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How does wearing a lumbar orthosis interfere with gait initiation?

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ABSTRACT

The interaction between medical devices and the human body must be evaluated in standardised laboratory tests. Since wearing a lumbar orthosis is assumed to reduce lower back mobility and reinforce trunk movement control through imposed lordosis, this device is expected to affect gait initiation which requires trunk and pelvic rotations. Thirteen healthy subjects were asked to initiate gait without orthosis (control) and orthosis with or without lordosis constraints. The biomechanical parameters usually reported for gait initiation were studied and no statistically significant effects were found. Indeed, the duration of the anticipation, and execution phases and maximal instantaneous velocity of centre of gravity at the end of the first step were not modified by the experimental conditions. The lack of interference underlines the robustness of the gait initiation parameters, which therefore may lead subjects to adopt adaptive strategies to retain this invariance. Future experiments should be conducted to highlight these strategies.

Practitioner Summary: The aim of this study was to investigate the effect of various lumbar orthosis characteristics on gait initiation organisation. The results, based on a dynamic analysis of balance strategies, showed that the medical device had no repercussions on movement control. Several explanations are proposed, which should be validated by future studies.

1. Introduction

A better knowledge of medical devices is clearly a prerequisite for offering products that are better suited to patients’ needs. This objective applies as well for lumbar orthotic devices, which are generally used to treat low back pain sufferers. Many types of device exist today (soft, semi-rigid, rigid), so that therapists can prescribe the best fitting model. Past studies have highlighted that lumbar orthosis restricts the range of motion of the lumbar spine (postural effect) (Krag, Fox, and Haugh 2003; Jegede et al. 2011; Munoz et al. 2012). The compression contributed by lumbar orthosis could also compensate for the proprioceptive deficit of these patients (McNair and Heine 1999). However, compression alone may decrease lumbar lordosis (Thoumie et al. 1998). Since a change in spinal curvature is thought to generate a proprioceptive deficit (Dolan and Green 2006; Chow, Leung, and Holmes 2007), the lumbar orthosis sometimes raises controversy. To remedy this contradiction, a number of models with a lordosis mechanism have been designed to retain the patients’ physiological lordosis (mechanical effect). Researchers have highlighted the role of the lumbar spine on gait initiation (GI) and more particularly the role played by lumbar axial rotation (Ceccato et al. 2009). The relationship between the lumbar spine and the pelvis has been described in phase, i.e. both pelvic and lumbar parts rotating in the same direction as compared to the upper trunk which rotates in the opposite way. During anticipatory phase of GI described below, lumbar-pelvic complex rotates in the swing limb direction, then inverses in the stance limb direction at the first foot off. However, rotations of the lumbar–pelvic complex are necessary to initiate gait and can be disturbed by the restriction of motion induced by the lumbar orthosis; therefore, the GI task could be used for testing this medical device.

A number of studies have characterised GI (Carlsöö 1966; Mann et al. 1979; Brenière, Do, and Sanchez 1981; Brenière, Cuong Do, and Bouisset 1987; Brenière and Do 1991; Elble et al. 1994; Winter 1995), which is a dynamic process defined by a transition between static and dynamic equilibrium. Since balance is necessarily disrupted in first step execution, anticipatory postural adjustments (APA) are required. These two phases (APA and step execution) correspond to the total duration of GI and determine the velocity of the centre of gravity (CG) at the end of the first step (Brenière, Cuong Do, and Bouisset 1987). This task also involves complex dual
roles of APAs. Indeed, APA is considered a dynamic phenomenon that precedes the onset of voluntary movement and acts as a perturbation and a counter-perturbation for balance (Bouisset and Do 2008). The aim of these APAs is to dissociate the centre of pressure (CP) from the CG in order to create the initial forward fall of the latter and concomitantly towards the stance limb to transfer body weight and ensure dynamic stabilisation. Along these lines, dynamic equilibrium does not respect the fundamental mechanical principle of static objects. Between the onset and the end of the motor act, a transitory state of disequilibrium is indeed observed. Thus, postural kinetic capacity (PKC) can be established, which corresponds to the ability to respond to the perturbation induced by performing a voluntary movement (Bouisset and Do 2008).

To our knowledge, GI is an activity of daily living that has never been studied with subjects wearing a lumbar orthosis. Some authors have taken an interest in other activities (Krag, Fox, and Haugh 2003; Jegede et al. 2011), most particularly walking, but no conclusion was drawn considering the transverse plane with lumbosacral orthosis (Jegede et al. 2011). Consequently, the aim of this study was to describe how healthy subjects control GI with and without lumbar orthosis. Performing the same task, i.e. a constant step length, was used to avoid any confounding factor.

Based on the literature, we hypothesised that with a lumbar orthosis, subjects’ motions would be restricted (axial rotation), likely inducing decreased step amplitude in gait. Therefore, to follow the previously mentioned requirement (step length constancy across conditions), subjects may behave as if they were aiming to take a longer step. As a result, APA, first step durations and maximal CG velocity at the end of the first step were expected to be modulated. (Brenière, Cuong Do, and Bouisset 1987; Caderby et al. 2014).

Furthermore, this study also focused on the lordosis effect. In the orthosis without lordosis condition, compression was viewed by some authors as positively impacting proprioception (McNair and Heine 1999), whereas for others, a deterioration was seen with a decreased lordosis (Dolan and Green 2006; Chow, Leung, and Holmes 2007). Thus, disturbed proprioception may affect movement control and consequently PKC. Along these lines, adding a mechanical effect to spare the physiologic lumbar curvature could in turn remove the proprioceptive impact of the non-lordosis condition. As a result, improved capacity for controlling the trunk may lead the biomechanical parameters of GI to be close to those reported in the control condition.

2. Method

2.1. Subjects

Thirteen healthy young men (age: 21.6 ± 1.7 years; weight: 73.1 ± 5.3 kg; height: 179 ± 6.2 cm; mean ± SD; nine right-footed dominant, four left-footed dominant) participated in this study. The subjects gave their written informed consent and the experimental procedure was consistent with the ethical standards in the declaration of Helsinki. Recruiting only men was not an inclusion criteria, it results from population recruitment which was mainly masculine.

2.2. Measurements

A three-dimensional force platform (60 x 120 cm, AMTI, Watertown, MA, USA) was used to measure external forces (sampling frequency at 250 Hz). Reaction forces in newton ($F_x, F_y, F_z$) and moments in newton meter ($M_x, M_y, M_z$) were obtained. Force plate data were low-pass filtered using a fourth-order recursive Butterworth filter with a cut-off frequency of 10 Hz. From Newton’s laws, it can be written that $a_{CG} = F/m$, where $a_{CG}$ is the linear CG body acceleration, $m$ is the mass of subject and $F$ is the ground reaction force along the progression axis. Instantaneous CG velocity ($v_{CG}$) was calculated by integrating $a_{CG}$ with the trapezium rule (constant of integration equal to zero). The instantaneous coordinates of CP, mediolateral (ML) and anteroposterior (AP) displacements were also studied through $-M_y/F_z$ and $M_x/F_z$, respectively.

2.3. The lumbar orthosis

The semi-rigid lumbar orthosis used in this study (Figure 1) has been previously described (Munoz, Rouboa, and Rougier 2013). This orthosis maintains physiological lordosis with a frontal vertical panel and a curved rigid shell at the back. The textile part of the device is made of polyamide, polyethylene foam, cotton, elastane and elastodiene. The rigid back part as well as the front frame are made of polyethylene, aluminium, steel and stainless steel. On the orthosis, two rows of rubber pads can be added or removed which impact the geometry of the lumbar spine inferring the two orthosis conditions detailed in the next section. With the pads, the orthosis is in lordosis configuration, whereas without, the orthosis has no lordosis configuration. The weight of the orthosis was 1.1 kg. Weight impact on GI was recently seen with an additional load of 15% body weight (Caderby et al. 2013), but no evidence can be drawn with the low load of the current orthosis. For all the conditions, the orthosis was always adjusted by a single investigator in order to optimise reproducible tightening and positioning. It was placed over casual t-shirts worn by the subjects.

2.4. Experimental procedure

Before the experimenter gave orally the onset walking signal, the subjects stood upright, barefoot and motionless
on the force platform. All subjects performed 15 trials and were asked to initiate gait with their favorite leg. This procedure was repeated through three randomised conditions: (i) control (without any orthosis), (ii) lumbar orthosis without lordosis (L) and (iii) lumbar orthosis with lordosis (LL). Overall, each subject performed a total of 45 trials on a five meters walkway. Five meter can be considered a fair distance for studying gait initiation. In fact, subjects can reach steady-state velocity and then stop their movement. Steps affected by the decreased velocity are those of the steady state and not initiation. A short time adaptation was allowed to the subjects before performing the recording trials. They were asked to try the future task once a new condition started. Indeed, a minimum of three preliminary trials were performed.

Step length was imposed to be identical for all subjects by markers on the platform in order to perform a similar task throughout all conditions, whereas the feet angle was let free. The step length was important because it conditioned the level of lumbar–pelvic rotation.

2.5. Dependent variables

For each trial, the first step was analysed and the biomechanical parameters of GI were provided (Figure 2): APA duration (dAPA), first step execution duration (dEXE), step length \( L_{\text{step}} \) and CG peak velocity \( v_{\text{CG peak}} \).

APA duration was assessed through ML instantaneous coordinates of CP (Figure 2(a)). This duration lasted from the first change on the ML-CP displacements to the onset of the first single support (foot-off of the first step). 

\[
dAPA = FO1-t0. 
\]

First step execution duration corresponded to the time between the onset of a single support (previously obtained) and the onset of the double support (foot contact). 

\[
dEXE = FC-FO1. 
\]

Step length \( L_{\text{step}} \) was obtained through AP-CP displacements (Figure 2(b)) and was defined as the distance between the initial CP position before any change occurred on the mechanical trace and foot off of the second step (FO2).

Lastly, \( v_{\text{CG peak}} \) peak corresponded to the \( a_{\text{CG}} \) first sign changes, which occurred at the end of the first step, shortly after the foot contact (Figure 2(c)).

2.6. Statistical analysis

For each trial, the biomechanical parameters were computed to obtain a mean for each subject and each condition. The means of all subjects for a given parameter were
4. Discussion

Before interpreting the results, it may be helpful to recall that the aim of this study was to assess the effects of a medical device, a lumbar orthosis, on the control of GI process in healthy individuals. Since subjects were required to perform a constant step length, this study focused on the possible modifications in this movement control.

Firstly, it should be pointed out that uniform step lengths were taken in all conditions (table 1), allowing the initial question to be answered.

The length imposed for the first step was long in order to amplify rotations of the lumbar–pelvic complex (Ducroquet, Ducroquet, and Ducroquet 1965; Inman, then compared. Since all samples were normally distributed (Shapiro & Wilk test) and variances were homogeneous (F test), the data were analysed using a one-way ANOVA with repeated measures for each parameter. The independent variable corresponds to the lumbar orthosis wearing. The OpenStat software was used and the significance level was established for all tests at $p < 0.05$.

3. Results

Table 1 displays the results of this study for all parameters. Overall, the statistical analysis was unable to highlight any significant effect for all the variables studied. This applies for:

- the length of the first step ($L_{\text{step}}$): ($F(2,24) = 0.119, p = 0.888$);
- the duration of anticipation phase (dAPA): ($F(2,24) = 1.359, p = 0.276$);
- the duration of first step execution phase (dEXE): ($F(2,24) = 1.024, p = 0.374$);
- the peak velocity of centre of gravity ($v_{CG}$ peak): ($F(2,24) = 1.040, p = 0.369$);

Table 1. Summary of results for each condition (means ± SD).

<table>
<thead>
<tr>
<th></th>
<th>Control (without orthosis)</th>
<th>Orthosis without lordosis (L)</th>
<th>Orthosis with lordosis (LL)</th>
<th>Statistical analysis</th>
</tr>
</thead>
<tbody>
<tr>
<td>$L_{\text{step}}$ (m)</td>
<td>0.85 ± 0.026</td>
<td>0.851 ± 0.036</td>
<td>0.848 ± 0.032</td>
<td>$p = 0.888$</td>
</tr>
<tr>
<td>dAPA (s)</td>
<td>0.563 ± 0.065</td>
<td>0.559 ± 0.080</td>
<td>0.574 ± 0.085</td>
<td>$p = 0.276$</td>
</tr>
<tr>
<td>dEXE (s)</td>
<td>0.349 ± 0.032</td>
<td>0.348 ± 0.050</td>
<td>0.343 ± 0.051</td>
<td>$p = 0.374$</td>
</tr>
<tr>
<td>$v_{CG}$ peak (m s$^{-1}$)</td>
<td>1.719 ± 0.14</td>
<td>1.695 ± 0.149</td>
<td>1.704 ± 0.145</td>
<td>$p = 0.369$</td>
</tr>
</tbody>
</table>

Figure 2. Biomechanical parameters of gait initiation. (a) Mediolateral displacements of CP ($t_0$: first sign of change of mechanical trace; $F_{O1}$: foot off, first step; $F_{C}$: foot contact). (b) Anteroposterior displacements of CP ($L_{\text{step}}$: step length; $F_{O2}$: foot off, second step). (c) Acceleration and instantaneous velocity of CG.
Rolston, and Todd 1981; Bruijn et al. 2008; Liang et al. 2014), leading to long dAPA (~0.56 s) and short dEXE (~0.35 s). These findings are in accordance with research on GI, most particularly with the pioneering study reported by Brenière, Cuong Do, and Bouisset (1987). Nonetheless, the results of the present study show that neither the anticipatory nor the execution phases were affected by the different orthosis conditions. Instantaneous CG velocity at the end of the first step also remained unmodified across conditions. The timing of movement appears unchanged when adding an orthosis. To explain the lack of an orthosis effect on the biomechanical parameters of GI, several hypotheses can be proposed.

The first hypothesis postulates that lumbar orthosis has no effect on lumbar–pelvic complex mobility within this range of motion, contrary to our initial working hypothesis. In fact, some authors have reported GI task amplitudes for rotation of the lumbar and pelvic areas of about 10° (Ceccato et al. 2009), even if gait velocity was described as comfortable in their study. Other authors (Jegede et al. 2011) showed that lumbar orthosis decreases the full active range of motion (maximal rotation) of these anatomical sites. Thus, restrictions on mobility while wearing the orthosis were only highlighted for the full range of motion, but not for reduced rotations as observed in GI. Along these lines, our lumbar orthosis may seem to poorly restrict the lumbar–pelvic mobility and therefore cannot really impact the subject’s motor and postural programmes. Coordination between postural movement (APA) and focal movement (step) (Bouisset and Do 2008) remains intact, inferring unchanged biomechanical parameters for GI.

The second hypothesis views lumbar orthosis as restricting the range of motion (always in the transverse plane), even for these amplitudes. Considering human gait initiation as an inverse pendulum (Brenière and Do 1991), one may consider orthosis as having an effect on mobility without influencing the biomechanical parameters of GI. Interestingly, this model explains some invariance in the characteristics of this task. In the sagittal plane, humans are indeed mainly articulated around the ankles and oscillate forward to create initial CG velocity, which is the functional role of APAs. These joints are highlighted and those above, such as the hips and lower back, are not taken into account. Transverse plane and consequently rotations are not considered in this model, explaining why a perturbation of central body mobility has only slight repercussions on biomechanical parameters. In this case, lumbar mobility would not impede GI. Durations of postural and focal movements persist even though subjects lose part of their lumbar–pelvic mobility.

However, even though rotations are not considered in this model, their existence cannot be ruled out (Ceccato et al. 2009). Therefore, if lumbar orthosis restricts lower trunk mobility, subjects could compensate for this restriction by using other parts of their body. This third hypothesis may explain why subjects succeed in performing the same task without changing outcomes. In this case, the organisation of movement would be modified to achieve the required task and respect the timing necessary to initiate gait. In other words, internal forces may be modified to preserve external forces, which act directly on GI parameters. This compensatory motor strategy could further involve the postural chain such as free segments (arms) for preserving postural kinetic capacities. Checking this possibility would require a study of the entire body’s kinematics.

The last point is the importance of lower trunk proprioception for controlling trunk as well as GI movement. In other words, how can the lack of a difference between L and LL conditions be explained? Obviously, including healthy subjects means that no proprioceptive deficit at the onset was detectable. Adding a mechanical effect through an imposed lordosis in this case may change the spinal curvature slightly and therefore slightly improve proprioception. However, without a mechanical effect, the compression would certainly have modified sagittal vertebral equilibrium (Thoumie et al. 1998), inducing a controversial proprioceptive contribution to movement control (Dolan and Green 2006; Chow, Leung, and Holmes 2007). Nonetheless, since no repercussions on the biomechanical parameters of GI were reported in these two orthosis conditions, GI should be mainly interpreted as an independent lower trunk sensory cues task since they are not taken into account. This would explain why the modifications of proprioceptive trunk cues only slightly impact movement control. In a recent study also involving healthy subjects (Munoz, Rouboa, and Rougier 2013), in sitting, another daily life activity, this same semi-rigid lumbar orthosis was shown to interfere only with unstable support (seesaw) resulting in biomechanical adaptations proportional to the lordosis constraints. In contrast, no effect was reported when subjects were sitting on a stable support. Taken together, these observations suggest that biomechanical modifications mostly appear during complex and constraining movements.

The current study attempted to propose a constrained task consisting in an imposed long step length to amplify lumbar–pelvic complex rotations. It should be noted here that standardised the step could at least in part explain this lack of effect. This length constraint was not the same for all subjects, as shown by the rather large standard deviation of the heights (6.2 cm). This is clearly the main limitation of this study since the taller the subject, the easier he performs a longer step. The level of lumbar–pelvic rotation varied therefore according to subjects’ heights and consequently, outcome measures were inauspicious to...
modifications in tall compared to small subject. Another limitation includes each subject’s unknown sagittal vertebral equilibrium profile. For ethical reasons, these profiles, based on X-ray, were not established and we considered that all subjects had a physiologic spinal curvature. Finally, including only men can also be considered as a limitation. As explained in the method, volunteers were recruited in a population which was mainly masculine. Unfortunately, the lumbar restriction of motion was not measured, simply because this research only focused on the impact of this medical device on the control of GI. Therefore, no hasty conclusion can be drawn concerning its lack of impact on this daily living activity. Other kinematic variables would need to be studied for clearer answers, in particular on the capacity of this device to have postural (reduction of the functional range of motion) and mechanical effects on the lumbar spine without distorting the timing of the successive GI phases. Finally, it should be underlined that these results are only valid and specific for the lumbar orthosis model tested.

5. Conclusion
Based on an ergonomic approach, this study demonstrated that no direct link exists between the characteristics of the device worn and the parameters measured. The semi-rigid lumbar orthosis does not appear to influence the biomechanical parameters of GI in healthy young adults and more specifically on the subjects’ ability to counteract the perturbation induced by step movement. However, as compared to healthy young subjects, it is likely that older healthy or pathologic (low back pain) individuals would display contrary behaviors, inferring a more powerful interaction with lumbar orthosis wear.

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