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Abstract: We investigated the individual muscle contribution to arm motion to better understand the complex muscular coordination underlying three-dimensional (3D) reaching tasks of the upper limb (UL). The individual contributions of biceps, triceps, deltoid anterior, medius, posterior and pectoralis major to the control of specific degrees of freedom (DOFs) were examined: Using a scaled musculoskeletal model, the muscle excitations that reproduce the kinematics were calculated using computed muscle control and a forward simulation was generated. During consequent perturbation analyses, the muscle excitation of selected muscles was instantaneously increased and the resulting effect on the specific DOF was studied to quantify the muscle contribution. The calculated muscle contributions were compared to the responses elicited during electrical stimulation experiments. Innovative in our findings is that muscle action during reaching clearly depended on the reaching trajectory in 3D space. For the majority of the muscles, the magnitude of muscle action changed and even reversed when reaching to different heights and widths. Furthermore, muscle effects on non spanned joints were reported. Using a musculoskeletal model and forward simulation techniques, we demonstrate individual position-dependent muscle contributions to 3D joint kinematics of the UL.

I, Annelies Vandenberghe, certify that all authors have seen and approved the manuscript being submitted:

# Quantifying individual muscle contribution to three-dimensional reaching tasks

**Annelies Vandenberghe [a,*], Lode Bosmans [a], Joris De Schutter [b], Stephan Swinnen [a], Ilse Jonkers [a]**

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I warrant that the article is my original work and that the article has not received prior publication and is not under consideration for publication elsewhere. On behalf of all co-authors, I shall bear full responsibility for the submission.

Scientific responsibility for this work remains with the authors. The authors would like to thank Katherine Holzbaur for the permission to use her upper limb model, and all volunteers who participated in this study.

# Conflict of interest

I, Annelies Vandenberghe, hereby certify that there are no conflicts of interest

***3. Response to Reviewers**

Dear Mr Theologis,

Dear reviewers,

Thank you very much for the detailed review of our manuscript that we have amended to include some of your comments – highlighted in yellow.

Reviewer comments that were unclear to us or that we have chosen not to revise in the manuscript are discussed and argumented below.

We also thank reviewer 2 for the interesting remarks on the role of mono- and biarticular muscles. Given the word count of the manuscript we decided not to further extend on this – but we do hope that we can elaborate on this issue in a follow-up manuscript.

I hope that you can approve with the current revised manuscript and thank you again for your useful suggestions.

Kind Regards,

Ilse Jonkers

Reviewer #1:
General Comments
1. I strongly recommend the author describe briefly the content about methods or protocol for those had been published, rather than a reference number only.

***In order to limit the word count and as the methodology was previously published in Gait & Posture, we have indeed limited the description of the measurement protocol to the specific trials used for the comparison between estim and free reaching conditions. However, we feel that the description in the methods section supplemented with the caption of figure 1, adequately represents the essentials of the measurement set-up and protocol.***

2. Can the isolated and visible muscle contraction cause the joint movement in non-antigravity during electrical stimulation? It's a different story from visible contraction between without joint motion and with joint motion. The criteria of the electrical threshold were rough and needed to be revised.

***We agree partially with the reviewer that the criteria were defined roughly. However, we did contribute considerable time to fine tune the approach for determining the stimulation level and considered specific ways to come to a more standardized determination of the stimulation intensity: We included the isolated check of the muscle contraction to avoid the comment that the stimulation was not selective and that surrounding muscles were stimulated as well. We did incrementally increase the stimulation level till we observed contraction of the surrounding muscles. Consecutively we then decreased the stimulation intensity again till a selective contraction was seen. Applying this stimulation level during motion did indeed cause the observed end-point perturbation. We carefully checked all individual trials to verify the presence of end-point deflection. For PMAJ no consistent end-point deviations could be elicited therefore these results were excluded from the paper?***

***The approach used here is similar to the approach used in (1) Stewart C, Postans N, Schwartz MH, Rozumalski A, Roberts AP (2008). An investigation of the action of the hamstring muscles during standing in crouch using functional electrical stimulation (FES). Gait Posture. 2008 Oct;28(3):372-7. A,d (2) Stewart C, Postans N, Schwartz MH, Rozumalski A, Roberts A. An exploration of the function of the triceps surae during normal gait using functional electrical stimulation. Gait Posture. 2007 Oct;26(4):482-8.***

***We have modified the description in the methodology slightly to better reflect the strategy for determining the stimulation intensity slightly to 'Original stimulation intensity (20mA) was increased till visible co-contraction was induced and the submaximal stimulation level was used during the experiment.'***

3. Why did the perturbation processing include PMAJ, but not stimulated during estim trials?

***We were unable to establish a stimulation intensity that induced a repetitive, selective stimulation that affected the end-point trajectory (that could be tolerated by the test subject). Given the variability in the response (e.g less than 3 trials of 5 with effective end-point perturbation, we decided not to include the estim results for the validation of the perturbation analysis. We added to the discussion'***

***'As the response for estim of PMAJ was highly variable between trials with less than 3 trials producing end-point deviation, these estim results were not included in this paper.'***

4. The agreement or the difference between simulation and estim trials should be quantified or statistically analyzed, rather than the descriptions in results.

***The estim experiments were used to validate the model response in a qualitative rather than a quantitative way. Given the 'rough' nature of the estim experiments we would not expect to find a quantitative agreement…..as this would assume that a direct relation between the estim intensity and the activity level of the muscle in the model could be established. Such an analysis would require an EMG calibration procedure and such an EMG driven analysis would not be appropriate as it still would require to relate estim intensity to the muscle force level produced. Therefore, the authors feel that a quantitative or statistical comparison is not felt to be appropriate for this reason.***
***Estim and model perturbation do both result in an instantaneous perturbation of the muscle force equilibrium. Therefore, agreement of these results in terms of the direction of the induced changes in joint angles is considered indicative of an agreement of the effect of the induced muscle force on the UL joint kinematics. In contrast, if both results would be opposite, this is indicative that the effect of changes in the muscle force affects the joint kinematics differently in the model than through estim.***

***The authors are clear about the purpose of the estim validation: it serves to validate the direction of the muscle action rather than the magnitude of the muscle action. This validation procedure is presented additional to anatomical studies as we feel that many anatomical studies are unable to account for altered joint configurations as present in this 3D reaching task. Both anatomical and estim results contribute to the validation of the model response.***

5. Can the position before stimulation be provided in Table 1? It can not make sense whether the effect is from stimulation or target position.

***The data presented in Table 1 are the position of the limb the instant prior to the stimulation i.e. the sample prior to the occurrence of the stimulation pulse. They therefore represent the angle configuration immediately prior to the time of the stimulation. As 3D data is acquired at 100 Hz but analogue data (including the stimulation pulse) is acquired at 1000 Hz, this approach allows to reliably detect the joint angles prior to the stimulation. If we would aim for the 3D data closest to or immediately following the stimulation, we cannot reliably assess if the joint angle configuration was not yet influenced by the stimulation.***

***The current statement in the caption of table 1. 'Joint positions and standard deviation of shoulder and elbow joint at the moment of excitation' is not entirely correct given the data analysis approach used as it could indeed be related to the effect of the estim. We altered this to***

*'Joint positions and standard deviation of shoulder and elbow joint immediately prior to the moment of excitation for the nine different target conditions (in degrees).'*

***I do not understand the second remark by the reviewer that questions the effect to target position. Target position indeed influences the UL kinematics – this is exactly the point we want to make. As experimentally the stimulation intensity and as the perturbation of the excitation is identical for different targets, changes in kinematic response can therefore be related to changes in the muscle action/contribution to segmental acceleration and therefore changes in joint angle configuration. The fact that we only considered the kinematic changes during a limited time window following the perturbation reassures this statement.***

Reviewer #2:

The authors present an interesting paper investigating muscle contributions to three-dimensional reaching tasks. The focus on reaching tasks is important and has applications in many areas, eg. ergonomics and clinically. I feel that the paper should be accepted, subject to minor changes approved by the editor.

One of my concerns, is that the authors have not reviewed, contrasted and compared their work to earlier work by authors, eg. Van der Helm and De Groot. Although the current work may be more sophisticated, forward dynamic modelling of the upper limb to investigate the contribution of muscles to anatomical movements has been undertaken. This should be integrated into the paper.

***A thorough literature review was conducted at the time of the manuscript submission. No forward dynamic simulations of the upper limb that include a similar perturbation analysis were found in literature. The reviewer is correct to direct towards the work of Van der Helm and De Groot. Many of their publications however relate to inverse dynamic analysis that do per definition not allow the here presented perturbation analysis e.g. Steenbrink F, de Groot JH, Veeger HE, van der Helm FC, Rozing PM. Glenohumeral stability in simulated rotator cuff tears. J Biomech. 2009 Aug 7;42(11):1740-5. As well as : Nikooyan AA, Veeger HE, Westerhoff P, Graichen F, Bergmann G, van der Helm FC. Validation of the Delft Shoulder and Elbow Model using in-vivo glenohumeral joint contact forces. J Biomech. 2010 Nov 16;43(15):3007-14. Therefore and given the word count, we decided not to include a discussion of the manuscripts related to these inverse dynamic analyses.***

***Since initial submission of the manuscript we became aware of the manuscript by Asadi Nikooyan A, Veeger HE, Chadwick EK, Praagman M, van der Helm FC. Development of a comprehensive musculoskeletal model of the shoulder and elbow. Med Biol Eng Comput. 2011 Oct 29. This manuscript is closely related to the current work and provides an extensive comparison with other shoulder models. However, it however does not include a perturbation analysis and therefore does not present material for comparison with the results presented in this paper.***

***Given the substantial contribution of this paper and its use as future reference for the readers, we decided to include a reference in the discussion: '. Recent work[28] cautions for the sensitivity of these assumptions to the model predictions in terms of muscle forces and calculated joint contact forces. Duplication of the current perturbation approach in different models is therefore mandatory.'***

I would also appreciate a better explanation in the Discussion, eg. page 9, about "reversal of contribution". The authors state that the muscle moment arm can change and therefore change the muscle function, but some precise examples are required.

***Rereading this section, I think that the reviewer is right in questioning its phrasing.***

***Furthermore, for specific muscles a reversal of the contribution was reported when the reaching trajectory was modified; e.g. muscle action of DELTM changed from horizontal shoulder abduction for medial to adduction for lateral reaching, whereas muscle action of TRI changed from horizontal shoulder adduction for medial to abduction for lateral reaching.***

***Now the manuscript states:***

***Both these findings confirm that when reaching in 3D space, the task-specific kinematics of the shoulder has the potential to affect the instantaneous moment arm of the muscle, affecting the magnitude of and even invert the muscle moment and the consequent observed muscle action. These changes relate to the joint position-specific muscle length and moment arms and the consequent muscle moment generating capacity.***

***A more correct and cautious wording would be:***

***Both these findings confirm that when reaching in 3D space, the task-specific kinematics of the shoulder affects the instantaneous moment generating capacity of the muscle through the joint position-specific effect of muscle length and moment arms. Consequently, although increasing the muscle excitation with a fixed amount, the moment balance at the joints is affected differently and opposite kinematic responses may be observed. As these mainly relate to secondary functions that do not oppose gravity, a reversal of the associated instantaneous moment arm should be considered as well.***

***This phrasing does accounts for a potential effect of gravity on the observed kinematic response. However, it additionally states that it is unlikely that this is the dominant effect as the reversal of function is mainly observed for movement directions that are not against gravity and therefore reversal of the instantaneous moment arm is the most plausible explanation. However, it cannot be guaranteed. This can only be explored in simulations that exclude the effect of gravity. Although technically feasible, the value of such simulations would be limited for real life situations.***

The role of mono- and bi-articular muscles is very interesting. Maybe the authors do not have enough space to expand on this topic. IS it possible to comment on the synergistic role of the muscles in specific tasks ? For example, does the bi-articular muscle that appears to have negligible effect on the shoulder have an important effect in orienting the head of the humerus so that the mono-articular muscles are more effective ? Does the model itself limit the conclusions that can be drawn about these muscles ? Validating the results against anatomical literature is not a good method. One would hope that the model is anatomically correct, therefore muscle function should reflect what has been postulated anatomically. The alternative method would be to examine discrete conditions where specific muscles are not working, and then simulate them. That might confirm and explain what has been observed clinically.

***In my view, there are two main replies towards this comment:***

- ***To evaluate the interaction effects of the bi- and monoarticular muscles further data analysis of the perturbation results is needed. It would require to first establish if and to what extent mono-articular muscles are capable to compensate for the action of bi-articular muscles (the authors did present similar work for the lower limb Jonkers I, Stewart C, Spaepen A. The complementary role of the plantarflexors, hamstrings and gluteus maximus in the control of stance limb stability during gait. Gait Posture. 2003 Jun;17(3):264-72. ) We do consider a similar analysis for the upper limb. However, if we want to extent the results of such an analysis above the level of a pure sensitivity analysis, the main difficulty is in finding a good clinical model to evaluate the predicted compensatory muscle reactions. Ongoing research is focusing exactly on this.***
- ***The assumption that the model is valid as it reflects anatomy is not entirely true. Not taking into account concerns related to the variability of data reported in anatomical literature, the current dynamic model also includes parameters of the muscle force model that are less well described in literature. Their effect on muscle force prediction and therefore the moment generating capacity is known. Therefore, the presented model response also accounts for the validation of these parameters as well as the known geometrical relationships.***

Title Page (with authors and addresses)

# Quantifying individual muscle contribution to three-dimensional reaching tasks

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## Abstract

We investigated the individual muscle contribution to arm motion to better understand the complex muscular coordination underlying three-dimensional (3D) reaching tasks of the upper limb (UL). The individual contributions of biceps, triceps, deltoid anterior, medius, posterior and pectoralis major to the control of specific degrees of freedom (DOFs) were examined: Using a scaled musculoskeletal model, the muscle excitations that reproduce the kinematics were calculated using computed muscle control and a forward simulation was generated. During consequent perturbation analyses, the muscle excitation of selected muscles was instantaneously increased and the resulting effect on the specific DOF was studied to quantify the muscle contribution. The calculated muscle contributions were compared to the responses elicited during electrical stimulation experiments. Innovative in our findings is that muscle action during reaching clearly depended on the reaching trajectory in 3D space. For the majority of the muscles, the magnitude of muscle action changed and even reversed when reaching to different heights and widths. Furthermore, muscle effects on non spanned joints were reported. Using a musculoskeletal model and forward simulation techniques, we demonstrate individual position-dependent muscle contributions to 3D joint kinematics of the UL.

# 1. Introduction

Arm movements in three-dimensional (3D) space rely on multi-joint coordination. Despite the task-specific coordination of a large number of degrees of freedom (DOFs), arm movements are performed effortlessly through specific control of a large number of muscles spanning the different joints. A better insight into muscle actions during arm movements will provide a better understanding of functional consequences of muscle impairment and the effect on the coordination of the different arm segments.

Previous descriptive studies analyzed arm movements in two[1-5] and to a lesser extent in three dimensions[6-8]. Relating muscle activation sequences to the kinematics of the different DOFs, individual muscle action was derived. However, these studies are unable to document muscle coordination in terms of the contribution of individual muscles to the 3D movement of the individual upper limb (UL) segments and the resulting inter-segmental angles. A muscle-driven forward simulation of arm movement that uses a model describing the musculoskeletal geometry in combination with a muscle model can relate muscle excitation to muscle force production and consequent segmental motion. Major challenge is to define an appropriate set of muscle excitation patterns that exactly reproduces the measured kinematics[9-11]. Once defined, the excitation pattern of the individual muscles can be altered (perturbation analysis) and the resulting changes in the kinematics provide an indication of the individual muscle contribution. This approach was already used to examine the muscle contributions during walking[12,13]. A similar analysis has to the best of our knowledge not yet been attempted to analyze muscle contributions to reaching tasks.

Validation of these model-based muscle contributions is an obvious concern. Classically, anatomical studies that apply loads to the individual muscle-tendon units in a static limb position and measure joint rotations are used for comparison. Alternatively, kinematic changes induced through electrical muscle stimulation (estim) were used to validate muscle contributions during pathological gait[14]. The use of estim to validate the contribution of individual UL muscles is to our knowledge nonexistent.

This study quantifies the contribution of six major UL muscles (biceps, triceps, deltoid anterior/medius/posterior and pectoralis major) to the UL kinematics during a 3D reaching task using muscle-driven forward dynamic simulations. For the selected muscles, the calculated contributions are compared to published anatomical studies and estim experiments.

# 2. Methods

### 2.1. Subjects

Five healthy right-handed subjects (20-30 years) participated in this study. The protocol was approved by the ethical advisory board of KULeuven. All subjects gave written informed consent.

### 2.2. Experimental 3D reaching task

Subjects performed 3D reaching tasks to nine standardized targets according to a previously described protocol[15] (see caption figure 1): (1) During the *free reaching condition,* subjects reached five times to each target holding the pointer with minimal trunk movement. (2) During the *estim trials,* subjects performed identical reaching movements holding the pointer fixed to a robot arm (5DOF Performer MK2, Eshed Robotec) that passively followed the movement and that determined the instant of stimulation. At one third of the reaching distance, individual muscles (deltoid anterior/medius/posterior, biceps or triceps longus) were stimulated by a short, single electro impulse (Grass Stimulator S88, Astro-Med Inc.) through a surface stimulation electrode on the muscle belly (Seniam protocol[16]) and with the reference electrode placed on the plexus brachialis. Prior to the reaching experiment, a stimulation verified if the intensity elicited an isolated, visible muscle contraction. Original stimulation intensity (20mA) was increased till visible co-contraction was induced and the submaximal stimulation level was used during the experiment. Stimulation intensity varied between 15mA and 35mA for different subjects and muscles. Given the intensity and duration of the total measurement protocol, this stimulation was only explored for five targets (MH, ML, NH, NM and NL). Subjects reached 35 times to each target: randomly five stimulations of each muscle and ten intermittent trials without stimulation were included. The subjects reached 220 times in total. Muscle fatigue was avoided by resting periods between the different reaching conditions, resulting in total measurement duration of more than two hours.

A dual beam camera system (Krypton, Metris, Leuven) registered the trajectories of 26 active infrared markers (figure 1) at 100Hz during both reaching conditions.

### 2.3. Data analysis

Using the measured marker trajectories, dynamic simulations of the different reaching movements were generated. Figure 2 shows the workflow in Opensim 1.8.1.[19] (see appendix): (1) A generic musculoskeletal model of the UL[17] modified to include three DOFs in the shoulder[18] and two in the elbow, was scaled for each subject, (2) inverse kinematics and (3) computed muscle control (CMC)[20] were conducted. (4) Using the initial states and muscle excitations computed during CMC, a muscle-driven forward dynamic simulation was generated that reproduced the measured kinematics (reference simulation). (5) In a final perturbation step, muscle excitations of deltoid anterior (DELTA), medius (DELTM) and posterior (DELTP), biceps longus (BIC), triceps longus (TRI) and pectoralis major pars clavicularis (PMAJ) were increased instantaneously (duration of 0.005 s) to half of the maximal muscle excitation (0.5) at one third of the reaching movement.

To quantify the individual muscle contributions, the effect of the increased excitation on individual DOFs was evaluated by calculating the mean difference between the kinematics of the reference and perturbed simulation over a 0.1s window following the perturbation. This window allowed characterizing the representative kinematic changes without destabilizing the forward simulation. The main effect of reaching height and width on the individual muscle contribution was evaluated statistically for the nine targets using a repeated measure ANOVA. Only results with significance level $p \leq 0.05$ are reported.

To quantify the effect of estim of individual muscles, the mean difference between the estim kinematics and the kinematics of the reference reaching task was calculated over a 0.1s window following estim. To compare the muscle contributions observed during the simulations and the estim, both effects were expressed as a % of the total change in DOFs for each target and each muscle. Muscle contribution of the simulations and estim were judged to agree if their directions were in agreement.

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# 3. Results

### 3.1. Target-dependent UL kinematics

Table 1 presents shoulder and elbow kinematics immediately prior to the instant of perturbation for the nine target conditions. Reaching width (medial to lateral) increased horizontal shoulder abduction, external shoulder rotation and elbow flexion. Reaching height (low to high) increased shoulder elevation and elbow flexion.

### 3.2. Individual muscle contribution (Figure 3)

Main actions of DELTA were horizontal shoulder adduction, shoulder elevation and elbow extension, followed by internal rotation and pronation. With increasing reaching height, horizontal shoulder adduction (p=0.045), elevation (p=0.010), internal rotation (p=0.008) and elbow extension (p=0.018) decreased significantly. When reaching laterally, horizontal shoulder adduction (p=0.020), internal rotation (p=0.001) and forearm pronation (p=0.001) increased significantly.

Main actions of DELTM were shoulder elevation, internal rotation and horizontal ad- or abduction, depending on target location. An additional action was elbow extension. With increasing height, shoulder elevation (p=0.008), internal rotation (p=0.001) and elbow extension (p=0.000) decreased significantly. When reaching laterally, shoulder internal rotation (p=0.010) decreased significantly and horizontal abduction decreased and reversed to horizontal adduction for lateral reaching.

Main action of DELTP was horizontal shoulder abduction. Additional actions were shoulder elevation or depression, depending on target location, external shoulder rotation, elbow flexion and forearm supination. With increasing height, shoulder horizontal abduction (p=0.034) decreased significantly. When reaching more laterally, shoulder depression (p=0.015) decreased significantly. Furthermore, for LH and LM reaching, its contribution changed from shoulder depression into elevation.

Main actions of BIC were respectively forearm supination and elbow flexion, followed by minimal horizontal shoulder abduction and elevation. Despite a small effect on shoulder rotation, its contribution changed from external rotation for high reaching to internal rotation for low reaching (p=0.002).

Main action of TRI (elbow extension) reduced significantly when reaching more laterally (p=0.009). Additional actions were, depending on target location, horizontal ad- or abduction, forearm supination and shoulder depression. With increasing reaching height, forearm supination (p=0.040) increased significantly.

Main actions of PMAJ were horizontal shoulder adduction and internal rotation, followed by forearm pronation, shoulder depression and elbow extension. With increasing height, horizontal adduction decreased significantly (p=0.003). When reaching laterally, internal rotation (p=0.016), elbow extension (p=0.046) and shoulder depression (p=0.010) increased significantly.

### 3.3. Validation with estim

The validation of the calculated muscle contribution with estim is shown in figure 4.

Estim of DELTA did confirm the main muscle actions from the simulation for low and middle reaching. For high reaching, only horizontal adduction during NH reaching and elbow extension during MH reaching were confirmed.

Estim of DELTM confirmed shoulder elevation, internal rotation and horizontal ad- or abduction, but not shoulder abduction during NH reaching. The response in forearm rotation was more pronounced during estim.

Estim of DELTP confirmed horizontal abduction, except for NH reaching. For the secondary actions, the sign of the results of the estim were more variable.

Estim of BIC confirmed the main muscle actions found in simulation for all targets. Only its role in elbow flexion during NH reaching and in supination during NL reaching was not confirmed.

Estim of TRI confirmed elbow extension, except for NH. For the secondary actions, the sign of the results were more variable.

# 4. Discussion

In this study, a muscle-driven forward simulation was perturbed to investigate the contribution of six UL muscles to the control of shoulder and elbow DOFs. We hypothesized that due to the underlying kinematics, the calculated muscle contribution would change when reaching to targets at different heights and widths. Therefore, the space dependency of the muscular control of the DOF of the UL can be investigated. To validate the calculated muscle action, electrical stimulation (estim) was applied at the same instance of the reaching trajectory for a subset of the targets.

The specific contribution of individual muscles to the UL motion can be described (figure 3):
*At the shoulder*, DELTA and, to a lesser extent PMAJ, induce horizontal adduction. Activity of DELTP induces horizontal abduction assisted by BIC. DELTA and, to a lesser extent DELTM and BIC, induce elevation. TRI, PMAJ and DELTP cause depression. DELTM, PMAJ and, to a lesser extent, DELTA induce internal rotation. DELTP and TRI induce external rotation. The effect of DELTM and TRI on the elevation angle and BIC on shoulder rotation depends on the 3D position.
*At the elbow,* TRI assisted by DELTA, DELTM and PMAJ, induce elbow extension. BIC assisted by DELTP, induce elbow flexion. Whereas BIC and, to a lesser extent DELTP and TRI induce supination, DELTA assisted by PMAJ and DELTM cause pronation.

The position-dependent magnitude of the muscle contribution confirms that when reaching in 3D, the task specific kinematics determine the contribution of individual muscles to the control of individual DOFs. On the one hand, effects of reaching height are described with e.g. increased horizontal shoulder adduction, elevation, internal rotation and elbow extension for DELTA for low reaching and increased supination for TRI for high reaching. On the other hand, additional effects of reaching width are found with lateral reaching inducing reduced elbow extension for TRI but increased horizontal shoulder adduction, internal rotation and forearm pronation for DELTA.

Furthermore, for specific muscles a reversal of the contribution was reported when the reaching trajectory was modified; e.g. muscle action of DELTM changed from horizontal shoulder abduction for medial to adduction for lateral reaching, whereas muscle action of TRI changed from horizontal shoulder adduction for medial to abduction for lateral reaching. Both these findings confirm that when reaching in 3D space, the task-specific kinematics of the shoulder has the potential to affect the instantaneous moment arm of the muscle, affecting the magnitude of and even invert the muscle moment and the consequent observed muscle action. These changes relate to the joint position-specific muscle length and moment arms and the consequent muscle moment generating capacity.

UL muscles present a secondary action at joints they do not span[11]. Mono-articular shoulder muscles have a dominant effect on the shoulder and only a minor effect on the elbow: DELTA and to a lesser extent DELTM and PMAJ, induce shoulder elevation that due to the segmental position of upper- and forearm, induces elbow extension and pronation. DELTP induces elbow flexion and supination. Bi-articular muscles spanning the elbow and the shoulder present the dominant effect at the elbow with minor effect at the shoulder: Whereas BIC presents only a negligible effect on the shoulder the effect of TRI on the DOFs of the shoulder must be recognized. In addition to the remotely induced segmental accelerations, changes in segment orientation with respect to gravity will contribute to these remote responses.

A first way of validating the calculated muscle contribution is through comparison with anatomical literature[21-25]: The main actions of DELTA (shoulder horizontal adduction and elevation) and the increasing role of DELTA in shoulder elevation with increasing horizontal adduction are in line with literature. Literature confirms the main actions of DELTM (shoulder elevation, internal rotation and horizontal ad- or abduction). However, the literature describes DELTM mainly as a vertical abductor and elevator, with the effect on horizontal adduction and internal rotation being less well documented. Only Kronberg[23] described DELTM as an internal and external rotator, especially with vertical abduction of the UL. The primary actions of DELTP reported in this study (horizontal shoulder abduction, external rotation and depression) are in line with previous studies. However, the primary action in horizontal abduction contradicts the previously reported main action as a shoulder depressor. The main actions of PMAJ (horizontal adduction and internal rotation) are described in literature. Whereas our study identifies this muscle to contribute to shoulder depression, most studies mentioned PMAJ as a shoulder elevator. Only Brown[21] described PMAJ pars clavicularis as a shoulder depressor. The actions of the elbow muscles were compared with literature[22-26]: In agreement, BIC presents a main action in supination and secondly in elbow flexion. We found however no previous study reporting the effect of BIC on shoulder rotation. Whereas the description of TRI as primary elbow extensor is in line with literature, the width-dependent reversal of horizontal adduction for the medial targets, but abduction for the neutral and lateral targets has not been described before.

We conclude that for most muscles, the main actions are in agreement with literature. Differences are mainly related to position-dependent effects. Anatomical literature mainly investigated isometric shoulder tasks, standard movements in 2D, or was based on cadaveric studies. Studies describing muscle action during 3D reaching tasks are nonexistent, but the results of this study indicate the need to further explore the effect of 3D motion on joint kinematics and underlying muscle contributions. The use of forward simulation and perturbation analysis is innovative and potentially very useful for this as it allows identifying muscle strategies underlying 3D reaching task while accounting for the effect of target location in 3D space.

A second method to validate the calculated muscle contribution is by comparison with the results of the estim. As the response for estim of PMAJ was highly variable between trials with less than 3 trials producing end-point deviation, these estim results were not included in this paper. Estim confirms the main muscle action of DELTM, DELTP and TRI, but not the main actions of DELTA and BIC. Estim confirms the sign of all the main muscle contributions (DELTM, DELTP, BIC and TRI, but not DELTA high reaching). In contrast, for the secondary muscle actions more discrepancies are found. The difficulty of stimulating one specific shoulder muscle is an important confounding factor. Despite the careful verification of the stimulation effect prior to the experiment, the actual shoulder kinematics during the experiment could influence the actual stimulation site slightly. The use of a robot arm during estim is known to only influence minimally shoulder and elbow kinematics[27]. It is therefore expected that this will only minimally contribute to the observed differences.

Several critical considerations can be made related to this study: The forward simulations are based on a simplified shoulder model and the assumptions made by CMC to estimate muscle excitations depend on one specific mechanical cost function i.e. minimization of muscle activity. Recent work[28] cautions for the sensitivity of these assumptions to the model predictions in terms of muscle forces and calculated joint contact forces. Duplication of the current perturbation approach in different models is therefore mandatory. During the perturbation analysis, we increased the individual muscle excitation to 0.5. As the excitation level in the reference simulation differs for the individual muscles, the resulting relative increase in muscle excitation differs.

Furthermore, we only report muscle contributions related to the kinematics at one third of the reaching cycle. However, the current analysis can easily be extended to other time instants.

Finally, we only report the results for a selected muscle group. Muscle selection was however based on the feasibility to perform estim as this allows validation of the model and the presented workflow. The validity of the experimental platform used for estim (robot arm) to control the 3D reaching movements was studied in previous research[27].

To conclude, this study analyzed the contribution of six individual muscles to the control of the UL-DOFs during 3D reaching using forward dynamic simulations combined with perturbation analysis. The main findings of our study are the position-dependent magnitude and the position-dependent reversal of the calculated muscle contribution for the different DOFs. Furthermore, our results confirm muscle action effects on non-spanned joints in the UL. We conclude that this model and workflow is valid for investigating the individual contribution of selected muscles. The proposed method has potential not only to identify the impact of individual muscle actions during different 3D-UL tasks but also to explore the effect of muscle weakness on UL motion.

## Acknowledgements

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## Captions

**Figure 1:**

Test setup: targets were placed at arm length, (1) at three different widths: at center line (at line of sternum) (Medial), 45° lateral from center line (Lateral), intermediate position (Neutral), and (2) three different heights: at shoulder height (Middle), 45° above shoulder height (High) and 45° below shoulder height (Low). This resulted into nine reaching conditions: medial and low (ML), medial and middle (MM), medial and high (MH), neutral and low (NL), neutral and middle (NM) – the reference reaching task, neutral and high (NH), lateral and low (LL), lateral and middle (LM), and lateral and high (LH). 26 active infrared LEDs were placed on upper arm, lower arm and trunk consisting of 5 clusters fixed on lower arm, upper arm, clavicle, back and thorax and 6 LEDs attached to anatomical body points (radial styloid, ulnar styloid, lateral epicondyle, medial epicondyle, acromion, and processus xyphoideus). One LED was attached to the back of the hand, three LEDs were placed on the head. The LEDs were registered by two cameras, placed four meter away from the subject, one at 90° at the right side and one at 45° at the right/front side of the subject.

**Figure 2:**

Workflow used for the calculation the muscle contribution: (1) scaling of the musculoskeletal model using measured marker positions, (2) inverse kinematics using the experimentally measured 3D marker trajectories to calculate the time-dependent changes in DOFs, (3) the muscle excitations that track the experimental kinematics were calculated using CMC, (4) muscle-driven forward dynamic simulations were generated to reproduce the experimentally measured kinematics, and (5) in a final perturbation step the individual muscle excitations was increased up to 0.5 over a period of 0.005 seconds to calculate the instantaneous effect on the kinematics of the different DOFs.

**Figure 3:**

Effect of increased excitation of the six muscles in the reaching simulation on the DOFs of the shoulder and elbow for the nine targets. The effect is expressed as average of the differences between the kinematics of the reference and perturbed simulation over a 0.1s window following the perturbation (in degrees).

**Figure 4:**

Muscle contribution as deduced from the electrical stimulation experiments (black bars) compared to the effect of perturbation analysis (grey bars). For each muscle, the effects are expressed as a % of the total change in DOFs for each target.

**Table 1:**

Joint positions and standard deviation of shoulder and elbow joint immediately prior to the moment of excitation for the nine different target conditions (in degrees). Conventions used for the different DOF: Elevation angle: + horizontal adduction and – abduction, shoulder elevation: + elevation and – depression, shoulder rotation: + internal and – external rotation, elbow flexion: + flexion and – extension, forearm rotation: + pronation and – supination.

## Figures

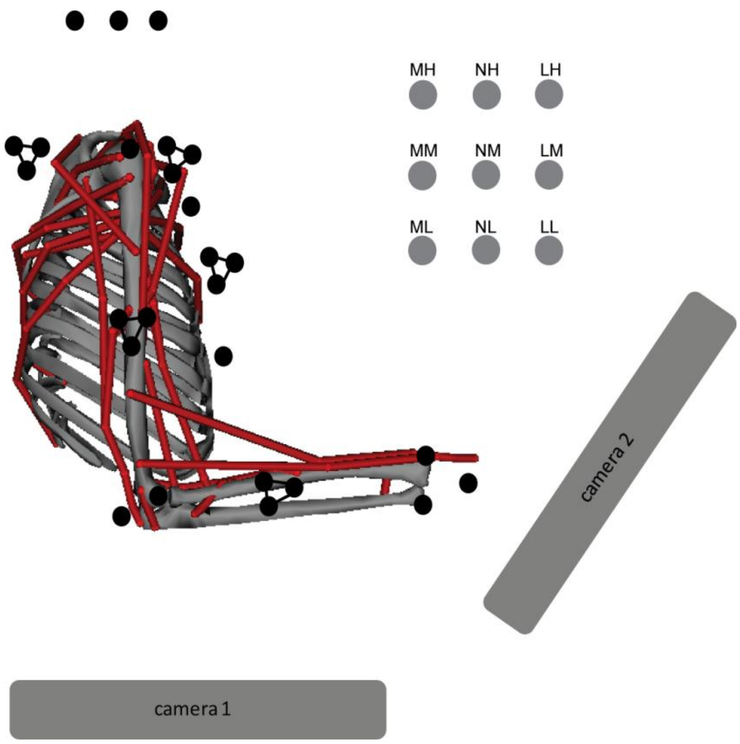
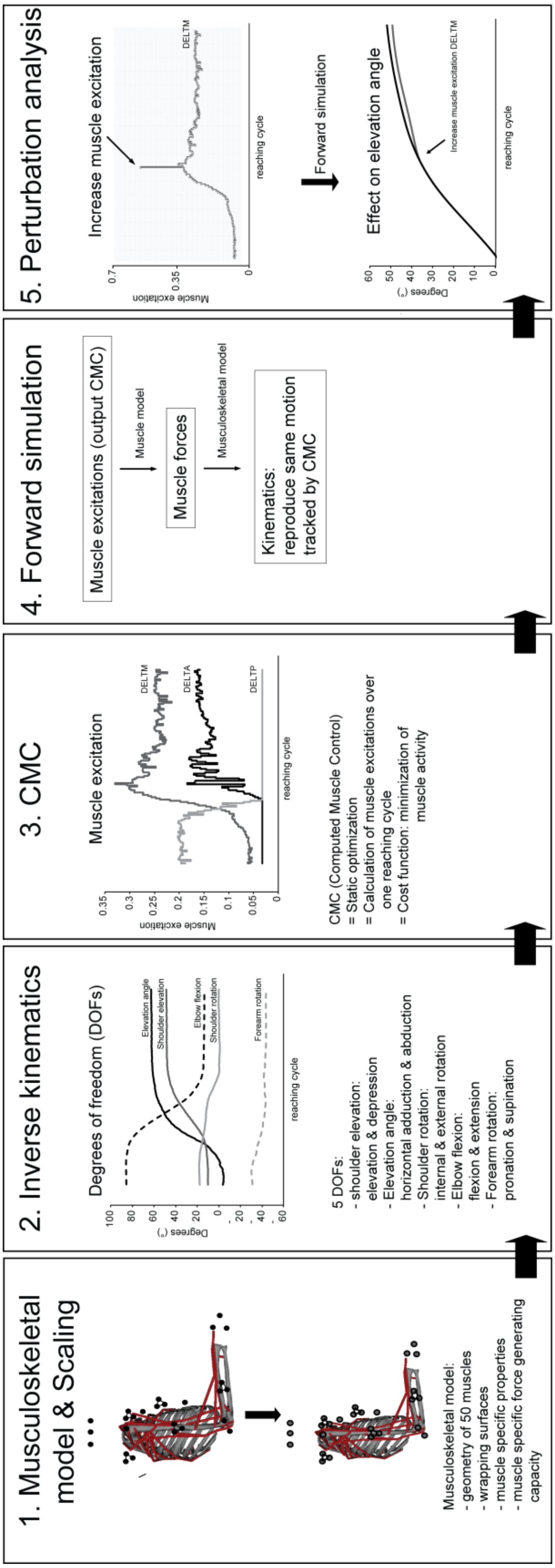



Figure 1



1. Musculoskeletal model & Scaling
Musculoskeletal model:
- geometry of 50 muscles
- wrapping surfaces
- muscle specific properties
- muscle specific force generating capacity
2. Inverse kinematics
Degrees of freedom (DOFs)
Degrees (°)
Elevation angle
Shoulder elevation
Elbow flexion
Shoulder rotation
Forearm rotation
reaching cycle
5 DOFs:
- shoulder elevation:
elevation & depression
- Elevation angle:
horizontal adduction & abduction
- Shoulder rotation:
internal & external rotation
- Elbow flexion:
flexion & extension
- Forearm rotation:
pronation & supination
3. CMC
Muscle excitation
Muscle excitation
DELTM
DELTA
DELTP
reaching cycle
CMC (Computed Muscle Control)
= Static optimization
= Calculation of muscle excitations over one reaching cycle
= Cost function: minimization of muscle activity
4. Forward simulation
Muscle excitations (output CMC)
Muscle model
Muscle forces
Musculoskeletal model
Kinematics:
reproduce same motion tracked by CMC
5. Perturbation analysis
Increase muscle excitation
Muscle excitation
DELTM
reaching cycle
Forward simulation
Effect on elevation angle
Degrees (°)
Increase muscle excitation DELTM
reaching cycle


Figure 2

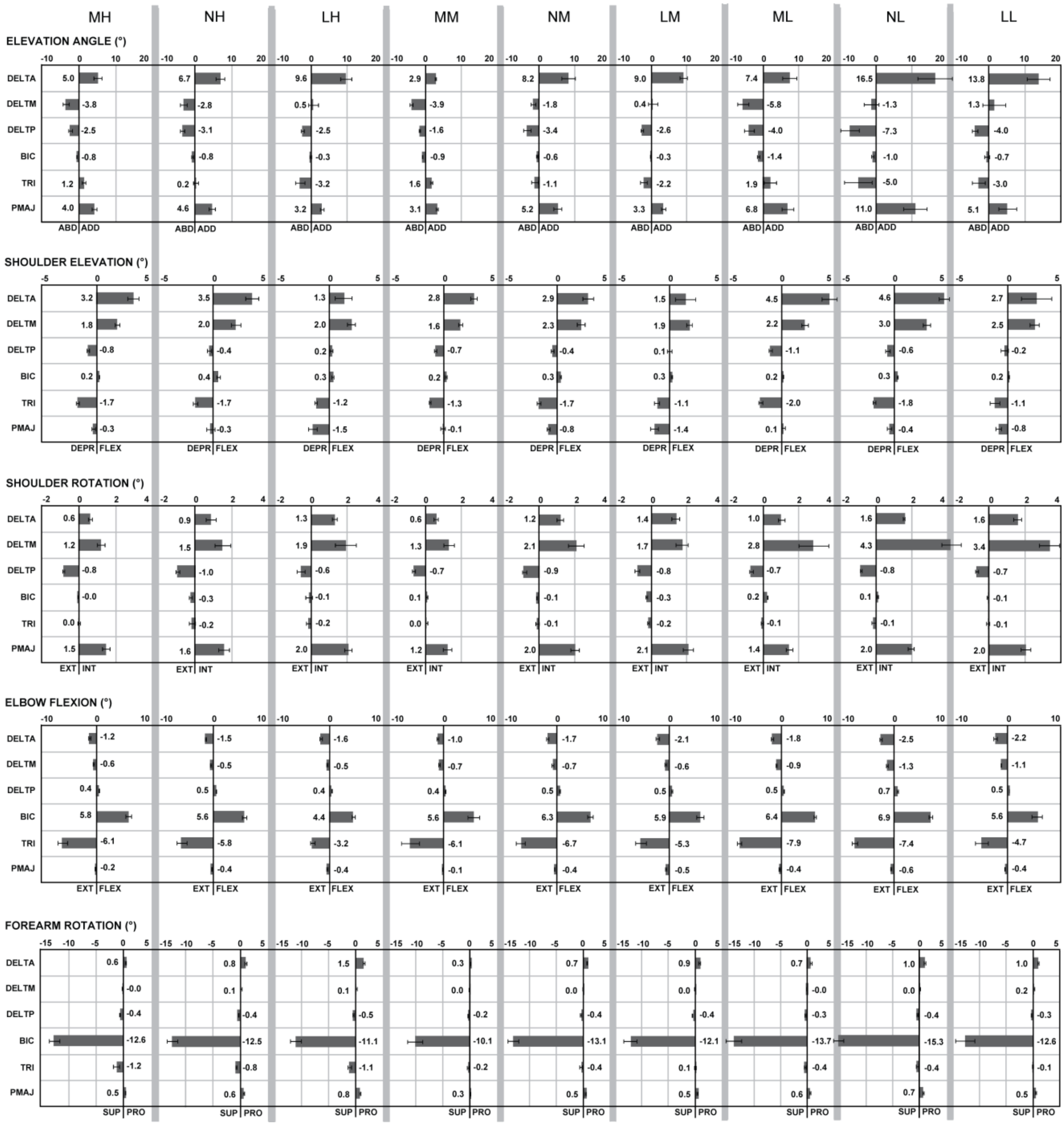

MH
NH
LH
MM
NM
LM
ML
NL
LL
ELEVATION ANGLE (°)
SHOULDER ELEVATION (°)
SHOULDER ROTATION (°)
ELBOW FLEXION (°)
FOREARM ROTATION (°)
DELTA
DELTM
DELTP
BIC
TRI
PMAJ
ABD ADD
DEPR FLEX
EXT INT
EXT FLEX
SUP PRO

Figure 3

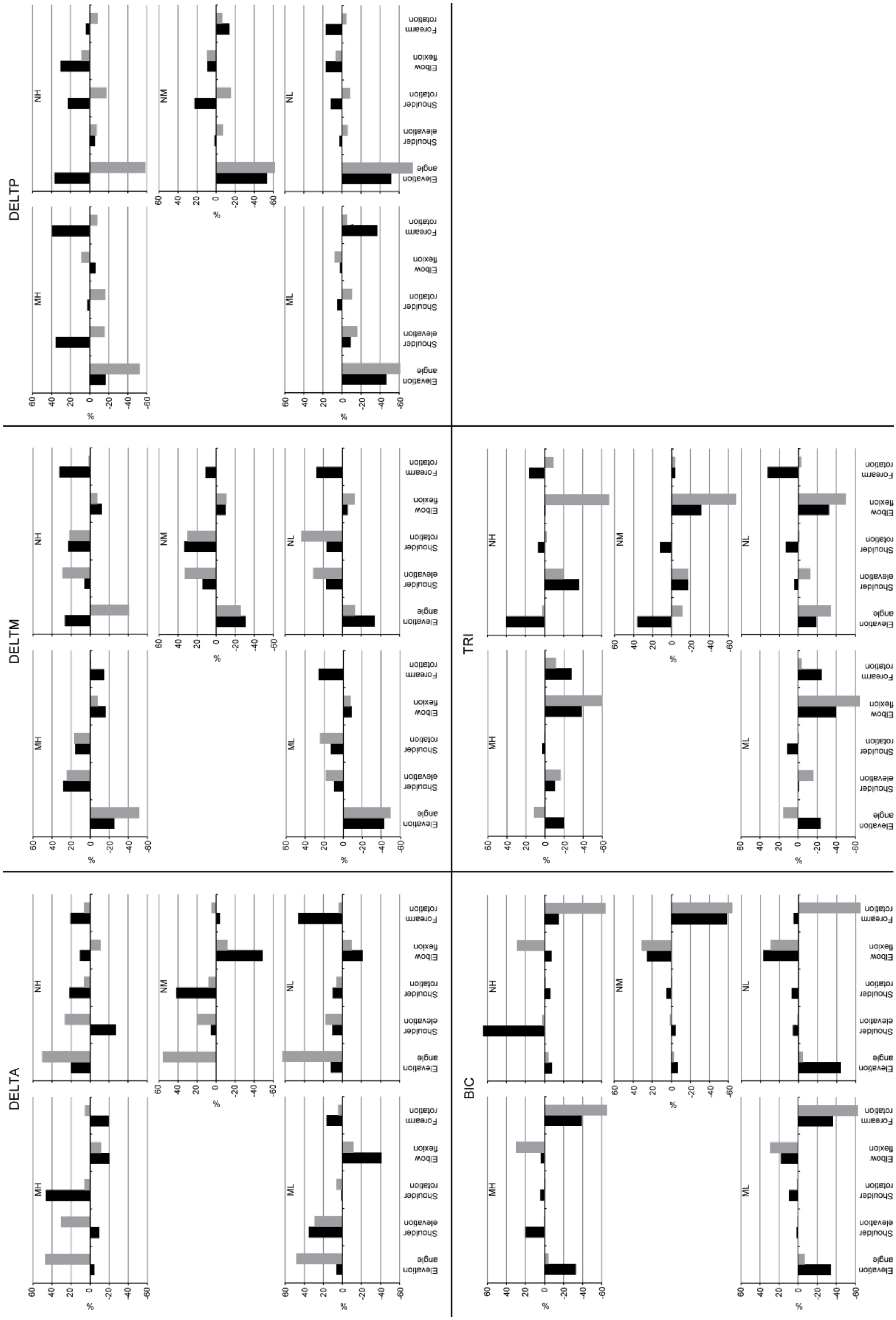

DELTP
DELTM
DELTA
TRI
BIC
MH
NH
NM
ML
NL
Elevation angle
Shoulder elevation
Shoulder rotation
Elbow flexion
Forearm rotation
%
60
40
20
0
-20
-40
-60

Figure 4

| | MH | NH | LH | MM | NM | LM | ML | NL | LL |
|---|---|---|---|---|---|---|---|---|---|
| **Elevation angle (°)** | 59 (±11) | 51 (± 12) | 24 (± 29) | 67 (± 9) | 47 (± 20) | 21 (± 32) | 60 (± 13) | 40 (± 29) | 17 (± 53) |
| **Shoulder elevation (°)** | 26 (± 8) | 26 (± 10) | 27 (± 8) | 23 (± 8) | 23 (± 9) | 24 (± 6) | 17 (± 9) | 15 (± 8) | 13 (± 4) |
| **Shoulder rotation (°)** | 17 (± 4) | 14 (± 6) | 11 (± 8) | 18 (± 2) | 13 (± 4) | 8 (± 7) | 18 (± 3) | 11 (± 2) | 3 (± 8) |
| **Elbow flexion (°)** | 86 (±7) | 87 (± 4) | 93 (± 8) | 80 (± 7) | 84 (± 7) | 89 (± 8) | 74 (± 6) | 77 (± 4) | 83 (± 6) |
| **Forearm rotation (°)** | -29 (± 9) | -29 (± 7) | -29 (± 6) | -26 (± 8) | -30 (± 7) | -33 (± 6) | -24 (± 7) | -26 (± 8) | -26 (± 5) |

Table 1

Appendix

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