An inverse dynamic analysis on the influence of upper limb gravity compensation during reaching

J.M.N. (Hans) Essers[a], Kenneth Meijer[b]
Department of Human Movement Sciences
Maastricht University Medical Centre +
The Netherlands
[a]hans.essers@maastrichtuniversity.nl,
[b]kenneth.meijer@maastrichtuniversity.nl

Alessio Murgia
Center for Human Movement Sciences
University Medical Center Groningen
The Netherlands
a.murgia@umcg.nl

Arjen Bergsma
Department of Rehabilitation, Nijmegen Center for Evidence Based Practice, Radboud University Nijmegen Medical Centre and Department of Human Movement Sciences, Maastricht University Medical Centre +
The Netherlands
a.bergsma@reval.umcn.nl

Paul Verstegen
Focal Meditech BV, Tilburg
The Netherlands
p.verstegen@focalmeditech.nl

***Abstract*—Muscular dystrophies (MDs) are characterized by progressive muscle wasting and weakness. Several studies have been conducted to investigate the influence of arm supports in an attempt to restore arm function. Lowering the load allows the user to employ the residual muscle force for movement as well as for posture stabilization. In this pilot study three conditions were investigated during a reaching task performed by three healthy subjects and three MD subjects: a control condition involving reaching; a similar movement with gravity compensation using braces to support the forearm; an identical reaching movement in simulated zero-gravity. In the control condition the highest values of shoulder moments were present, with a maximum of about 6 Nm for shoulder flexion and abduction. In the gravity compensation and zero gravity conditions the maximum shoulder moments were decreased by more than 70% and instead of increasing during reaching, they remained almost unvaried, fluctuating around an offset value less than 1 Nm. Similarly, the elbow moments in the control condition were the highest with a peak around 3.3 Nm for elbow flexion, while the moments were substantially reduced in the remaining two conditions, fluctuating around offset values between 0 to 0.5 Nm. In conclusion, gravity compensation by lower arm support is effective in healthy subjects and MD subjects and lowers the amount of shoulder and elbow moments by an amount comparable to a zero gravity environment. However the influence of gravity compensation still needs to be investigated on more people with MDs in order to quantify any beneficial effect on this population.**

Keywords—gravity compensation; zero gravity environment; lower arm support; joint moments; inverse dynamics.

I. INTRODUCTION

Muscular dystrophies (MDs) are characterized by progressive muscle wasting and weakness. Although the degree of decline and the severity of the conditions vary, MDs are generally disabling in time [1]. Most people with MDs eventually lose the ability to walk and their arm function is often impaired. People who are no longer able to walk can be helped with a wheelchair; however restoring a person's ability to perform daily tasks with the upper limbs remains a difficult problem to overcome.

Several studies have been conducted to investigate the influence of arm support systems in an attempt to restore the arm function by compensating for the weakened muscles [2-5]. An arm support system with braces can compensate for the gravitational pull by applying an upward force at local points on the arm [5]. Similarly exoskeletons can provide gravity compensation of both arm and forearm segments [6]. The activity patterns of shoulder and upper arm muscles during reaching movements using a forearm support system have been shown to be significantly reduced in healthy subjects [5]. However there is still little evidence on how joint moments are changed by a support system and whether zero gravity support is the best biomechanical solution that designers should aim for. Moreover although muscle activities have also been shown to be affected in stroke subjects using an arm support system [3-4], it is still unclear how joint moments are affected in an impaired person by a gravity free environment.

In this preliminary study, a comparison is made on the influence of gravity compensation between joint moments in healthy subjects and subjects with MD performing a reaching task. Biomechanical models and inverse dynamic software were used to calculate the shoulder and elbow joint moments in three different conditions; I) a control set-up, II) a gravity compensation set-up and III) a simulated zero gravity environment. The motivation behind this study is that lower moments at the shoulder and elbow can possibly benefit people with MD because the loadings on the weakened muscles are reduced. This research is part of a larger study whose objective is to quantify changes in joint moments in people with MD and to develop better support systems for this target population.

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II. METHODS

A. Subjects

Three healthy males (age: 30.7 ys ± 7.2; height: 1.80 m ± 2.8; weight: 75.7 kg ± 4.0) with no reported upper limb impairments and three MD male subjects (age: 54.7 ys ± 6.8; height: 1.80 m ± 6.2; weight: 79.7 kg ± 14) were measured. Ethical approval was obtained for the study (Radboud University Nijmegen Medical Centre Ethical Committee NL39024.091.11).

B. Movements

Motion analysis data from a reaching task were recorded in two different set-ups: unassisted movement and assisted movement with the SLING arm support (Focal Meditech). In both cases the subject was asked to move the dominant hand from an initial position resting on a table in the sagittal plane to a target placed at a distance of a stretched arm, at shoulder height and one shoulder width on the ipsilateral side. The tasks were performed while sitting on a chair in front of a table with the trunk upright. In the initial position the upper arm was slightly abducted alongside the upper body with a flexed and pronated elbow and a flattened hand touching the table. After the target was reached, the hand was returned to the starting position, completing the movement. The movements were recorded with a 3D camera system (Vicon Motion Systems) at a frequency of 200 Hz. Reflective markers were attached on the subject's body following the guidelines of the Upper Limb model [7]. Four repetitions were performed for each reaching movement of the two set-ups. The second repetition was used in the calculation of joint moments.

C. Arm Support

The SLING arm support mechanism was used for gravity compensation [8]. The device works using counterweights. In this study the SLING supported the lower arm at two points on the forearm at 33% and 67% of the ulnar length respectively, measured from the elbow to the wrist (Fig. 1). In this set-up the forearm was pulled upwards, while the subject had the freedom to perform movements in all three directions of space. The counterweight of the SLING was adjusted to compensate the gravitational pull, based on proprioceptive indications from the subject.

D. Model Environment

The coordinates of the reflective markers during the unsupported and supported movements described previously were used to drive the simulation model in the AnyBody Modeling System (AnyBody Technology). With the subject's biometric information derived from marker coordinates, the software's GaitFullBody model was scaled according to body length and mass among others. An inverse dynamic analysis was then carried out to calculate the net joint moments at the shoulder and elbow. The analysis on the unsupported movement consisted of two parts: a normal gravity situation, referred to as Control, and a simulated zero gravity situation, in which the same motion data for the unsupported movement were used but gravity was set to zero in AnyBody's model parameters. The analysis of the supported movement was carried out with location and direction of the upward force as specified in the arm support section. As a result the outputs of the calculation were the net joint moments in three conditions: I. Control, II. Gravity compensation with SLING, and III. Zero gravity environment. These conditions were chosen to assess the influence of gravity compensation (I vs. II), the influence of a zero gravity environment (I vs. III) and the difference between gravity compensation induced either by a mechanism or resulting from a zero gravity environment (II vs. III).

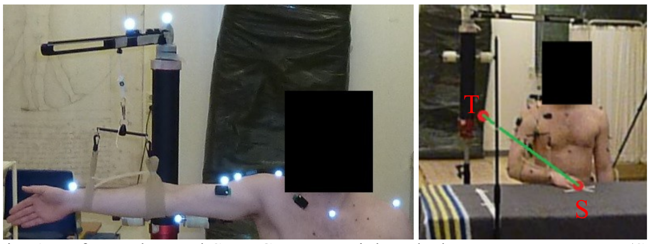

Fig. 1. Left: Marker and SLING set-up. Right: Ideal movement to target (S: start, T: target). The target was a plastic strip fastened to the black support pole and it is partly hidden in the figure.

E. Data Analysis

The measurements were normalized for time (0-100%) to account for inter- and intra-subject variation. The average joint moments were calculated per group and condition. The ratio of the SLING counterweight was calculated relatively to the BMI to account for possible variations in body compositions. Moreover, the time to complete the tasks was measured to investigate the speed per group. Data were summarized using descriptive statistics (mean and standard deviation). The conditions were compared within each group and within themselves.

III. RESULTS

The task was completed by all subjects in all conditions. The MD subjects required more time to complete the task in the Control and the SLING condition than the healthy group (Table 1). Both groups required more time to complete the task in the SLING condition than in the Control one. The healthy group used a relatively larger counterweight with respect to the BMI than the MD group (Table 1). In the Control condition the maximum value of the moment was greater by more than one order of magnitude than the moment in the SLING and the Zero gravity conditions in both groups (Fig. 2 and Table 2). Between the two groups the signs of the average moments in the SLING condition were different, showing for the MD group a trend to maintain the arm more elevated and the elbow more flexed. The healthy group presented a lower mean moment in the SLING condition than the MD group, showing a trend to maintain the arm less elevated and the elbow more extended when using the SLING.

TABLE I. SLING WEIGHT AND TIME TO PERFORM THE TASK

		Healthy Group	**MD Group**
SLING/BMI (kg/(kg/m2))		0.098 ± 0.002	0.079 ± 0.008
Time (s)	***Control***	2.62 ± 0.23	3.67 ± 0.38
	SLING	2.84 ± 0.20	4.69 ± 0.48
	Zero Gravity	2.62 ± 0.23	3.67 ± 0.38

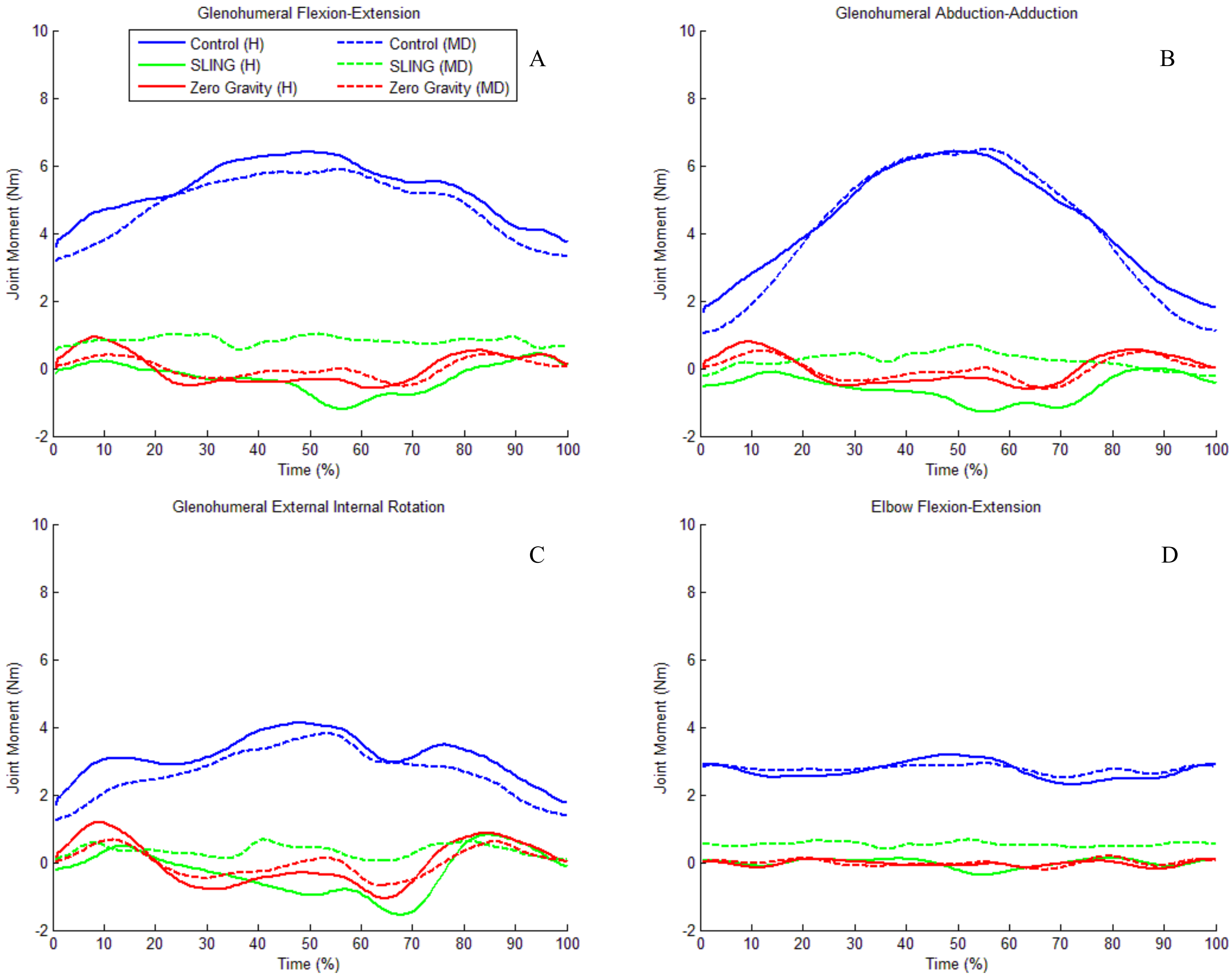

Fig. 2A-D. Glenohumeral and elbow joint moments for the two groups. Positive moments are Flexion, Abduction and External rotation.

The glenohumeral joint moments are shown in Fig. 2 ABC. The three conditions are shown for the healthy group (H) and the MD group (MD) per joint plane. The tasks started at 0% and ended at 100%. The target was reached at approximately 55% for the task with and without support of the SLING. At the beginning of the movement (0%) the elbow was flexed at about 90 degrees with the hand in front of the subject at table height. In the Control condition the shoulder flexion moment at 0% and 100% was lower because the arm was held in a posture with lower elevation. In the Zero gravity condition, the moments at 0% and 100% were almost negligible. When the arm was moved towards the target, the shoulder moments in the SLING condition were lower than in the Control as the arm was pulled upwards. For the Zero gravity condition, the arm needed to be accelerated and decelerated, resulting in a positive and negative shoulder moment. The arm was then moved back to the starting position, resulting in decreased shoulder moments and a minimum at 100% for the Control condition. In the SLING condition the healthy paticipants generated higher negative shoulder moments to move the SLING donwards in order to counteract the upward force of the mechanism. The MD subjects moved slower, using gravity to lower the arm. In the Zero gravity condition, the arm needed to be accelerated downwards and decelerated upwards, which resulted in the shoulder moments changing sign from negative to postive.

The elbow flexion moments for the three conditions are shown in Fig. 2D. The elbow pronation moment was constant and negligible in all cases (data not shown). In the Control condition the elbow flexion moment oscillated around about 3 Nm. This offset was caused by the constant flexion moment required to control the forearm from moving away from the upper arm. The moment did not decrease to 0 Nm at 0% and 100% since the hand was not supported by the table in the inverse dynamic model. This is a realistic approximation since at these points the table was only touched for a brief period of time. The moment patterns in the SLING condition and the Zero gravity condition fluctuated around offset values between 0 to 0.5 Nm, with changes resulting from the joint being either

TABLE II. AVERAGE JOINT MOMENTS AND SD PER GROUP AND CONDITION

	Healthy Group			**MD Group**		
	Control (Nm)	***SLING (Nm)***	***Zero Gravity (Nm)***	***Control (Nm)***	***SLING (Nm)***	***Zero Gravity (Nm)***
Glenohumeral Flexion (+) / Extension (-)	5.30 ± 0.33	-0.27 ± 0.17	0.00 ± 0.00	4.85 ± 1.16	0.82 ± 0.40	0.00 ± 0.01
Glenohumeral Abduction (+) / Adduction (-)	4.34 ± 0.47	-0.58 ± 0.04	-0.01 ± 0.01	4.14 ± 0.64	0.22 ± 0.33	-0.01 ± 0.00
Glenohumeral External (+) / Internal Rot. (-)	3.16 ± 0.43	-0.27 ± 0.07	0.00 ± 0.00	2.67 ± 0.33	0.34 ± 0.20	0.00 ± 0.00
Elbow Flexion (+) / Extension (-)	2.71 ± 0.16	-0.03 ± 0.07	-0.04 ± 0.01	2.76 ± 0.52	0.53 ± 0.20	-0.02 ± 0.02

flexed during reaching or extended when the hand was moved back to the table.

IV. DISCUSSION/CONCLUSION

The objective of this study was to investigate the influence of gravity compensation on the moments of the shoulder and elbow joint during a reaching task in order to understand if the SLING provides sufficient and adequate gravity compensation support. As a preliminary study, this experiment used three healthy subjects and three MD subjects. It was found that the gravity compensation mechanism not only lowers the joint moments in the shoulder and elbow but also alters the moments' patterns.

The moments at the shoulder and elbow joint in the unsupported conditions were comparable to the ranges reported in the literature for healthy subjects [9]. The moments in the SLING and in the Zero gravity conditions were more than 70% lower than in the Control condition. Hence, the supported condition would require less strength to perform the task than the unsupported one and in the long term it would also be less straining for the muscles. The mean joint moments were higher in the SLING condition than in the Zero gravity one. This finding is the result of the different way the subjects interacted with the support in the two conditions. In the SLING, as opposed to Zero gravity, the subjects received a constant upward force which had to be counteracted to move downwards. Although the SLING did not provide perfect zero gravity compensation support it did not lead to excessive compensatory moments greater than the original moments it was designed to compensate for. Comparable conclusions for the unsupported and supported conditions were found in studies on healthy elderly and stroke patients [4, 5]. These studies measured muscle activation of shoulder and elbow muscles to determine the difference between an unsupported and supported reaching task. The activations were found to be lower in the supported task. This indicates that the muscle groups were less active, and that less strength was needed to perform the task, which is equivalent to requiring a lower joint moment.

Some simplifications were made in the model, which constitute limitations to the present study. The distribution of the upward force between the two brace points of the SLING can change dynamically during the movement in a real life situation. A distribution shift would change the upward support moment and thus the resulting joint moments. This variation in the distribution of the support force was not accounted for in AnyBody. A possible solution to this problem is to measure the actual force on each brace point during the entire task.

An arm support system using counterweights is a cheaper solution than, for example, a robotic support but its passive nature also contribute to its limits. As it was shown for the healthy group in the supported SLING condition, shoulder moments had to be generated to counteract the upward force from the mechanism thus creating an additional shear load at the shoulder. This load was probably reduced in the MD group because the upward supporting force, relative to the BMI, was lower. A possible reason for the different supporting force between the two groups could be ascribed to the different perception of load and the preferences of the MD subjects during the set-up of the mechanism. Since the counterweight load was set on the basis of the user indication that the arm was feeling weightless, a less accurate perception of weight and a preference towards a lower value of the upward force would result in a lower weight, preventing the need to counteract a greater upward force thus reducing fatigue and shear loads on the joints. Such loads can compromise the stability of the shoulder in people with MD, particularly when the upper limbs are affected, as these subjects are known to present weakness in the rotator cuff muscles [1].

In addition, the MD group performed the tasks slower, which could possibly have influenced the quality of the movement and resulted in shoulder moments of a lower magnitude. From the above discussion it follows that the ideal application and distribution of the upward force between the brace points could be actively balanced to provide optimum support in the crucial phases of the movement where most effort is required to accelerate or decelerate the arm and to sustain the arm in general, while minimizing the shear loads on the joints. In this respect a redesign of the arm support system mechanism would be needed although this could compromise the simplicity of the device and increase its cost.

In conclusion designing better support systems should include biomechanical considerations. In this study it has been shown how the SLING lowers the moments at the shoulder and elbow joint, which can benefit subjects with MD in their daily tasks. However, further research should be focused on expanding the investigation on the influence of gravity compensation in MD subjects in order to quantify the real benefits of an arm support system.

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