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Effects of load placement on back muscle activity in load carriage

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Summary. The effect of two different load placements (just below mid-back or just above shoulder level) on erector spinae EMG, trapezius EMG, and heart rate were investigated during load carriage. The EMG and heart rates were telemetered from 11 subjects while they walked on a smooth level surface at an average velocity of 5.6 km · h^{-1} carrying a load of 19.5 kg in a specially designed backpack. The average rectified EMG amplitude was calculated digitally for both load placements. The high load placement resulted in significantly higher levels of muscle activity than did the lower placement. Heart rate was not significantly different between the two placements. A qualitative biomechanical analysis suggests that the EMG differences are primarily due to differences in the moments and forces arising from the angular and linear accelerations of the load and trunk. The results indicate that metabolic measures alone are not sufficient to adequately assess tasks which evoke primarily local muscle demands.

Key words: Load carriage – Load placement – EMG

Introduction

Winsmann and Goldman (1976) reported the results of a study comparing two backpack designs differing in their weight distributions. One design had no hip belt, and accordingly placed most of the load on the carrier's shoulders. The other had a hip belt, and distributed the load more evenly between shoulders and hips. Winsmann and Goldman could find no difference between the two packs using metabolic energy measures. They concluded that there appeared to be considerable latitude for backpack design "without demonstrable physiological penalties for one design over another."

Although this conclusion may be valid, it should be recognized that not all "physiological penalties" manifest themselves in metabolic rate measures. These measures reflect the total workrate of the body, but give no indication of whether local loading of relatively small but potentially vulnerable muscles is excessive. A workrate that is acceptable for a large muscle may severely overload a smaller one. Excessive levels of muscular tension must certainly qualify as a physiological penalty, as they can lead to muscular pain and fatigue (Chaffin 1973) and perhaps ultimately to bone and joint disease (Bjelle et al. 1981). For some of the particularly vulnerable muscle groups in the body, excessive levels of tension may be harmful even if the total muscle masses involved are too small to produce a measurable metabolic response. The lack of a significant difference, in metabolic rate, between two different distributions of weight on the back is thus no guarantee that both are equally suitable physiologically. Some measure of the activity level of important muscle groups is necessary before conclusions can be drawn as to physiological suitability.

The purpose of this study was thus two-fold: (1) to investigate the effects of two different load distributions on the activity of selected muscles during load carriage; and (2) to determine whether heart rate measures, used alone as a correlate of metabolic rate, differentiated between the effects of the two backpack load distributions. It was reasoned that if heart rate measures did not detect any difference between the two load distributions, despite a demonstrated difference in muscular tension, then one could conclude that heart rate measures were insensitive to changes in the demands placed upon the musculoskeletal system in this task.

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Materials and methods

The muscles selected for analysis were the erector spinae and upper trapezius *(pars descendens)* muscles. The erector spinae was selected because of the likely relation between its tension and degenerative intervertebral disk disease (Chaffin 1969), and because it is the principal back extensor. The upper trapezius was selected because previous work which also included the vastus lateralis, gastrocnemius, tibialis anterior, and biceps femoris muscles (Bobet and Norman 1982) had shown the upper trapezius to be sensitive to changes in the conditions of load carriage. Moreover, backpackers often notice fatigue and soreness in this region of the body, and the level of tension in this muscle has been implicated in cervico-brachial disorders (Bjelle et al. 1981). In addition, both muscles could be assumed to meet the criteria for assuming a linear EMG/tension relation: during load carriage, their length does not change appreciably, their electrical activity can be readily isolated from that of neighboring muscles, and their level of contraction is submaximal.

Eleven healthy men (age 19–22; height 166–190 cm; mass 53–85 kg) carried a 19.5-kg load around a flat 90-m course at a speed of 5.6 km · h^{-1}, as timed by photocells. The load carriage device was a specially constructed backpack in which the load's centre of mass could be placed either just below mid-back (level of the xiphoid process) or high on the back (level of the ear lobe). In the horizontal plane, the mass center of the load was located about 7 cm directly behind the spine.

The EMG from erector spinae (level of the fourth lumbar vertebra) and trapezius (level of the sixth cervical vertebra) from the right side of the body was recorded using bipolar surface electrodes placed 2 cm apart. An FM telemetry system worn by the subject transmitted the myoelectric signals to a demodulator, thence to an analog FM tape recorder and to a chart recorder. The effective frequency response of the system was 30–300 Hz. Following data collection, the tape-recorded EMGs were played back through 6-Hz linear envelope detectors. The resulting smoothed EMG signal was A/D converted at 50 Hz and stored on floppy disk using a microcomputer.

The subjects practised the route unloaded until they could consistently reproduce the required pace (± 2%). Once they had mastered the pace, two trials of EMG, each of four strides, were recorded. Inspection of the EMG of these strides revealed that they were quite consistent, so two strides from each trial were analyzed. This resulted in four strides being analyzed per subject per load placement, a number which Arsenault (1981) has shown to be adequate to obtain reliable EMG measures during normal walking. Subjects then donned the pack and adjusted the shoulder straps and lap belt to suit themselves. The same procedure as for the unloaded trials was repeated, first with one placement then the other. The order of presentation of the load placements was counterbalanced. In order for the trapezius activity to be comparable across load placements and subjects, arm position was standardized by having subjects maintain their elbows flexed and their hands at chest height, with their thumbs under the backpack straps if a load was present.

Following the last loaded trial, the pack was removed, and a sample of the myoelectric signal was recorded with the subject lying relaxed and supine. The average value of this "noise" signal (after removal of any bias, and rectification) was subtracted from all previous trials for that subject, in order to remove any contribution of electrical equipment noise and ECG crosstalk to each muscle's EMG. The noise trial was performed after the walk rather than before because much of the noise was due to interference from the ECG, and hence varied with heart rate. In order for the noise value to be appropriate, the noise trial had to be recorded while the subject's heart rate was still elevated.

The average amplitude of the envelope was calculated over the four strides for each muscle, and expressed relative to the average EMG amplitude during unloaded walking for that muscle. These normalized average EMG values (hereinafter "AEMG") were analysed using a one-way analysis of variance.

Heart rate was monitored using electrodes placed on the chest. The heart rate signal was telemetered and recorded as if it was another channel of EMG. Heart rate was calculated by manually counting the number of QRS complexes in 16 s of the chart record. At least 3 min of load carriage were allowed to elapse before any heart rate measures were taken, in order to ensure that the heart rate sample was representative. The heart rate results were also analysed using a repeated measures analysis of variance.

Results

The means are presented in Fig. 1. Envelope curves from a representative subject are given in Fig. 2. AEMG means for the lower load placement were significantly ($P < 0.05$) lower than those for the corresponding high load placement. The erector spinae means for both placements fell below 100% (59% and 86%, respectively), indicating that the addition of the load actually decreased erector spinae activity levels over unloaded walking. The trapezius mean for the lower placement fell somewhat below 100% (92%), while that for the high placement fell slightly above (108%). The heart rate mean for the lower placement (108 bt · min^{-1}) was not significantly higher than that for the high placement (105 bt · min^{-1}). As would be expected with the addition of a 20-kg load to the body, both placements produced a significantly higher heart rate than that observed during unloaded walking (97 bt · min^{-1}).

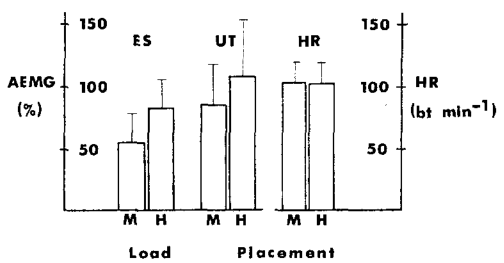

Fig. 1. Means for erector spinae *(ES) AEMG,* upper trapezius *(UT) AEMG,* and heart rate *(HR)* for midback *(M)* and high *(H)* load placements. Thin bars are 1 SE

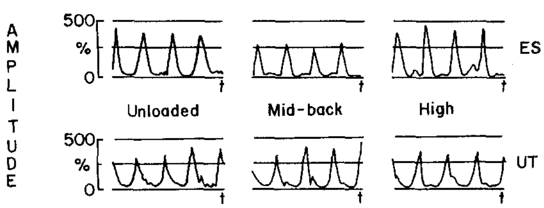

Fig. 2. Smoothed, rectified (envelope) EMG for the erector spinae *(ES)* and upper trapezius *(UT)* for two strides for one subject. Left to right: unloaded walking, walking carrying a load placed at mid-back, walking carrying a load placed high on the back

Discussion

Unloaded versus loaded walking

The observation that heart rates were elevated by 8–10 bt $\cdot$ min^{-1} with a 20-kg load on the back compared to unloaded walking was expected since overall muscular demands are greater to sustain the load. The observation that the low back and shoulder girdle AEMG's under loaded conditions were either lower than or about the same magnitude as those observed during unloaded walking seems, at first glance, anomalous. The reasons for this observation become apparent when both static and dynamic moments of force produced by the load and trunk masses during the walking stride are considered.

Figure 3 is a free body diagram of the trunk (head and arms included) and pack system. The moment of force produced by the erector spinae musculature and other low back tissue must resist the moments produced about a centre of rotation in the low back, for example at the L5/S1 intervertebral disk, denoted by point C. The static forces which can produce moments about this point are the weights of the load and the head, arms, and trunk. These are denoted by W acting at G_{TPC}, the centre of gravity of the trunk and pack combined. The dynamic forces producing moments about C are the horizontal and vertical inertial forces, ma_x and ma_y, respectively. The latter force is not shown because the accelerations in the vertical direction are relatively small. The rotational acceleration of the pack and trunk also contributes to the moment about C. Its effect is denoted by the term $I_G \propto$.

The equation of motion which describes the moment of force *(M)* acting at point C is:

$$M_C = I_G \propto + \, y \cdot ma_x + x \cdot ma_y + x \cdot W. \tag{1}$$

The terms are defined in the figure caption of Fig. 3.

In unloaded walking Winter (1979) has provided data which shows that the horizontal and vertical linear accelerations and the angular accelerations are very small. The mass and moment of inertia of the unloaded trunk are not sufficiently large to produce a significant moment of force when combined with these small accelerations, thus, the contributions of the first three terms in the equation to the low back moment are negligible in unloaded walking. With no load on the back the dominant moment is the static moment *($x \cdot W$)* and is one of trunk flexion because the line of gravity of the combined head arms and trunk (HAT) is located somewhat forward of the lumbosacral joint. This moment must be resisted by erector spinae activity. In loaded walking, the

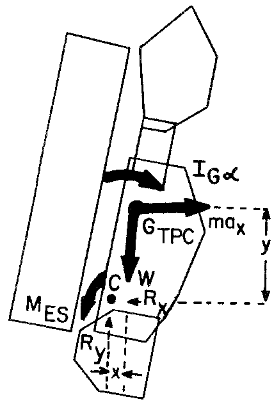

Fig. 3. Free-body diagram of the trunk and pack system. *TPC:* trunk and pack combined; *G: TPC* mass center; *C:* lumbosacral joint; *W: TPC* weight: $\propto$: *TPC* angular acceleration; m: *TPC* mass; a_x and *ay*: horizontal and vertical *TPC* accelerations; x and y horizontal and vertical distances from C to G; I_G: centroidal moment of inertia of *TPC*; M_{ES}: muscle moment due to erector spinae musculature

presence of a load on the back creates a back extension moment which partly offsets the flexion moment of the HAT. This reduces erector spinae activity. The magnitude of the reduction depends upon the weight of the HAT, the particular angle of inclination adopted to balance the moments of force, and the ability of the subject to maintain this balance during the accelerations and decelerations associated with the walking stride.

The elevation of the upper trapezius activity in unloaded versus loaded walking may have been an artifact of the standardized arm position required of the subjects in unloaded walking. This arm position required some arm abduction, and hence some trapezius activity. With the addition of the backpack, subjects could hold their arms in the required position simply by gripping the backpack straps. Some subjects may have employed this method of reducing the trapezius activity to levels slightly lower than those seen in unloaded walking.

High versus Mid-back load centers of gravity

The mid-back placement of the load center of gravity resulted in consistently lower EMG levels in both muscles studied than did the high placement. This difference also is explained on the basis of relative moments of force acting on the low back. Film analysis in a previous study (unpublished results) showed that the postures assumed by the subjects resulted in approximately the same static moment (about 53 N $\cdot$ m) due to the weight of the TPC whether the load was placed high or low on the back. The dynamic moments, however, were appreciably larger with the center of gravity of the load in the high location. The moments due to angular acceleration were 30 N · m for the high placement and 21 N · m for the low, a difference due primarily to the larger moment of inertia in the high position. Those attributable to the horizontal linear acceleration of the system mass center were 33 and 20 N · m in the high and low placement, respectively. Although the peak horizontal accelerations and decelerations were about the same with each step regardless of load placement (1.8 and 2.0 m · s^{-2}) the vertical distance from the center of rotation at the L5/S1 disk to the TPC mass center (y, Fig. 3) was larger with the load placed high than with it placed lower on the back (0.24 and 0.16 m, respectively).

The upper trapezius results can also be explained by the mechanics of load carriage. During loaded walking, the vertical and horizontal accelerations of the trunk are transmitted to the pack through the straps and the hip belt. Because the load is attached to the body, it is accelerated and decelerated by the movements of the trunk. These accelerative forces must pass either through the shoulders (via the straps) or through the pelvis (via the hip belt). Any forces transfered through the shoulder straps inevitably either push or pull on the shoulders, and thus give rise to trapezius activity.

With the high placement, the load is less stable, and tends to sway much more over the course of the stride. This increased swaying must be compensated for by the trapezius musculature, if the carrier is to be able to walk with any stability, and this compensation is what leads to the higher AEMG levels seen in the trapezius with the high placement. As with the erector spinae, the trapezius activity seems to depend more upon the accelerations of the pack than upon its weight.

Based on back muscle tension levels, a mid-back placement appears to be preferable. The moments this placement generates are smaller, demanding lower levels of periodic activity in the back muscles, and its weight actually aids the back muscles to extend the back. Moreover, unexpected linear and angular accelerations of the pack and trunk, such as occur during a stumble, can be handled more safely because of the relatively smaller inertial and gravitational moments associated with a midback rather than a high mass center.

Physiological consequences of load placement

The results here show that it is possible to have considerable differences in local muscle tension, as indicated by differences in EMG activity levels, with no corresponding change in a more global metabolic measure, heart rate. Two mechanisms may be responsible for this observation. First, the erector spinae muscle is only one of many muscles actively involved in load carriage (Bobet and Norman 1982). The energy demands of the erector spinae are therefore only a small fraction of the total energy demand. Accordingly, the erector spinae could change its tension level drastically with little effect on heart rate. A second mechanism is the participation of other muscle groups. Small changes in the position of the trunk, for example, may result in some of the load normally borne by the erector spinae being transferred to some other muscle group. In this way, the total amount of energy required to carry the load could remain constant while some muscles show a decrease in activity. Similar arguments also hold for the trapezius muscle.

Whatever the mechanisms involved, the results here show that heart rate measures are not sufficient to evaluate the physiological demands of differences in load placement on the back during load carriage. Whether other muscles than those monitored here participated or not, the fact remains that different tension levels in the back musculature went undetected by the heart rate measure. Using heart rate measures alone, one might erroneously conclude that the two placements were no different physiologically. Furthermore, under the conditions of this study, the heart rate results can be taken as representative of results that would have been obtained with metabolic measures that correlate with heart rate. Because of this shortcoming, care must be taken in the interpretation of non-significant differences in metabolic rate measures. In many occupational tasks, the cause of pain, fatigue, or injury may often be excessive local muscle tension demands rather than excessive energy demands. For this reason, one should consider supplementing metabolic measures with some measure of muscle tension in the assessment of tasks which evoke primarily local muscle demands.

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