Gait of a deafferented subject without large myelinated sensory fibers below the neck

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Article abstract—We evaluated the gait pattern of a deafferented subject who suffered a permanent loss of large sensory myelinated fibers below the neck following an acute episode of purely sensory neuropathy 21 years ago. The subject has developed several strategies to achieve a secure gait, namely: (1) a reduction of the degrees of freedom by freezing the knee articulations during the stance phase, (2) a preservation of body balance by enlarging his base of support, and (3) visual monitoring of his step by stabilizing the head-trunk linkage together with a characteristic forward tilt. As a result, the gait of the deafferented subject lacks the fluidity of normal gait. Compared with normal subjects, the gait pattern of the deafferented subject is characterized by a shorter cycle length, a longer cycle duration, a slower speed, and a lower cadence. Using a dual-task paradigm, the attentional demands for walking were particularly important (as indexed by longer probe reaction times) during the double-support phase, suggesting that the deafferented subject uses the double-support phase as a transitory stable phase to update cognitively the postural features necessary for generating his next step.

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Walking is a highly practiced activity, often considered to be automatically controlled. Nevertheless, continuous regulation and integration of sensory inputs of various origins is required to preserve the postural balance and to achieve a permanent adaptation to an ever-changing environment. Experiments with experimental animals have led to significant understanding of these mechanisms. Pathology may also offer unique opportunities to evaluate the role of sensory feedback in human motor control processes, especially patients suffering partial or acute loss of proprioception and touch, as observed in the sensory neuropathy syndrome. This syndrome, which follows an acute infection of unknown origin, is presumably associated with a cross-reaction between antibody to the foreign antigen and the sensory neurons of larger myelinated fibers. The structural loss is definitive and most patients do not show any recovery of nerve function, although they generally develop compensatory strategies by using substitutive feedback channels, mainly visual. Studies of such cases may lead to a better understanding of the way and the extent to which cognitive strategies and supplementary sources of information overcome the lack of muscular and cutaneous proprioception in the control of coordinated actions.

We recently have had the opportunity to study a rare case of extensive sensory neuropathy. This deafferented subject is deprived of all contingents of large myelinated sensory fibers below the neck. In spite of this drastic loss of somesthetic and proprioceptive information, he recovered a walking capacity that prompted us to analyze carefully the kinematics of his walking pattern and to measure the attentional cost of his performance compared with that of normal subjects. Attention is often considered, at least by cognitive theorists, as an extension of sensory processes, thus providing a link for studying the interdependence between sensorimotor and cognitive systems for the control of posture. Compared with control subjects, we wanted to examine whether the deafferented subject, because of the extensive loss of sensory information, needs to allocate a greater portion of his attentional resources to the walking task. Without going into the specific details of cognitive resource models, there are three basic assumptions to the use of such models: (1) central processing capacity is limited, (2) performing a task requires a given portion of this capacity, and (3) if two tasks performed simultaneously require more than the total capacity, the performance on one or both will be affected negatively. When the primary task (walking in the present experiment) is unaffected...
by the introduction of a probe reaction time task (discrete secondary task, consisting of a simple reaction time to an auditory or visual stimulus), an increased probe reaction time reflects an increased attentional load.\textsuperscript{11,14} Using this general approach, walking requires more cognitive processing than simple sitting or upright standing posture.\textsuperscript{15,16} In the present experiment, we used a dual-task experimental paradigm similar to that employed by Lajoie et al. and others\textsuperscript{16,17} to examine how much attention the deafferented subject needed to allocate to the walking task.

Methods. The deafferented subject (I.W.; aged 40 years) suffered a permanent loss of large sensory myelinated fibers following an acute episode of purely sensory neuropathy at the age of 19. There was no recovery and this episode left him without light touch or proprioception below the neck. Neurophysiologic tests showed no sensory nerve action potentials and no H reflex below the neck.\textsuperscript{8} In contrast, muscle power and bulk were normal, as was concentric needle electromyography. Cutaneous-muscular reflexes were absent.\textsuperscript{18} He had a similar perception of pain, heat, and cold over his unaffected head and over his affected body, suggesting that there was significant sparing of small myelinated and unmyelinated fibers. His perception of muscle fatigue and aching were normal. A more detailed clinical description of the case can be found elsewhere.\textsuperscript{19}

The data reported come from two different sessions of testing spanning a period of 2 years. In a first session, the subject walked at his own pace across a laboratory floor. The gait was recorded using a CODA motion analysis system. Markers were placed on the shoulders, elbows, wrists, hips, knees, ankles, and feet. The position of each marker was output at a sample rate of 200 Hz. The monitoring system comprised three scanners with divergent beams in the X, Y, and Z axes. The lateral two scanners were 1 m apart. By a reflectant system and comparison with the emitting scanner, it was possible to reconstruct position within the three dimensions. Six cycles were recorded, of which the middle three cycles were quantified. In a second session, the subject walked on an 8-m long pathway providing the recording of approximately seven walking cycles. The subject's shoes were equipped with conductive material under the heel and toes of each foot. Contacts with the pathway, covered with aluminum-wire netting, were digitally coded to provide accurate temporal values corresponding to the onset and offset of right and left single-support and double-support phases. The horizontal foot displacement of the two feet was recorded via small mono-filament wires attached to the rear of each shoe cover. Each wire was wrapped around a plastic wheel (30-cm circumference) fixed onto a gear-box system having a 4.6:1 ratio. A 10-turn high-precision potentiometer was mounted on the shaft of the rotating axis and provided a voltage proportional to the distance covered. The wires were maintained stable with a constant resistance spring (0.66 N) fixed into the rotating mechanisms. This low resistance served only to prevent the wires from shivering; it did not affect the walking and was not perceived by the subject. The system provided a resolution of 3 mm. Additional gait parameters were recorded with a 3D Selspot two-camera system. The sampling rate was 500 Hz and cameras were positioned on the left side of the subject (20 feet from the subject, 27 feet apart). The camera placement allowed the recording of a little more than one complete walking cycle (i.e., the third walking cycle). Active infrared markers also were placed on the left shoulder, elbow, hip, knee, ankle, and foot. Electromyographic activity from the vastus lateralis, medial hamstring, tibialis anterior, and medial gastrocnemius were also recorded with the use of Ag/AgCl surface electrodes with preamplification at the source (Therapeutics Unlimited). The signals were full-wave rectified and integrated (2.5 msec time constant). All signals were sampled at 500 Hz.

For comparison purposes, kinematics and electromyographic normative data were taken from Winter.\textsuperscript{20} The mean of the young adults (n = 11) for Winter's normative data was 24.9. Subjects walked at their natural cadence and data were collected on a minimum of eight walking trials.

To evaluate the mental load required for locomotion, the deafferented subject walked at his preferred pace and was asked to consider the walking task as the primary task. The secondary task was to respond as rapidly as possible to an auditory stimulus (1 kHz, 50 msec duration) with a vocal response ("top"). The subject wore a helmet equipped with a microphone. The analog signal from the microphone was used to detect the onset of the verbal response. Reaction times (RTs) were evaluated by computing the temporal difference between the presentation of the stimuli and the onset of the verbal responses. Signals from foot contacts and the microphone were sampled at 500 Hz. In addition, the secondary task was performed alone when the subject was seated. For the sitting task, 10 stimuli were given following one of five randomly presented preparatory periods: 3, 3.5, 4, 4.5, and 5 sec. Before the walking trials, the subject familiarized himself with the walking environment (five practice trials). After these trials, data were collected for six trials without the secondary task (control condition). These trials served to establish the subject's walking pattern. Then, the subject performed 24 trials. The stimuli were randomly presented (3 to 5 seconds after a verbal warning) on left foot toe off (i.e., at the onset of the single-support condition of the third walking cycle; 10 trials) or on left foot heel contact (i.e., at the onset of the double-support condition of the third walking cycle; 10 trials). Four catch trials (i.e., without stimulus) served to prevent any anticipation. The data of Lajoie et al.,\textsuperscript{9} wherein eight healthy young adults (five men and three women, mean age = 26.0, range 22 to 34 years) were tested in similar conditions, were used for comparison purposes.

Results. Kinematics of the gait. Figure 1 presents stick figures of the gaits of the deafferented subject and a control subject viewed from the sagittal plane. Several differences between the deafferented subject and the control subject are observed. The shoulder of the deafferented subject tilted forward during the beginning of the step-through phase (i.e., onset of the double-support phase), and the knee was locked before and during weight-bearing by that foot. The gait of the deafferented subject was also characterized by a wider base of support. Table 1 presents cycle length, cycle duration, speed, and cadence of the deafferented subject's walking pattern for the two sessions. Data for the second session are taken from the control trials (i.e., walking without a
secondary task). For comparison purposes, normative data from Winter and Lajoie et al. also are presented. Compared with normal subjects, our subject had a shorter cycle length, a longer cycle duration, a slower speed, and a lower cadence. The deafferented subject walked similarly in the two sessions, indicating well-established and stable gait. Data obtained for the second session were submitted to separate one-tailed t tests (mean of a subject compared with a population). Results are presented in table 1.

Electromyographic activity when walking. The electromyographic patterns of four major muscles involved in walking are presented in figure 2. They are also compared with the normative data of Winter. To allow a comparison with controls, EMG signals obtained for the third cycle (second session, control trials without stimulus) were time-base normalized to 100% of the cycle. Clear differences between the deafferented subject's EMG patterns and the controls were observed. The vastus lateralis (VL) shows a significant level of activation for about 75% of the walking cycle while it is solicited for only 25% of the cycle in normal subjects, with a peak of activity at weight acceptance (at about 10% of the cycle). Winter has proposed that the VL activity serves to control the amount of knee flexion and to assist in the extension of the knee in mid-stance. To insure stability when the leg is weight-bearing, the deafferented subject contracts the VL for a longer period (i.e., from 0 to 40% of the cycle). This activity may explain why he shows no flexion of the knee as the foot strikes the ground. For the deafferented subject, the medial hamstring (MH) shows a peak level of activation at 45% of the cycle (i.e., late in SS while the leg was in stance phase); controls, on the other hand, have no significant activity during this portion of the cycle. The VL and MH are biarticular muscles that serve to stabilize the hip and the leg. For the deafferented subject, the peak MH activation may serve to stabilize the trunk in the forward tilt position. The deafferented subject adopts this position presumably to permit vision of the lower segments throughout the walking cycle. The EMG pattern of the tibialis anterior shows major difference at the beginning of the walking cycle. The deafferented subject's muscle is relatively silent compared with a peak activation in controls. For controls, the peak activation serves to maintain the ankle in a dorsiflexed position. The deafferented subject walks with the medial surface of the foot pointed forward, creating less need for foot retention near heel contact. Finally, the medial gastrocnemius of the deafferented subject shows a similar pattern of activity to that of normal subjects.

Attentional demands necessary for walking. An essential prerequisite of the double-task methodology is that the addition of the secondary task (i.e., the verbal response to the auditory stimulus) does not modify the primary task (i.e., walking). For this reason, the average cycle length, duration, speed, and cadence, when probes were given in single- and double-support conditions, were compared with the control no-stimulus condition (table 2). The kinematic parameters were similar across the three walking conditions. The deafferented subject did not modify his walking pattern when stimuli were presented, therefore validating the RTs as a representative index of the attentional demands for walking.

Data for the controls are taken from Lajoie et al. In the seated position, our subject was faster than most control subjects to react to the auditory stimuli (166 msec versus 235 msec, respectively; \( t = 1.69, p > 0.05 \)). When walking, the deafferented subject's RTs were similar to those of controls when the stimuli were given in the single-support phase (371 msec versus 321 msec, respectively; \( t = 1.10, p > 0.05 \)).
Figure 2. Normalized electromyographic activity (average and standard deviation) of the vastus lateralis, medial hamstring, tibialis anterior, and medial gastrocnemius muscles for the deafferented (left portion of the figure) and control (right portion of the figure) subjects. Data for control subjects are taken from Winter's normative database. The EMG data are time-base normalized to 100% of the stride before ensemble averaged.

Table 2 Parameters for three walking conditions*

<table>
<thead>
<tr>
<th></th>
<th>No-stimulus</th>
<th>SS</th>
<th>DS</th>
</tr>
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<tbody>
<tr>
<td></td>
<td>Average</td>
<td>SD</td>
<td>Average</td>
</tr>
<tr>
<td>Cycle duration</td>
<td>1,335</td>
<td>26</td>
<td>1,335</td>
</tr>
<tr>
<td>Cycle amplitude</td>
<td>1,278</td>
<td>46</td>
<td>1,346</td>
</tr>
<tr>
<td>Speed (m/sec)</td>
<td>0.95</td>
<td>0.016</td>
<td>1.008</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>89.9</td>
<td>2.0</td>
<td>89.3</td>
</tr>
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</table>

*Cycle length, cycle duration, speed, and cadence in the control no-stimulus condition and when stimuli were given at toe-off (onset of the single-support phase—SS) and at heel-contact (onset of the double-support phase—DS) for the deafferented subject.
efficient speed of progression. Comparison with controls, the slower RTs observed when stimuli were given at the onset of the double-support phase suggest that this phase requires an increased attentional load.

**Discussion.** The deafferented subject has developed several strategies to achieve a secure gait: (1) a reduction of the degrees of freedom (normally controlled by local peripheral loops in the control of posture) by "bracing" himself while moving; (2) a preservation of body balance by enlarging his base of support; (3) the visual monitoring of his step by stabilizing the head-trunk linkage together with a characteristic forward tilt. Compared with normal subjects, the gait pattern of our subject is characterized by a shorter cycle length, a slower speed, and a lower cadence. This general gait pattern is reminiscent of the adaptations observed for elderly people. For the elderly, these adaptations serve chiefly to produce a more secure and less destabilizing gait and an energy-efficiency speed of progression.

Following the classical distinction introduced by Hess, walking is the product of two interactive mechanisms: (1) the ereismatic or postural component that provides the support for the propulsive force and postural adjustment for preserving overall body orientation in the field of gravity forces, and (2) the teleokinetic or instrumental component, that is, the goal-directed displacement of body and limbs. Reference frames may be the basis for the underlying mechanisms necessary to insure that the body and the limbs are rightly positioned with regard to environmental landmarks. Two such frames are generally considered: (1) an egocentric reference frame regarded as a dynamic structure that stems from a continuous updating of the relative positions of the body segments by way of static and dynamic proprioceptive signals, and that tunes the motor commands for spatially oriented movements; and (2) an exocentric reference frame that derives from a memory-based internal construct built from extracting the stable covariant features of the visual environment that remain stable as the body moves.

One of our initial basic assumptions concerning the state of the deafferented subject is that a dynamic egocentric frame of reference, which needs a permanent updating of the postural schema, is no longer available in complete absence of muscular proprioception. The deafferented subject has developed several strategies to achieve a secure gait: (1) bracing himself while moving; (2) visual monitoring of his step by stabilizing the head-trunk linkage; (3) the preservation of body balance by enlarging his base of support. These operations, presumably automatic in normal subjects, heavily load the deafferented subject's cognitive system (as indexed by the longer RTs). Conversely, the single-support phase does not produce a different load in our subject and the control subjects. This phase is mainly conducted with passive forces and appears to require less or no on-line regulation.

Previous results obtained in pointing tasks with another deafferented subject support the hypothesis that a dynamic egocentric frame of reference, which needs a permanent updating of the postural schema, is not available in complete absence of muscular proprioception. (This subject, G.L., suffers a permanent and specific loss of the large sensory myelinated fibers in the four limbs following two episodes of sensory polyneuropathy that affected her whole body below the nose.) Moreover, the model of "coherent copies," proposed by Droulez and Darlot, accounts for the capacity of the visual system and of visual kinesthesia to complement the direct proprioceptive information originating from the muscular sensors, or even more, to provide a valid substitute when the dedicated sensors are temporarily or completely disabled. Mittlestaedt proposed that, in normal subjects, the head is referred to the "idiotropic vector" as an internal egocentric reference related to the trunk main axis. Conversely, our subject is more likely to use a head-centered frame to which his deafferented trunk and body segments (albeit vividly represented in a "visual body image") can be referred via neck afferent information. This strategy allows the leading head segment, which is visually anchored on stable environmental landmarks, to steer the locomotion of the deafferented body.

Further empirical observations support this viewpoint. Indeed, the deafferented subject G.L., completely devoid of neck proprioception, in contrast to I.W., cannot walk unassisted, mainly because of her inability to maintain balance without support, in spite of being able to stabilize her head with respect to the environment via visual and vestibular cues. This observation suggests a crucial role for neck proprioception in articulating the ereismatic and teleokinetic constraints of locomotion and the functional

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leading role of a stabilized head in the featuring of the body schema. Berthoz and Berthoz and Pozzo suggested that the brain uses the head as a stabilized inertial guidance platform and as a mobile frame of reference to provide all the necessary requirements for generating coordinate multilimb activities. For this purpose, and owing to continuous changes of the forces exerted on the musculoskeletal system by gravity during body displacement, a continuous updating of the internal representation of head motion appears mandatory. Compelling evidence now exists to show that the vestibular system cooperates with the visual system to compute central configuration of head motion and position in space. Such a mechanism is obviously preserved in both deafferented subjects. However, because of the preservation of his neck afferents information, our subject alone can elaborate efficient but costly cognitive strategies to meet the postural requirements for walking.

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References

