ELECTROMYOGRAPHIC ANALYSIS OF PARTIAL BODY WEIGHT SUPPORT AND LOAD CARRYING ON THE NORMAL HUMAN GAIT

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INTRODUCTION

Walking is characterized by a pendulum-like exchange of the gravitational potential energy and horizontal kinetic energy of the body's center of mass. This mechanism conserves mechanical energy and presumably metabolic energy. However, Cavagna et al. (1963) found that the vertical position and forward velocity of the body's center of mass fluctuate out of phase during each step cycle. Unlike an ideal pendulum, which has 100% recovery, humans normally recover only \( \frac{67}{65} \) of the total mechanical energy (Cavanagh et al. 1989), because the energies do not fluctuate at opposite equal rates. As a result, mechanical energy must be added to the system by the leg muscles during each step.

Cavagna et al. (1963) were among the first to speculate about the effect of reduced gravity on the mechanical energy fluctuations in walking humans. Gravity not only controls the actions but also the forms of all organisms. Despite our intuitive appreciation for the influence of gravity, we do not understand how gravity interacts with other forces, such as inertia, to affect many biological and physical processes. Such are the cases for normal walking, as well as a gait retraining strategy.

The effect of body weight support (BWS) on the EMG, kinematic, and temporal distance characteristics of normal human gait during treadmill walking has been shown by Finch et al. (1991). However, Cavanagh et al. (1989) have reported that, having performed a kinematic analysis of astronauts exercising while tethered to a treadmill, the posture that increased with walking speed was the forward lean. Consequently, understanding how reduction in vertical force, isolated from horizontal force, affects the musculoskeletal loading may provide important insights into the determinants of the gait retraining strategy.

The purpose of this study was to investigate the effect on the EMG of various degrees of BWS and load carrying (LC), as compared to full weight bearing (0%), on the normal human gait by using a rolling-trolley BWS and LC system.

METHODS

Subjects
Ten healthy male subjects (21.3±1.4yrs, 66.2±8.0kg, 170.9±4.7cm) volunteered to participate in this study after informed consent had been obtained. None of the subjects had a history of neuromuscular disease.

Procedure
The protocol consisted of two testing sessions. 1) Manual muscle testing: fair (MMT.F) by performing in a stable cadence (0.5Hz), which was kept constant with the use of a metronome. It was performed in an isotonic manner more than 10 times. 2) Walking on the floor with 10%, 25% and 50%BWS / LC, and with full weight bearing (0%) at the cadence of 120 steps/m with the use of a metronome.

Body-Weight-Support (BWS)
The subjects walked on a 10m walkway while 10%, 25% and 50%BWS were provided. The percentage of BWS provided was normalized to each subject's weight. The BWS system (Figure 1) was mounted in the ceiling (in height 4.5m) over the 10m walkway located in the laboratory. The BWS apparatus consisted of the rolling-trolley and a harness, which mechanically supported the subject vertically over the walkway. The harness supported the subject primarily about the pelvis, lower abdomen, and chest to avoid interfering with lower limb movement. The rolling-trolley, which was connected to the cable line, was dragged by the staff, the operation of which was synchronized to the periods of subject's walking, to free the subject from the need to drag the cart or to pose while in the forward leaning stance.

Figure 1: Schematic drawing of the body weight support (BWS) system. The reduced gravity simulator applied a nearly constant upward force near the body’s centre of mass via a pelvic harness.
Load carrying (LC)
The subjects walked on a 10m walkway while 10%, 25% and 50%LC were provided. The load was normalized to each subject's weight. The load carrying apparatus consisted of two sacks, which carried on the subject's back and abdomen, were confirmed to the trunk.

Recordings
Electromyographic activity was detected by surface electrodes (bipolar silver electrodes 0.5cm in diameter) applied on the skin, 2 cm apart (center to center) longitudinal to the direction of muscle fibers. Recordings were obtained for the following muscles on the subjects' right side: m.tibialis anterior (Ta), m.gastrocnemius (Ga), m.soleus (So), m.vastus medialis (Vm), m.rectus femoris (Rf), m.biceps femoris (Bf), m.gluteus medius (Gme), m.gluteus maximus (Gma). The electrodes were centered over the muscle belly except the So, following conventional skin preparation. Electromyographic signals were band-pass filtered (5-1000Hz), differentially amplified (common mode rejection rate=80 dB), and recorded together with footswitch signals, and auditory signals, on a 21-channel magnetic tape at 9.5cm/s (frequency response=2500Hz). Footswitches were placed bilaterally beneath the heel and the big toe. Each footswitch produced a distinct voltage, allowing for determination of temporal-distance characteristics. The time between one-heel contacts to the next heel contact of the same limb was considered as a 100% cycle.

Normalization technique
The raw EMG signals digitized at a rate of 1000 Hz, full-wave rectified and low-pass filtered with second-order critically damped low-pass filters (3Hz), producing a linear envelope (LE) using a computer. Once the digitized version of the LE was available, it was further processed in two ways. First, the real time (ms) of the gait cycle was normalized (0-100 per cent) so that pooling of the data across subjects would be feasible. Secondly, the amplitude of the LE, expressed mV, was transposed as a percentage of a manual muscle testing: fair (MMT.F), which had been obtained for each muscle previously during the gait session. The EMG signal produced by this MMT.F was transformed into a LE and averaged (5 times peak amplitude) in the same fashion as for the EMG gait data. It is believed that this procedure allowed for the possibility of comparing EMG data in a between-subjects manner, as well as days or experimental trials for a given subject.

For each of the ten subjects, ten gait cycles of data were analyzed. The sampling and normalization procedures allowed for the cumulation of mean and standard deviation values for each of the sample points, obtained in either a within- or between-subject manner. These techniques were used to create an ensemble average profile (with variation around it) for a given muscle (within-subject data) or a grand ensemble average profile (with its variation) for a given muscle (between-subject data).

Statistics
For comparison of an average amplitude value of the EMG signal from different trials within subjects, pair wise Wilcoxon tests were performed with an adjusted alpha value ($p < 0.05$) whenever the preceding Friedman two-way analysis of variance was significant.

RESULTS

Averaged EMG profiles
Concerning the shank muscles, Figure 2 shows the averaged EMG profiles for the right Ta, Ga, So, muscle with 0%, and with 10%, 25%, 50% LC, and with 10%, 25%, 50%BWS. Vertical axis represents the activity level of the muscle expressed as a percent of MMT.F.

Concerning the thigh muscles, Figure 3 shows the peak activity level of the Vm (assessed in the interval from 0% to 10% of the normalized gait cycle) and Bf (assessed in the interval from 90% to 10% of the normalized gait cycle) during three LC conditions, tended to increase with...
increasing LC but stayed unchanged during the activity patterns.

Figure 3: Averaged EMG for the right Vm, RF, Bf, muscle of the normal subject walking on the floor full weight bearing (0%) and with 10%, 25% and 50% body weight support (BWS), and with 10%, 25% and 50% load carrying (LC). Vertical axis represents activity level of the muscle expressed as a percent of Manual Muscle Testing Fair (MMT.F)

The activity of the second peak (assessed in the interval from 50 to 70% of the normalized gait cycle) of the Rf tended to diminish with increasing BWS than the 0% and three LC conditions; a more appropriate EMG profile was noted for 50%BWS, with the initial burst occurring earlier in this swing phase.

Concerning the pelvic muscles (Figure 4), the mean functional activity of the Gme (assessed in the interval from 90% to 40% of the normalized gait cycle) diminished with increasing BWS but increased with increasing LC; however, there were no significant differences across each EMG profiles. At 50%BWS, Gma continued to show a high level activity during the foot contact phase.

Figure 4: Averaged EMG for the right Gme, Gma, muscle of the normal subject walking on the floor full weight bearing (0%) and with 10%, 25% and 50% body weight support (BWS), and with 10%, 25% and 50% load carrying (LC). Vertical axis represents activity level of the muscle expressed as a percent of Manual Muscle Testing Fair (MMT.F)

DISCUSSION

Effects of LC

With increasing LC, the EMG activity of eight muscles tended to increase linearly. The EMG activity of Ga, So, Gem and Gma with 25%LC and all eight muscles with 50%LC was significantly higher than that of 0% (p<0.05). Our data suggest that the LC might have an effect on strength training of the lower limb muscles when treating normal subjects.

Effects of BWS

In this study, with increasing BWS, the EMG activity of Gma, Ta, Bf and So tended to increase, Ga and Vm remained unchanged, and, Gme and Rf tended to decrease. The EMG activity of Ta, So, Bf with 50% BWS and So with 10%BWS was significantly higher than that of 0% (p<0.05).

A motor-driven treadmill and suspension system based on partial body-weight support combined with enforced stepping movements was introduced in the rehabilitation of
nonambulatory hemiparetic patients.

Some reports documented an improved stride-length and decreased EMG mean burst amplitude (Visintin et al. 1989), facilitated gait and decreased clonus (Visintin et al. 1994), increased speed and endurance (Elizabeth et al. 2001) during the treadmill training with 40%BWS, when compared with regular physiotherapy. Others documented the diminished EMG amplitude of Va and So (Hesse et al. 1997), improved higher symmetry (Hesse et al. 1999) during the treadmill training with 30%BWS. The patients in these studies might have exhibited a larger forward leaning posture on the treadmill, probably because of the backward movement of the belt and overhead suspension system using a single cable. Pillar et al. (1991) have reported on another system that utilized a pneumatic system mounted in the ceiling over the 3m conductive walkway which was placed between parallel bars, and found a substantial increase in gait symmetry and velocity of the patients’ ambulation although natural gait velocity of the healthy subjects was impeded by 20%BWS through this system.

These reports suggest that the effects of the treadmill or a pneumatic system training with BWS in patients, which allow them to practice a favorable gait, might be affected not only reduced gravity but also by cable suspension force vector. Our data suggest that BWS did not facilitate a less musculotonic gait when normal subjects walked on the floor by using a 10m rolling-trolley BWS system at the cadence of 120steps/m. Consequently, the reduction of the EMG activity of the eight muscles studied suggests a limit of 10%BWS that can not be exceeded.

**REFERENCES**


