Balance and stabilization capability of paraplegic wheelchair athletes

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Abstract—The orientation of paraplegic athletes toward adapted sport activities requires good knowledge of their functional characteristics. Wheelchair locomotion, especially for highly dynamic situations, poses the problem of trunk equilibrium management and head stabilization. The study aimed at designing a quantitative method to assess the ability of paraplegics to obtain trunk balance under dynamic stresses, and to analyze the various balance strategies, according to the spinal lesion level of the subjects. High (HPA) and low (LPA) paraplegic athletes were subjected to four series of antero-posterior stresses of increasing intensity, generated by an oscillating platform. By means of a computerized video-based movement analyzer, acceleration in the sagittal plane was measured at four different spinal levels and, for each one, a damping factor was determined. This factor, computed at the head level, appeared to be representative of the subjects’ ability to maintain balance. We attempted to differentiate balance strategies in the LPA and HPA groups through analysis of the relative contributions to damping of the thoracic and cervical spinal segments. The first results show an increasing tendency of neck reflex stiffening according to the neurological level.

Key words: athletes, balance, damping, paraplegic, postural, wheelchair.

INTRODUCTION

High performance in wheelchair locomotion requires a combination of low energy cost and optimal comfort. Handrim wheelchair propulsion is a means of locomotion with quite a high exertion demand. Weak propulsive output is obtained (1–4) with relatively high cardiorespiratory stress (5–7). Thorough studies of metabolic and physiological responses to muscular exercise (8–10) and of propulsion techniques (11–13), as well as investigations on wheelchair designs (14–16), have all contributed to reaching the optimal “man-machine” interaction objective.

Due to the increasing interest in wheelchair athletics, it is now critical to improve the man-machine interface, which involves adapting available standard equipment for potential athletic use. This approach takes the functional behavior of wheelchair athletes into consideration and emphasizes prevention of injuries related to handrim wheelchair propulsion (17,18). Head movement and mobilization of the cervical and thoracic spine play major roles during propulsion.

For wheelchair athletes, these spinal areas are crucial for efficient ambulation and trunk balance, thus increasing the risk of musculoskeletal trauma. Various authors have described spine traumatisms and injuries related to wheelchair propulsion and disabilities (19–22), others have carried out kinematic analyses of wheelchair locomotion and related spine movements (23–25). All of these studies have enhanced the overall understanding of spinal behavior in wheelchair users.

Prevention of traumatic risks for wheelchair athletes implies definition of suitably adapted physical activities, which requires precise knowledge of individual real functional potential. Our work falls within the scope of the functional characterization of wheelchair athletes (26,27). Performance analysis of the body balance regulation sys-
tem in the sitting position is a basic parameter for functional evaluations. The aim of the present study was thus to define balance control capabilities of paraplegic wheelchair athletes with different levels of neurological lesion.

METHODS

Principles

In order to study characteristics of the balance system, wheelchair athletes with paraplegia were subjected to acceleration in the sagittal plane. This acceleration component represents a major traumatic risk for the spine.

Identification of a control system generally consists of analyzing the response of that system to well defined input signals, mainly in the form of square, triangular, or sinusoidal waves. The first one leads to excessive levels of acceleration (shock) and is thus unsuitable for application to persons with paraplegia. In addition, it is technically difficult to implement. Triangular input, which produces (only at the slope inversion) short and powerful acceleration changes, has also been rejected.

Sinusoidal input therefore appeared to be most appropriate in our experimental context. It generates repetitive, alternate, and continuously variable acceleration with time, which eliminates the risk of abrupt stress at controlled levels of frequency and amplitude. Moreover, this type of input has some similarities to real cyclic movements of the trunk of the wheelchair user.

Acceleration transmitted to the head was measured to determine the subjects’ balance performances. To define the strategies of the wheelchair users, the relative contribution to balance of cervical and thoracic spines was also evaluated by measuring acceleration at various sites along the spinal column.

Equipment

The study required an original device to move the wheelchair and a computerized video-based movement analyzer for kinematic analysis. Technical difficulties to implement horizontal translational alternate displacements of the wheelchair led us to design the simple moving platform shown in Figure 1. This platform oscillates on a horizontal axis (A). It is activated by a servo-controlled hydraulic jack (B) linked to a hydraulic power supply (C). Control is maintained with a function generator (D), enabling us to subject the entire man-machine system to sinusoidal oscillations in the sagittal plane. Due to the relatively low position of the platform rotation axis, its distance from the various spinal measurement sites, and, as described later, the reduced angle rotation range, the main acceleration parameter taken into account was the horizontal (antero-posterior) component. This corresponds to dynamic conditions close to those obtained with a true antero-posterior translation movement generator.

The computerized video-based movement analyzer was an ELITE system, equipped with two video cameras. The movements of passive markers fixed on the subjects were recorded, while processing the tridimensional kinematic data. The system represents the functional pattern in harmonious groups of points bound by segments. The tracking procedure and kinematic analysis over a time-course provided us with data on linear and/or angular displacements, velocity and acceleration, and reconstruction of the trajectories of an unlimited number of markers. The sampling frequency was 100 Hz.

Procedure

The subject sat in a wheelchair fixed to the oscillating platform. He was fitted with four reflecting markers placed along the articular truncal axis as shown in Figure 2: at the temporal bone (M1), the superior head rotation center (M2), the inferior head rotation center (M3), and the iliac crest (M4). The two markers at the head rotation centers were placed according to the model of Berthoz (1983). Two other markers completed this model, the first on the rotation axis of the platform (M6) and the second (M5) at a vertical distance (h0) of 20 cm from this axis.

The extent of acceleration transmitted to the subject depends on three main parameters: angular amplitude (a),
Parameter values were carefully set to meet with the following conditions: 1) to produce, at the head level, a main acceleration component in the antero-posterior direction, which requires relatively low platform rotation angle amplitude and 2) to subject the head to various acceleration levels corresponding to those reached in usual daily life situations, up to dynamic levels encountered in sport activities, while still remaining well below risky levels.

Four oscillation levels A, B, C, and D were thus defined:

A. $a = \pm 3.5^\circ$ and $f = 1$ Hz which, for $h = 1$ meter (reference value), corresponds to a maximum acceleration of $2.4 \text{ ms}^{-2}$
B. $a = \pm 3.5^\circ$ and $f = 1.25$ Hz, maximum acceleration $3.8 \text{ ms}^{-2}$
C. $a = \pm 5.5^\circ$ and $f = 1$ Hz, maximum acceleration $3.8 \text{ ms}^{-2}$
D. $a = \pm 5.5^\circ$ and $f = 1.25$ Hz, maximum acceleration $6 \text{ ms}^{-2}$.

Despite different amplitude and frequency values, oscillation levels B and C led to the same acceleration, which allowed us to estimate the relative effects of these parameters on balance control.

For each stress level and at each marker, the maximal amplitude of acceleration was measured with a computerized optoelectronic movement analyzer. The following postural conditions were required of the subjects: hands on handrim, feet on the footrest, back leaning on the backrest, and visual axis oriented horizontally forward. Each test lasted 15 seconds, and the kinematic data were only recorded during the last 5 seconds.

**Subjects**

Two groups of six paraplegic subjects were selected (Table I): one group was composed of high paraplegic athletes (HPA), with neurological levels between T4 and T8, and one group was composed of low paraplegic athletes (LPA), with neurological levels between T11 and L5.

The subjects were all members of the Fédération Française Handisport and the Montpellier Club Handisport, and did not have any spinal immobilization by mechanical means or bony fusion.

A group of six normal healthy athletes (NHA) was also used to provide us with a functional reference, and to assess variability in the normal balance process.

**Damping Factor Definition**

The maximum acceleration value as computed above corresponds to a theoretical situation where the marker, placed at a distance ($h$) on the rotation axis of the platform, is supposed to be attached to this axis by an infinitely rigid link.

Physiological reality is of course (and fortunately) quite different from this rigid model. Active and passive
Table 1:
Anthropometric and functional characteristics of the subjects.

<table>
<thead>
<tr>
<th>Subj</th>
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<th>wgt kg</th>
<th>hgt cm</th>
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</table>

*HPA = high paraplegic athletes; LPA = low paraplegic athletes; NHA = normal healthy athletes; SCI lev = spinal cord injury level; M = mean; SD = standard deviation; compl = complete; inj = injury*

The damping effects of the musculoskeletal components contribute to minimize acceleration at the head.

A damping factor (d) was determined to assess the subject’s ability to maintain the head at the lowest level of acceleration. This factor is defined as the ratio of the real measured acceleration \( A_{\text{meas}} \) at a given site, to the theoretical acceleration \( A_{\text{th}} \) calculated at this site, assuming that the system is infinitely rigid:

\[
d = \frac{A_{\text{meas}}}{A_{\text{th}}}
\]

At a given point, the theoretical acceleration is proportional to the distance \( h \) to rotation axis. Consequently, if \( A_0 \) is the measured acceleration at a fixed reference point, situated at a distance \( h_0 \) from the axis of the oscillating platform, the theoretical acceleration \( A_{\text{th}} \) at any given point of the body, situated at distance \( h \) from the axis, can be calculated as:

\[
A_{\text{th}} = A_0 \frac{h}{h_0}
\]

Theoretically the value of the damping factor is between 0 and 1. The closer its value is to 0, the higher the damping effect at the considered site.

Moreover, the individual balance strategies can be characterized by assessing the relative contribution to damping of the cervical and thoracic spine. This can be achieved by calculating the difference between the values of the damping factor at the extremities of each spinal segment.

If \( A_3 \) is acceleration at the inferior head rotation center and \( A_4 \) acceleration measured at the iliac crest, the corresponding damping factors are:
The contribution to damping $C_t$ of the thoracic spine is:

$$C_t = d_3 - d_4 = \frac{h_0}{A_0}(A_3/h_3 - A_4/h_4)$$

$C_t$, which is proportional to the difference of normalized accelerations $A_3/h_3$ and $A_4/h_4$ measured at each extremity of the thoracic segment, is fully representative of the damping effect of the considered spinal part. In the same way, we evaluated the contribution $C_c$ to damping of the cervical spine $C_c = d_2 - d_3$. The damping factor was calculated for each oscillation level A, B, C, and D.

**Data Analysis**

A mean comparison test was selected to investigate the relationship between damping factor values and the extent of medular lesion and to define the relative contributions of the cervical and thoracic spine for maintenance of balance. The data were presented as the means ± standard deviation, with a significance level of $p < 0.05$.

**RESULTS**

**Damping Factor at the Head**

Statistical analysis of the values of the damping factor at the head did not lead to significant results. Nevertheless, graphical representation of values of the damping factor at the head (M1 marker) for oscillation levels A, B, C, and D and for HPA, LPA, and NHA groups (Figure 3), showed (except for very low oscillation levels), a marked decrease according to the neurological level. Moreover, this factor appears globally decreasing with the intensity of oscillation level. For the LPA and NHA subjects, the decrease in the damping factor was slightly perceptible in oscillation levels A, B, and C; whereas, it was more marked and regular in the HPA group under all conditions.

**Cervical and Thoracic Spine Involvement in Balance**

As shown in Figure 4, the damping factor decreased from iliac crest to the head. Nevertheless, the decreasing mode markedly differed according to the group considered. LPA and NHA showed a steadily progressive decrease at all oscillation levels, while there was a sharp decrease in the damping factor in the HPA group between M4 and M3, even at the lowest level.

From the damping factor values along the articular axis of the spine, we determined the contribution of the

![Figure 3](#)

Damping factor at the head (M1) during forward and backward oscillations for oscillation levels A, B, C, and D for high paraplegic athletes (HPA), low paraplegic athletes (LPA), and normal healthy athletes (NHA).
cervical and thoracic spine to balance. Regarding the contribution of the thoracic spine during forward oscillations (Table 2a), we observed significant differences between HPA and NHA for oscillation level A (p < 0.05). There were no marked differences between the three groups during the backward oscillation phase.

Regarding the contribution of the cervical spine to balance during the forward oscillation phase (Table 2b), we observed significant differences between HPA and LPA for level B, between LPA and NHA for level C, and more generally, between HPA and NHA for levels A, B, C, and D (0.01 < p < 0.05). Regarding the contribution of the cervical spine to balance during backward oscillations, we observed significant differences between HPA and LPA for level C, between LPA and NHA for level A, and between HPA and NHA for levels B, C, and D.

DISCUSSION

Variations in the Damping Factor at the Head

On the whole, we observed that the damping factor values at the head decreased with the intensity of stress. This decrease was barely perceptible for NHA and LPA under moderate stress (oscillation levels A, B, and C). Subjects of the two groups with sufficient or normal physical capabilities did not have any particular difficulties in controlling balance. During oscillation levels B and C, nearly identical damping factor values were obtained at the head. This is normal considering, as before, that accelerations applied in these two tests were identical, despite the different amplitude and frequency values.

The behavior of HPA differed markedly, with an increase in the damping effect for the higher mechanical excitation amplitudes. Moreover, in this group we noted a regular decrease in the damping factor with increasing intensity of oscillation. The relatively high damping factor value (0.50) during oscillation level A could be explained by the subject’s search for the best strategy to obtain the most efficient stabilization of the head. For more intense stress (oscillation levels B, C, and D), there was a relation between the damping factor value and the neurological level of the subject.

We also observed that in all conditions, normal subjects had a damping factor value higher than that of the two groups of paraplegic subjects. This could reflect a reduced...
need for active damping of normal subjects who had integral sensori-motor potential. This was confirmed by particularly high damping factor values (0.6) in the low stress oscillation levels A, B, and C. It can be assumed that for HPA and LPA, the higher the neurological level, the more sensitive the subjects were to mechanical perturbations. Consequently, they tried to minimize the effect of the mechanical perturbation by minimizing the amplitude of the head acceleration. On the contrary, the nondisabled athletes showed less intense reactions to mechanical perturbations and were less sensitive to the conditions. All of the above observations were valid for the forward and backward stresses. Nevertheless, the damping factor value was noticeably higher during backward stresses. This phenomenon may be related to the decrease in the number of degrees of freedom of the thoracic and lumbar articular spine during the contact phase of the back with the backrest of the wheelchair. In the next section, we analyze the involved articular and muscular structures in order to determine a satisfactory damping effect at the head.

**Balance Strategy: Thoracic and Cervical Spine Involvement**

During forward stress, the comparison between HPA and NHA behavior in all tests showed significant differences of the contribution of the cervical spine to balance.
Table 2b:
Contribution to damping of the cervical spine by oscillation level.

<table>
<thead>
<tr>
<th>Oscillation</th>
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</tr>
<tr>
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HPA = high paraplegic athletes; LPA = low paraplegic athletes; NHA = normal healthy athletes; M = mean; SD = standard deviation; t = t value; p = significance

Nevertheless, a noticeable difference (p < 0.05) was obtained for the B oscillation level (favoring frequency of oscillations) between HPA and LPA groups.

Concerning the contribution of the thoracic spine, the only significant difference appeared between HPA and NHA during oscillation level A. During backward stress, we observed the same type of results as for forward stress. There were significant differences between HPA and NHA during oscillation levels B, C, and D. However, there was a marked difference (p < 0.05) between HPA and LPA for the oscillation level C (favoring amplitude of oscillation). No significant difference was observed between the three groups of subjects for thoracic spine involvement in balance.

Based on the obtained data, the following characteristic behaviors for subjects of the three groups can be presented:

- The involvement of the thoracic spine in HPA and LPA increased with the intensity of stress and was markedly higher than that of NHA for both forward and backward stresses.
- The involvement of the cervical spine in balance in HPA was zero or very weak during both forward and backward stresses.
- All subjects showed a tendency to fix the cervical spine when the stress increased.
The neck stiffening tendency and consequent decrease in the mobility of the cervical spine clearly appeared in HPA and, to a lesser degree, in LPA. This stiffness was probably obtained by intense muscular action, generating fatigue and considerable mechanical articular constraints at this level. Similarly, spinal immobilization from orthotic devices or vertebral fusion, also causing spine rigidification, might have direct consequences on the transmission of acceleration. It would be interesting to test the method on a population of subjects with spinal immobilization at various levels.

In addition, it appears that the neurological level was not completely related to the damping factor values and to the balance strategies of the paraplegic subjects. Indeed, we observed less stiffness for two subjects of the thoraco-neurological level group than for one subject of the lumbar neurological level group. In contrast to other wheelchair athlete assessments, while we observed a close correlation between neurological level and biological responses (8,10), the present study demonstrates that balance strategies are more difficult to characterize.

We consider that the orientation and training of paraplegic and tetraplegic wheelchair athletes will require association of different functional evaluation parameters. Adaptation of physical activities to the capacities of the subjects demands more than knowledge of the neurological level. A multiparametric approach to athletic behavior, with this kind of balance evaluation, offers a means to optimize capacities while preventing pain and traumatism related to sport activities.

**CONCLUSION**

In spite of the small subject sample size, this study reveals the balance capabilities of paraplegic athletes. It also offers a means of analyzing their behavior under well-defined mechanical conditions. There seems to be a relationship between the damping factor measured at the head and the subject’s neurological level, which indicates the importance of this parameter. This factor could be a relevant quantitative indicator for assessing the ability of the paraplegic subject to obtain efficient body balance in the sitting position. For instance, this factor might be taken into consideration before orienting subjects toward particular sport activities requiring good control of sitting postures under highly dynamic stress. The damping factor could also be used in wheelchair design to evaluate the effect of given fittings on balance.

Balance strategies can be described through analysis of the relative contribution to balance of the cervical and thoracic spine. There seemed to be a relationship between the degree of stiffening of the cervical spine and the subject’s neurological level. Cervical spine stiffening, generally obtained by violent contractions of the neck muscles, could be the cause of spinal impairment. It thus seems reasonable to assume that it is better to be among those with less neck stiffening.

The proposed method seems effective in quantifying individual postural balance and estimating the risk of practicing particular sport activities. Nevertheless, considering the small number of athletes tested, the complexity, and thus the variability of the balance process, some differences in subject behavior were not clearly seen. Furthermore, the observed lack of significant differences according to neurological levels indicates that knowledge of this clinical factor alone cannot determine real individual ability to maintain balance, and points out the need for complementary quantitative assessment techniques. More precise characterization of the balance strategies will require further research on a wider sample of subjects. This could be achieved by analyzing the synergic activity of cervical muscles during the voluntary activity of wheelchair propulsion. We will attempt to establish relationships between neck reflex stiffening and development of pain or musculo-skeletal traumatisms in the cervical spine area.

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**REFERENCES**


