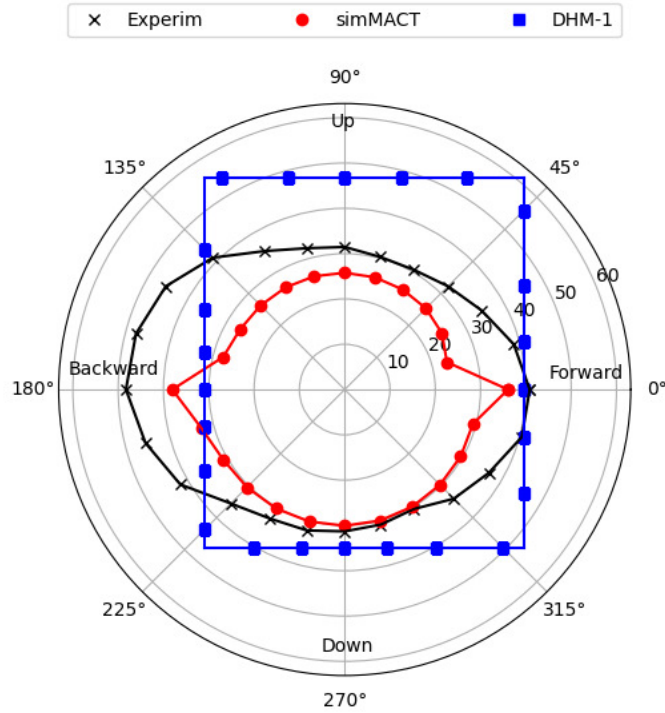


Fig. 4. Polar plot of experimental values and DHM computation of the maximum actuation torques in the 24 studied exertion directions described in Hodder's experiment. Black lines and crosses show experimental measures, red lines and dots correspond to *simMACT* simulation, blue lines and squares illustrate DHM-1 simulation.

		Relative error $\epsilon$ on actuation torques (%)	Difference $\delta$ on normalised actuation torques
<i>simMACT</i>	mean $\pm$ std	$-20 \pm 12$	$0.02 \pm 0.31$
	RMS	5	0.06
DHM-1	mean $\pm$ std	$18 \pm 31$	$0.08 \pm 0.53$
	RMS	7	0.11
DHM-2	mean $\pm$ std	N.A	N.A
	RMS	N.A	N.A

Table 1. Quantification of the error between the measured maximum torques and their simulated counterparts over the 24 studied exertion directions of Hodder's experiment.

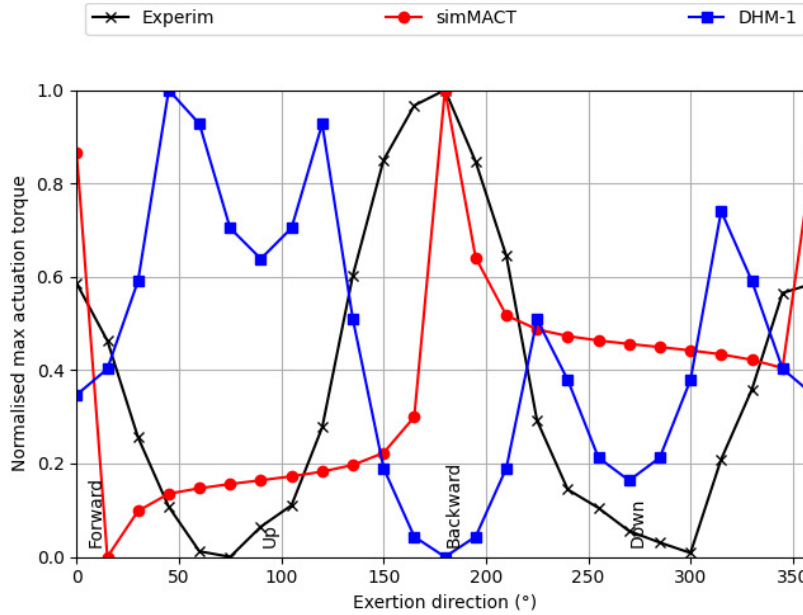


Fig. 5. Cartesian plot of the normalised experimental values and DHM computation of the maximum actuation torques in the 24 studied exertion directions. Black lines and crosses show experimental measures, red lines and dots correspond to *simMACT* simulation, blue lines and squares illustrate DHM-1 simulation.

Furthermore, the *simMACT* diagram is included within the experimental ellipsis. In other words, it never overestimates subjects' physical performances. The average relative error is about 5% for *simMACT*, slightly higher than that of the WAA model (1%).

## 5.2 Comparison between *simMACT* and DHM simulations

This section is intended to provide some outlooks on improving maximum actuation assessment in DHM software, rather than comparing the specific features of the DHM software under consideration.

According to Hodder's experiment, the shape of experimental maximum actuation torques is elliptical, thus less favourable to the IAA-based approximation used by both DHM software applications. DHM-1 simulations led to over-estimated maximum exertion in 19 out of 24 exertion directions and the signed average error is positive. Thus, in terms of risk prevention, this trend is not safe for future operators' health. In addition, its average and RMS differences in normalised actuation torques are 4 times and 2 times higher respectively than those of *simMACT*. Furthermore, normalised maximum actuations with DHM-1 reach their *minimum* value in the exertion direc-



tion where both experimental and musculoskeletal simulations have *maximal* values. Considering DHM-2, its database provides no maximum actuation value for one of the DoFs studied.

Hence `simMACT` appears to be a more generic and more reliable tool for occupational risk prevention than common DHM features and databases.

### 5.3 Perspectives

In this article, we described a methodology to estimate maximum actuation torques based on zonotopes and musculoskeletal simulation. The advantages of this method are as follows:

through 2D or 3D plotting, torque zonotopes allow a convenient graphical representation of unavoidable trade-offs between concurrent or coupled muscle/joint actuations. Plots as figure 4 may help biomechanics and risk prevention practitioners visualise how and why achievable performance varies depending on the posture of the subject / operators or the performed exertion;

torque zonotopes may also help researchers and experimenters designing experimental protocols or biomechanical models. For instance, they may help in identifying situations where the IAA may be a sufficient model or where WAA would work better. They would also help in identifying the order of magnitude of the ratio between the 2 DoFs maximum actuation, as proposed by [31] to relate  $\tau_{\text{forearm pronation}}^{\text{max}}$  (difficult to record) to  $\tau_{\text{elbow flexion}}^{\text{max}}$  (easier to record).

However, this paper leaves several questions unanswered:

the parametrisation of the musculoskeletal model is a fundamental issue as it conditions the upper and lower bound of individual muscle tensions. In this study, we used the MoBL musculoskeletal “as is”. We observed that maximal isometric shoulder actuation torques simulated for Hodder’s experiment were underestimated. Customising musculoskeletal models is a tricky task, as it requires customising the geometry, the kinematics and the dynamics of the model (body lengths, moment arms, positions of joints, masses and inertias) as well as muscle parameters. This involves dozens of parameters. Sensitivity analyses such as [47] may help in identifying the most influential ones. However, those issues are a matter of biomechanics and musculoskeletal modelling;

this study focuses on an isometric task. A similar study should be carried out for non-isometric tasks as well. The algorithms described above are still valid as long as the musculoskeletal simulation engine implements the force-velocity relationship (2). The simulation of isokinetic experiments such those described in [28, 32] is already implemented in `simMACT` and shall soon be laid out in a future paper;

this paper focuses on a 2-DoF task. [29] described a 3-DoF task implying shoulder elevation, elbow flexion and forearm pronosupination. This experiment has been implemented in `simMACT` and shall soon be simulated and analysed as well;

accounting for biomechanical constraints is still to be investigated. To date, `simMACT` does not account for stability constraints such as shoulder dislocation.

Once the above-mentioned issues are addressed, implementing this methodology in DHM software would greatly help in assessing maximum joint actuation in order to identify and reduce occupational risk from the early step of workstation design. It would provide workstation designers with more reliable maximum actuation values than those from the current sparse and imprecise databases. In particular, this methodology would account for the physiological characteristics of the musculoskeletal system through a “black-box”, requiring no extra skill in biomechanics from workstation designers and using only the same inputs as their current workflow and tools.

## **6 CONCLUSION**

In this paper we described a methodology based on zonotopes and musculoskeletal simulation in order to assess maximum actuation torques. This methodology has been implemented in `simMACT`, an opensource software demonstrator which relies on the musculoskeletal simulation engine OpenSim and the MoBL musculoskeletal model of the upper limb. Implementation of this software can be adapted to other musculoskeletal models and simulation engines as well.

This paper compares maximum actuation torques simulated according to this methodology to experimental data from one study described in the literature which focuses on a 2-DoF isometric task of the shoulder. This methodology leads to simulated actuations torque errors similar or smaller than those of mathematical models found in the literature (such as the Weighted Average

Approach or the Independent Axis Approach). Normalised simulated maximum actuation torques have been computed to compensate for the bias of musculoskeletal customisation issues. Errors on normalised torques are equivalent or smaller than those computed with 2 Digital Human Model software commonly used in workstation design.

This validation process should be continued with more generic tasks (isokinetic or dynamic tasks) and of higher dimension (more than 2 DoFs involved). If these encouraging results are confirmed, this methodology may open up interesting new perspectives for various fields such as sport, biomechanics or rehabilitation. It could also be applied to ergonomics, occupational health and safety and risk prevention during the early stages of workstation design, for instance through its implementation in DHM software.

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## A SIMULATION OF MAXIMUM ACTUATION TORQUES WITH DHM SOFTWARE

This appendix explains how the DHM assessments of maximum actuation torques have been computed. Within the Tecnomatix Jack® DHM (v 8.4) software, such computation is performed using the optional “Task Analysis Toolkit” module. It relies on the 3DSSPP® software. Within the Dassault Systèmes Human® DHM software (V5R20), such computation is performed using the “Biomechanics single Action Analysis” feature.

Considering Hodder’s experiment, We defined a female manikin for each DHM software, parameterised as the average subject of Hodder’s experiment (height 166.5 cm, mass 67.2 kg). Table 2 shows the maximum shoulder torques in the four main exertion directions for both DHM applications, if available. Figure 6 illustrates the associated DoF-components and the magnitude of maximum joint torques in the 24 experimental directions. See the following sections for specific computation details.

Max torque (N.m)	Forward	Backward	Upward	Downward
Jack®	39.5	30.8	46.8	34.9
Human®	27	27	NA	NA

Table 2. Max actuation torques of the shoulder in the four main directions for the average subject of Hodder’s experiment computed by Jack® and Human® DHM software applications (NA=not available).

### A.1 Simulation with Jack®

We used Tecnomatix’s female model, Jill®. Jill’s anthropometric waist-to-hip ratio was kept at its default value of 74%. Its posture was set to shoulder elevation angle 90°, anterior angle 0°, humeral rotation -90°, elbow flexion 90°. The external force was applied at the `right_lower_arm.left` site (see figure 7).

### A.2 Simulation with Human®

We used Human’s female model. Its posture was set to shoulder flexion angle 90°, abduction angle 80°(its max value), humeral rotation 0°, elbow flexion 90°. The external force was applied at

the bottom of the `fore-arm` segment (offset 145 mm from its centre of mass, see figure 8). No max actuation torque data is available for abduction angles greater than 0°.

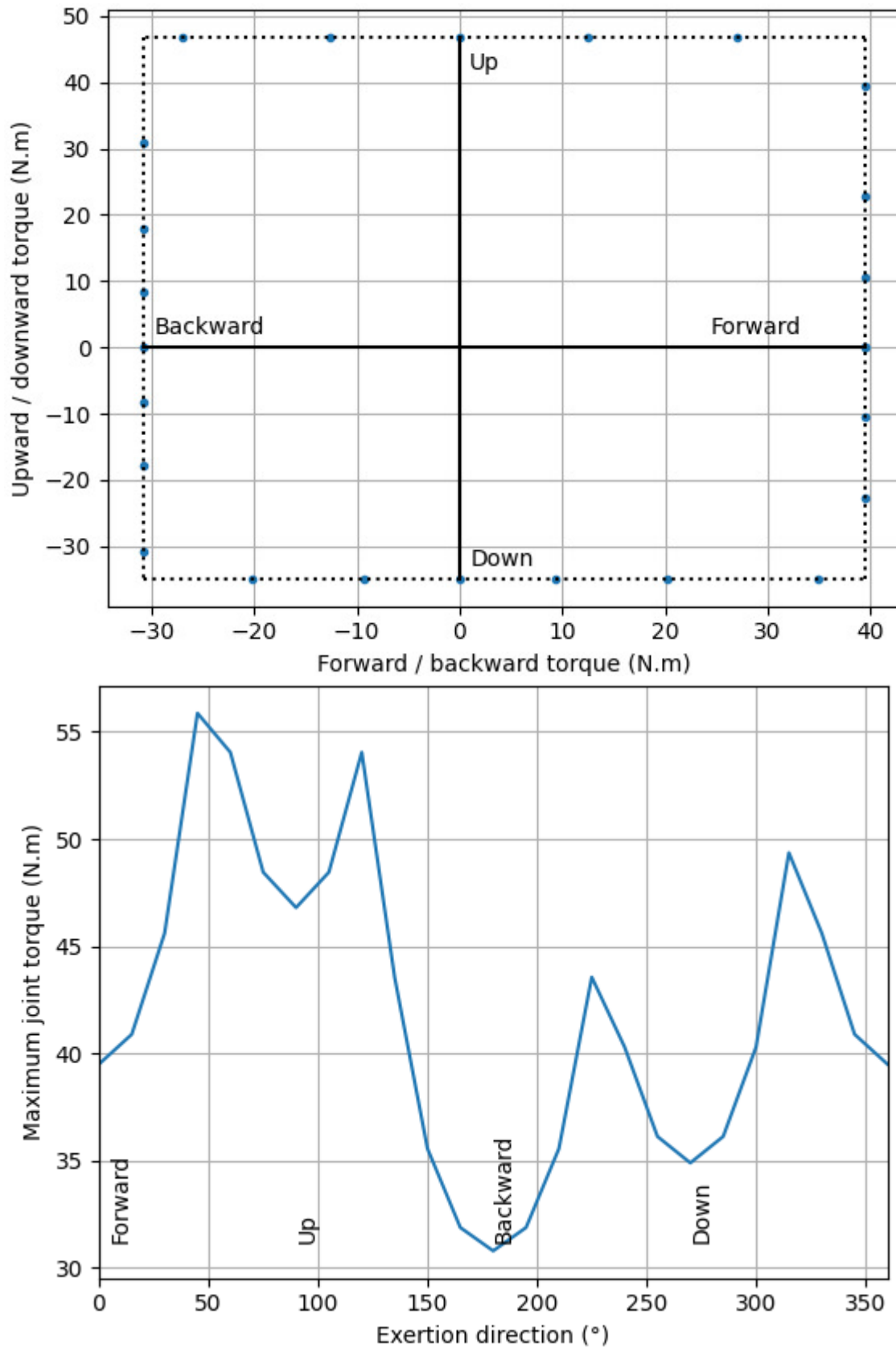
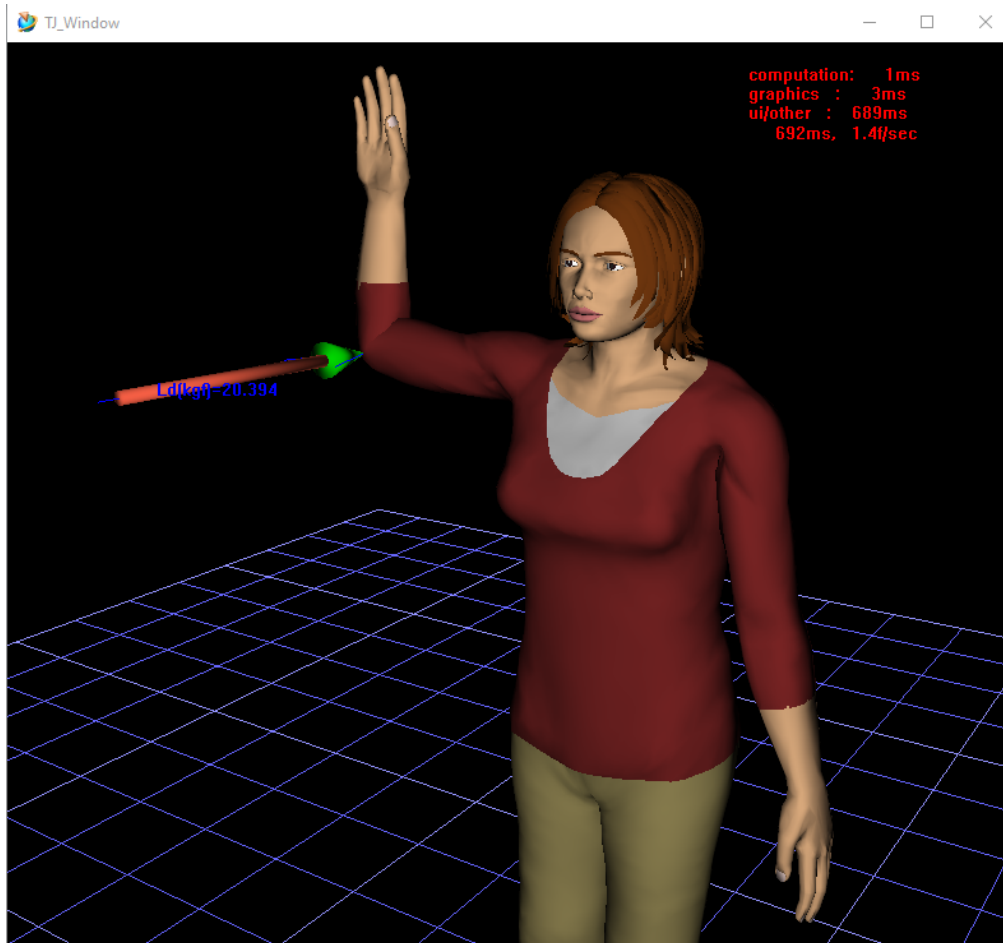


Fig. 6. The DoF components (top) and the magnitude (bottom) of the maximum joint actuation as computed by Jack® and Human® for Hodder's experiment.



Human Control Panel

Human: human

Standing

Right Left

### Loads and Weights

Force Distribution Strategy: Two Feet

right\_lower\_arm.left : global = 200.000newtons (-0.002)

Load Editor

Site: human.right\_low

Type:  Force  Weight 200.000 newtons

Direction

X Y Z: -0.002 -0.015 -1.000

Adjust

Local

### Bracing Settings

Lock task hand force direction

Display initial hand force(s)

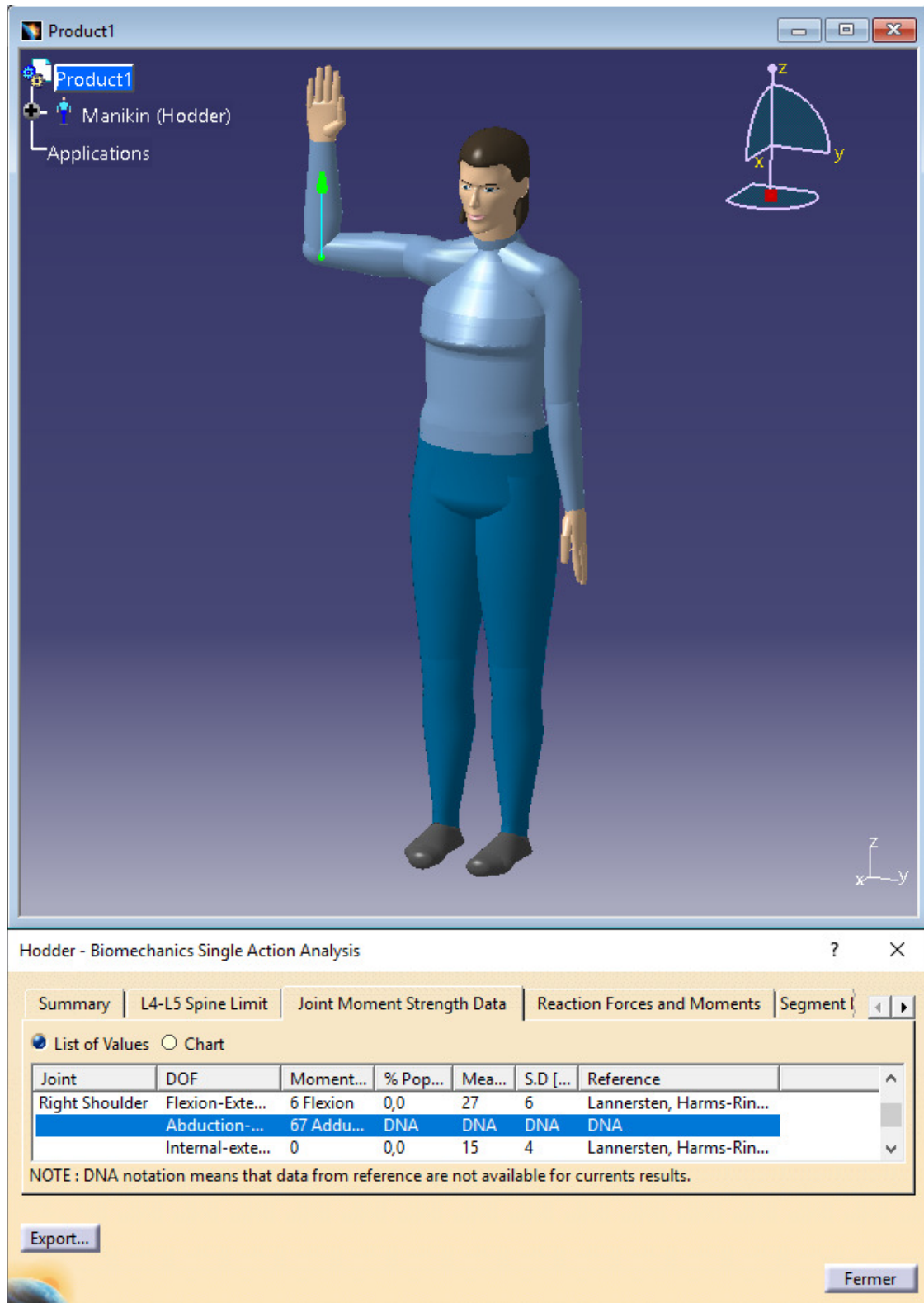


Fig. 8. Human® female DHM in the posture of Hodder's experiment. The green arrow represents the external force applied to the arm. No max actuation torque data is available for abduction angles greater than 0°.