Influence of gravity compensation on kinematics and muscle activation patterns during reach and retrieval in subjects with cervical spinal cord injury: An explorative study

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Abstract—Many interventions in upper-limb rehabilitation after cervical spinal cord injury (CSCI) use arm support (gravity compensation); however, its specific effects on kinematics and muscle activation characteristics in subjects with a CSCI are largely unknown. We conducted a cross-sectional explorative study to study these effects. Nine subjects with a CSCI performed two goal-directed arm movements (maximal reach, reach and retrieval) with and without gravity compensation. Angles at elbow and shoulder joints and muscle activation were measured and compared. Seven subjects reduced elbow extension (range 1.8°–4.5°) during the maximal reaching task with gravity compensation. In the reach and retrieval task with gravity compensation, all subjects decreased elbow extension (range 0.1°–11.0°). Eight subjects executed movement closer to the body. Regarding muscle activation, gravity compensation did not influence timing; however, the amplitude of activation decreased, especially in antigravity muscles, namely mean change +/– standard deviation of descending part of trapezius (18.2% +/– 37.5%), anterior part of deltoid (37.7% +/– 16.7%), posterior part of deltoid (32.0% +/– 13.9%), and long head biceps (49.6% +/– 20.0%). Clinical implications for the use of gravity compensation in rehabilitation (during activities of daily living or exercise therapy) should be further investigated with a larger population.

Key words: electromyography, goal-directed movements, gravity compensation, kinematics, rehabilitation, robot-assisted therapy, robotics, spinal cord injury, tetraplegia, upper limb.

INTRODUCTION

Damage to the spinal cord causes loss of motor and sensory function of the body parts below the level of the lesion. In patients with a cervical spinal cord injury (CSCI), the arm and hand function is affected to varying degrees according to the level and completeness of the lesion [1]. Compared with other spinal cord injury-related impairments, improvement in upper-limb function is one of the highest priorities in patients with a CSCI [2]. Exercise therapy integrated in an intensive rehabilitation program to learn or relearn motor functions is considered very important in optimizing the remaining upper-limb function [1,3]. Even in the chronic stage, intensive exercise therapy positively affects upper-limb motor control and functional abilities in patients with a CSCI [4].

Abbreviations: 3-D = three dimensional, ADL = activity of daily living, CSCI = cervical spinal cord injury, MRC = Medical Research Council, sEMG = surface electromyography.

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Literature indicates that motor learning or relearning is influenced by several key elements: active movements, intensity of practice (frequency, repetitions, and duration), use of feedback, task specificity, goal-orientated practice, and variation [5–6]. Exercise therapy based on these motor learning or relearning principles asks great physical effort from patients with a CSCI who have impaired upper-limb function. We presume that during goal-directed movements, a large part of the preserved muscle force is required to hold the arm against gravity; consequently, less muscle force is available to perform the actual movements.

To facilitate goal-directed arm movements during activities of daily living (ADLs) [7] or exercise therapy, therapeutic devices are often used to support the weight of the arm (e.g., with the Swedish Help Arm [Kinsman Enterprises, Inc; West Frankfort, Illinois]). In the last decade, several innovative therapeutic devices, including robotics, have been developed to support the affected upper limb during exercise therapy [8–9]. In these robotic devices, different treatment modalities have been implemented, such as passive, active-assisted, and active-resisted movements [8]; consequently, gravity compensation is incorporated in the design [8–9]. Until now, the effect of gravity compensation on motor control and functional abilities has mainly been investigated in nondisabled elderly [10] and stroke patients [11–15]. Although many applications in rehabilitation after a spinal cord injury include gravity compensation during ADLs or exercise therapy, the specific effects on kinematics and muscle activation characteristics (amplitude and timing) in patients with a CSCI are largely unknown. A cross-sectional explorative study that measured kinematics and surface electromyography (sEMG) during goal-directed movements with and without gravity compensation was conducted to study the effects of gravity compensation in subjects with a CSCI.

METHODS

Subjects

Nine subjects with a CSCI (at least 1 year since injury) were recruited from a local rehabilitation center. Inclusion criteria for participation were motor injury level C5–C7 (cervical) and age between 18 and 65 years. Exclusion criteria were extreme shoulder pain, contractions of the upper limb, and/or spasticity preventing performance of the required tasks. All subjects were assessed according to the standard neurological classification [16].

Apparatus

A mechanical, passive device called Freebal [17] (University of Twente, Enschede, the Netherlands; now available commercially as ArmeoBoon, Hocoma; Volketswil, Switzerland) was used to counteract the effect of gravity on the upper limb (Figure 1). The device has two slings—one is applied at the elbow and the other around
the wrist. Each sling is connected to an independent adjustable spring by way of an overhead cable and pulleys. During the goal-directed movements, this system enables a constant amount of support throughout the three-dimensional (3-D) working volume, irrespective of the position and orientation of the arm [17].

**Procedures**

During the measurements, subjects sat in their own wheelchairs (one subject was not wheelchair-dependent and sat on a normal chair) in front of a height adjustable table. In the starting position, subjects sat with their forearm flat on the tabletop, elbow flexed at 90°, and hand on the starting dot. Subjects performed two goal-directed movements with and without the Freebal:

1. Maximal reaching task. This task consisted of three maximum reaches in front of the subjects, without gliding the hand and arm along the tabletop.
2. Reach and retrieval task. Subjects were instructed to move at their own comfortable speed between a starting dot and target dot on the table for 30 seconds. Both dots were 10 cm in diameter, and the distance between the dots was 35 cm (Figure 2(a)).

**Measurement and Data Analysis**

**Kinematics**

Kinematics were recorded with a 3-D optical movement tracking system with six cameras (Vicon Nexus 1.3.109, Oxford Metrics Ltd; Oxford, United Kingdom). Reflective markers were placed on 10 bony landmarks of the arm and trunk: processus spinosus of the seventh cervical and eighth thoracic vertebra, incisura jugularis, processus xiphoideus, acromioclavicular joint, medial and lateral epicondyle, radial and ulnar styloid, and distal head third metacarpal (Figure 2(b)). Six cameras at 100 Hz recorded the 3-D marker trajectories. The acromion marker was used for estimating the glenohumeral rotation center. Scapular motion was disregarded because scapular motion was not likely to participate in the anteflexion movement if the angle of elevation remains below 60°.

The marker trajectories were visually inspected for recording errors and missing marker data. If one trunk marker was missing, we replaced it using the Vicon BodyBuilder model (Metrics Ltd; Oxford, United Kingdom). This model estimated the position of the missing marker by the position of the other three markers. We replaced missing marker trajectories over a short period (less than 10 samples) by linear interpolation. If data were missing for longer periods or at the end of the reach or retrieval movement, the movement cycle was removed.

Marker position data were converted to limb segments data according to the guidelines of the International Society of Biomechanics [18]; thereafter, joint angles were calculated with Euler rotation. The elbow joint angle (Figure 3(a)) was specified as the angle between the longitudinal axis of the upper arm and the forearm (full elbow extension was defined as 0°; forearm perpendicular to upper arm, 90°). We calculated two angles to describe the position of the upper arm related to the thorax: (1) the angle of elevation (Figure 3(b)),
defined as the angle between the upper arm and trunk (upper arm parallel with thorax, 0°; upper arm parallel with horizontal; 90°), and (2) the plane of elevation (Figure 3(c)), defined as the angle between the thorax and the upper arm in the transversal plane (arm extended forward, 0°; arm extended to the lateral, –90°).

For the maximal reaching task, we compared the maximum elbow extension with and without gravity compensation. To quantify the differences between the reach and retrieval task with and without gravity compensation, we derived joint rotations (in degrees) of the angles just mentioned and parameters of the movement cycles (mean duration of one movement cycle, number of repetitions within 30 seconds). Cycle parameters were averaged over all movement cycles within a series; the first two cycles were excluded for analysis. A movement cycle consisted of two parts, namely reach (maximum to minimum elbow angle) and retrieval (minimum to maximum elbow angle).

Electromyography

Bipolar sEMG of eight superficial muscles (descending parts of the trapezius, anterior and posterior parts of the deltoid, pectoralis major, long head of the biceps, long head and lateral head of the triceps, and latissimus dorsi) was recorded with circular, wet gel, silver/silver-chloride electrodes (ARBO, type S93SG, Tyco/Healthcare Deutschland; Neustadt/Donau, Germany) at a sample frequency of 1,000 Hz. Electrode placement, skin preparation, and recording protocol were in accordance with the SENIAM guidelines [19].

sEMG signals were synchronized with the marker trajectories (Figure 4). The time axis was normalized from 0 to 100 percent: reach 0 to 50 percent and retrieval 51 to 100 percent.

We converted the band-pass filtered sEMG signals to smooth rectified sEMG using a second-order Butterworth filter with frequency at 25 Hz. To visualize the differences in smooth rectified sEMG, we plotted movement trajectories (averaged data over all cycles) for two subjects with and without gravity compensation plotted in the same graph (Figure 5). Changes in the amplitude of muscle activation during movements with gravity compensation were expressed as a percentage of the change of the area under the curve of the same movement without gravity compensation. The area under the curve is calculated as the integral of the smooth rectified sEMG.

Timing of muscle activation was analyzed visually. The primary investigator assessed the sEMG recordings, and a coauthor with extensive experience in sEMG analysis checked it.

Statistical Analysis

This study had an explorative character; therefore, the effect of gravity compensation was described separately for each individual subject. Because of the small sample size and a heterogeneous population, a Wilcoxon signed rank test was performed and the median or ranges were
Figure 4. Elbow and shoulder joint angles (°) during 15 s repetitive reach and retrieval tasks with Freebal, performed by subject with identification number 2, simultaneously displayed with smooth rectified surface electromyography values (microvolt) of eight measured muscles.
found. From the Wilcoxon test, the test statistic $T$ (smallest of the two sums of ranks), its significance ($p$), and the effect size ($r$) were reported.

RESULTS

Subjects
A complete data set was available for nine participants. The physical characteristics of each of the nine subjects are displayed in Table 1.

Kinematics
Movement parameters are presented in Table 2. During the maximal reaching task with and without gravity compensation, the maximum elbow angle was significantly lower with gravity compensation (median 33.3°) than without gravity compensation (median 29.4°), $T = 2$, $p = 0.021$, $r = -0.77$.

During the reach and retrieval task with gravity compensation, all subjects showed decreased elbow extension (range 0.1°–11.0°). At the shoulder joint, seven subjects
had decreased plane (0.3°–6.9°) and six subjects had reduced angle of elevation (0.1°–15.1°). The movement times increased in four subjects (range 0.1–0.4 s), decreased in two subjects (0.2–0.4 s), and remained the same in three subjects. None of these parameters differs significantly between movements performed with and without gravity compensation (elbow extension: $T = 2, p = 0.214, r = –0.41$; shoulder plane of elevation: $T = 3, p = 0.767, r = –0.10$; shoulder angle of elevation: $T = 4, p = 0.515, r = –0.22$; and cycle duration: $T = 2, p = 0.484, r = –0.23$).

**Electromyography**

Based on the plotted smooth rectified sEMG (Figure 5) and calculated differences (in terms of percentage) in the areas under the curves (Table 2), we made three observations:

1. With gravity compensation, the amplitude of the sEMG decreased especially in the antigravity muscles. In six subjects, amplitude of the sEMG decreased in the descending part of the trapezius (range 17.5%–60.6%) and increased in three subjects (4.1%, 6.5%, and 59.7%). In all subjects, amplitude of the sEMG was decreased in the posterior part of deltoid (range: 12.8%–54.1%), the anterior part of deltoid (17.4%–73.6%), and the long head of biceps (22.9%–80.0%).

2. In four subjects (identification numbers 1, 3, 7, and 8) without triceps activity (Medical Research Council [MRC] score of 0), sEMG activity was recorded during flexion of the elbow.

3. In three of the five subjects with active triceps function (MRC score of at least 2), the amplitude of sEMG in the long head of triceps increased (25.2%, 1.2%, and 16.9%) and decreased in the other two subjects (16.4% and 56.6%). On a group level, a significant difference between the conditions with and without gravity compensation was found for the following muscles: descending part of trapezius during reach: $T = 1, p = 0.038, r = –0.69$; posterior part of deltoid during reach: $T = 1, p = 0.015, r = –0.81$, and during retrieval: $T = 0, p = 0.008, r = –0.89$; and anterior part of deltoid and long head biceps for reach as well as retrieval: $T = 0, p = 0.008, r = –0.89$.

Within subjects, the timing of muscle activation did not change visibly with gravity compensation. With respect to the patterns of timing between subjects, we found various different patterns. Some alternating activation patterns were found between agonists and antagonists. All subjects with at least some triceps function showed an alternating activation pattern between the long head of biceps and triceps (Figure 6(a)). We found a simultaneous activation pattern in four subjects between the activation of the anterior and posterior parts of the deltoid muscle (Figure 6(b)) and in six subjects between the anterior part of deltoid and pectoral muscles (Figure 6(c)).

Furthermore, the descending part of the trapezius was used in various different patterns. In one subject, an alternating activation pattern between the anterior and posterior parts of the deltoid occurred, and in another subject, an alternating activation pattern between the

### Table 1.
Physical characteristics of subjects ($N = 9$).

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ASIA = American Spinal Injury Association, C = cervical (fifth to eighth vertebra), F = female, ID = identification, L = left, M = male, MRC = Medical Research Council, R = right, T = thoracic.
posterior parts of the deltoid solely. These two combinations were also observed in a simultaneous pattern: in one subject, the descending part of the trapezius and anterior and posterior parts of the deltoid were simultaneously activated, and in another subject, the descending part of the trapezius was activated with the posterior deltoid solely.

**DISCUSSION**

The objective of the present study was to study the influence of gravity compensation on kinematics and sEMG characteristics of the upper limb during goal-directed movements of subjects with a CSCI.

With gravity compensation, most of the subjects showed less elbow extension and movement execution closer to the midline. Based on previous studies with stroke patients, one can expect that gravity compensation increases range of motion of the upper limb [12,14] because of the positive effect on pathological muscle synergies between shoulder abduction and elbow flexion [14]. In patients with a CSCI, this pathological coupling does not occur. However, an effect on kinematics is expected because less muscle force is necessary to overcome
results of this study showed less elbow extension during maximal reaching with gravity compensation in seven subjects. During reach and retrieval with gravity compensation, all subjects showed less elbow extension and, in eight subjects, a decrease in shoulder angle and/or plane of elevation.

Plausible explanations could be given for these results. First, subjects with a CSCI who have a lack of triceps function use their anterior part of the deltoid and upper pectoral muscles to produce an isometric extension torque in their elbow [20] or make a trick movement with their shoulder muscles to achieve passive elbow extension [21]. They use gravity to maintain the arm in extension below the horizontal plane [22] and to perform a passive elbow extension with a trick movement. In both compensation strategies, gravity is used to maintain elbow extension. Therefore, movement execution with gravity compensation might decrease elbow extension. Second, during goal-directed movements without gravity compensation, subjects use a large part of the preserved muscle force to hold the arm against gravity. If the primary agonists alone are not capable of generating the required anteflexion and extension torques, additional agonist muscles are recruited [23]. For example, the middle part of the deltoid might contribute to lift the arm, if the anterior part of the deltoid cannot generate enough force. The middle part of the deltoid also has an abduction function that can result in a reaching movement not truly in the sagittal plane [23]. Third, because of a decreased plane of elevation, the hand moves more in a direct line to the target dot. If the arm is extended closer to the midline, less elbow extension and angle of elevation are necessary to reach the target dot. Finally, with gravity compensation, the pectoral muscles can move the arm more easily to a position in front of the patient because the weight of the arm is counteracted.

The results of the sEMG data during the reach and retrieval task showed a decrease in sEMG activity during movements with the use of the Freebal, particularly in muscles that counteracted gravity, while timing remained unaffected. The results confirmed our presumption based on previous studies with nondisabled elderly [10] and stroke patients [11,13,15] that also showed a decreased sEMG in antigravity muscles and unaffected timing. Remarkably, despite subjects with an MRC score of 0 in the triceps, sEMG activity was seen mainly during elbow flexion. A plausible explanation for this sEMG activity is stretch or cocontraction. In the sEMG signal, however,
one cannot differentiate between activity because of stretch and voluntary motor activity [24].

A large variety in muscle activation patterns was seen between subjects because of heterogeneity of the study population. After a CSCI, the functional anatomy of the upper limb had to be redefined. Muscle synergies as seen in nondisabled subjects are often inappropriate for subjects with a CSCI [20]. The central nervous system is challenged to use a motor strategy to adjust to the new functional anatomy and biomechanics, with a reduced repertoire of innervated muscles to deal with the mechanics [21], leading to different movement patterns between subjects with a CSCI [20].

To our knowledge, our study was the first explorative study about the effect of gravity compensation on kinematics and sEMG in subjects with a CSCI. Another type of arm support by subjects with a CSCI was studied by Atkins et al. [7]. They reported about the effect of mobile arm support on ADLs. Based on Delphi questionnaires, they concluded that some ADLs were possible with the use of a mobile arm support, which without the use of such a device, patients with very weak biceps and deltoid muscles were unable to perform.

Besides being used for compensating lost functions, gravity compensation can be used for training purposes. Further studies should be performed with a larger population because of the small effect size, especially on kinematic parameters, and should be able to test the following hypotheses: (1) patients with an MRC score of at least 2 in the triceps muscle can train their primary agonists of the shoulder and elbow in goal-directed movements more intensively and, (2) for patients without active triceps function (MRC score of 0 or 1), gravity compensation may not seem useful to train extension movements because they perform these movements with the use of gravity. However, gravity compensation might be beneficial for training muscles required to cross the midline or to perform bimanual tasks. Also, the influence of gravity compensation on the patients’ ability to stabilize the shoulder in a certain position would be an interesting parameter.

CONCLUSIONS

This explorative study showed that gravity compensation influenced the kinematics and amplitude of the sEMG of the upper limb during goal-directed movements in CSCI. A larger study is needed to firmly conclude whether training with gravity compensation is clinically relevent.

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Study concept and design: G. J. Snoek, M. Kouwenhoven, A. V. Nene, M. J. A. Jannink.
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Interpretation of data: M. G. M. Kloosterman, G. J. Snoek, M. Kouwenhoven, A. V. Nene, M. J. A. Jannink.
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REFERENCES


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