Dynamic analysis of load carriage biomechanics during level walking

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Abstract

This paper describes an investigation into the biomechanical effects of load carriage dynamics on human locomotion performance. A whole body, inverse dynamics gait model has been developed which uses only kinematic input data to define the gait cycle. To provide input data, three-dimensional gait measurements have been conducted to capture whole body motion while carrying a backpack. A non-linear suspension model is employed to describe the backpack dynamics. The model parameters for a particular backpack system can be identified using a dynamic load carriage test-rig. Biomechanical assessments have been conducted based on combined gait and pack simulations. It was found that the backpack suspension stiffness and damping have little effect on human locomotion energetics. However, decreasing suspension stiffness offers important biomechanical advantages. The peak values of vertical pack force, acting on the trunk, and lower limb joint loads are all moderated. This would reduce shoulder strap pressures and the risk of injury when heavy loads are carried.

Nomenclature

m_i	mass of segment i
$\vec{a}_i$	translational acceleration vector for the i^{th} segment's mass centre
$\vec{F}_{ji}$	j^{th} resultant joint force acting on the i^{th} segment
$\vec{F}_{ei}$	resultant external force acting on the i^{th} segment
$\vec{g}$	gravitational vector

I_i	moment of inertia of the ith segment
α_i	angular acceleration of the ith segment
M_{ji}	net muscle moment acting on the ith segment at the jth joint
M_{ei}	resultant external moment acting on the ith segment
M_{ki}	moment of the resultant joint force at the kth joint acting on the ith segment
$\vec{F}_{gi}$, M_{gi}	ground reaction force and moment
$\vec{F}_p$, M_p	pack force and moment
$\ddot{x}_p$, $\ddot{y}_p$	normal and tangential accelerations of the pack mass centre
α_p	angular acceleration of the pack
$\ddot{x}_t$, $\ddot{y}_t$	horizontal and vertical accelerations of the trunk mass centre
θ_t, ω_t, α_t	angular displacement, velocity and acceleration of the trunk
d_x, d_y	normal and tangential positions of the pack mass centre relative to the torso mass centre
u, $\dot{u}$, $\ddot{u}$	displacement, velocity and acceleration of the pack suspension system
m_p	pack mass
I_p	moment of inertia of the pack
F_{xp}, F_{yp}	normal and tangential pack interface forces acting on the pack mass centre
M_{zp}	pack interface moment about the pack's centre of mass
a_1, a_2, a_3	elastic parameters of the pack suspension system
b_1, b_2, b_3	damping parameters of the pack suspension system
c_1, c_2, c_3	inertial parameters of the pack suspension system

F_{x_pack} , F_{y_pack}	horizontal and vertical pack forces acting on the torso mass centre
M_{z_pack}	pack interface moment about the torso mass centre

Introduction

Backpacks have been widely used to increase load carriage capacity. Some groups, such as hikers and infantry soldiers, often carry substantial loads using a variety of backpack systems. Many associated medical injuries have been reported involving tissue damage under straps, back problems and lower limb injuries (Jones, 1983; Knapik, 1996). The objective of this study was to investigate the biomechanical effects of different backpack suspension characteristics, during level walking, in terms of joint loadings, net muscle moments and mechanical energy expenditure. A combined experimental testing and computational modelling approach has been adopted, the aim being a better understanding of load carriage dynamics, its effect on locomotion performance, and the implications for backpack design.

Most load carriage studies can be classified as either physiological or biomechanical. Many researchers investigating load carriage physiology have focused on the energy expenditure of locomotion (Pandolf, 1977; Epstein, 1987). Biomechanical analyses are being increasingly used, including studies of the effects of load carriage on the electromyographic activities of muscle groups (Bobet and Norman, 1984; Ghori and Luckwill, 1985; Harman et al, 1992), gait and posture (Kinoshita, 1985; Martin and Nelson, 1986) and ground reactions (Kinoshita, 1985; Harman et al, 1992; Tilbury-Davis and Hooper, 1999). Most of these studies have considered the load being carried and its distribution on the torso. However, little is known about the biomechanical effects of a backpack's suspension characteristics. The coupling

between a backpack and the wearer is dynamic rather than static, due to the inertial properties of the pack and the compliance of the shoulder straps, the hip belt and the pack frame, which lead to relative motion of the pack with respect to the trunk.

It is well known that the muscle-tendon complexes function as springs, which absorb, store and return energy during different parts of the gait cycle, which leads to substantial savings in energy expenditure (Alexander, 1988). This type of internal elastic mechanism is probably also responsible for the surprising energy efficiency of carrying loads on the head, which is widespread in Africa (Alexander, 1986). External elastic mechanisms can also be used for load carriage. An example of this is the use of springy bamboo or wooden poles, which are employed by people throughout Asia to carry loads anterior and posterior to the body. The use of compliant poles when running has been investigated experimentally (Kram, 1991). Although it was found that carrying loads with compliant poles is not particularly energy efficient, it offers important biomechanical advantages by reducing the peak shoulder forces and loading rates. Achieving similar advantages with a backpack would be more difficult because the motion of a pack is more closely coupled with that of the trunk, which is important for the bearer's balance and agility. Conversely, pole-carried loads are unwieldy and free to swing beneath the poles in a way that reduces the user's balance and agility.

When walking or running, the backpack and the human torso interact in a dynamic way as a result of the cyclic motion of the torso. Therefore, to investigate the effects of different backpack suspension characteristics requires a biomechanical model of human gait, a dynamic model of the backpack suspension system, and a model of the way in which the backpack's characteristics affect gait. Previous experimental studies have found that the human gait pattern is affected by backpack load, the load distribution, the backpack type and

the bearer's gender (Martin and Nelson, 1986). To theoretically predict the effects of different pack designs, ideally we would need to be able to predict how the human nervous system will adjust the gait in response to the backpack behaviour, presumably in pursuit of energy efficiency, comfort or acceptable injury risk. However, this is extremely difficult to achieve by computational means due to the complexity of the human musculoskeletal system and the ambiguity of the underlying control strategies (Yamaguchi, 1990; Pandy, 2001).

The solution adopted by the authors is to use a combination of computational modelling and gait measurement. Gait measurement is used to determine the effect of those factors not being investigated (load, load distribution, and gender). The resulting gait kinematics are then used as input data for the computational modelling, in other words, any changes in gait patterns as a result of changing pack suspension characteristics are neglected. However, by using inverse dynamics, it is possible to predict the changes in joint forces and moments, ground reactions and mechanical work. In this way, the possible effects of different backpack suspension characteristics on locomotion efficiency, comfort and the risk of injury can be investigated.

Methods

Gait modelling

In this study, a whole body, multi-segment gait model has been adopted (Figure 1), which includes the head, torso, pelvis, both arms (forearms and humeri), and both legs (thighs, shanks and feet). All body segments are assumed to be rigid and their motions are modelled in the sagittal plane only. Anthropometric data for each body segment are based on Leva,

1996, and have been modified for the forearm and torso segments to allow for the different segment definitions used in this work (see Table 1).

An inverse dynamics approach has been adopted, where the measured motions of all the major body segments, while carrying load, are given as the only input data. This differs from the conventional application of inverse dynamics used in gait laboratory studies, where the ground reactions are measured using force plates and are inputs to the calculations (Winter, 1990; Siegler and Liu, 1997). Basing the inverse dynamics on measured kinematic data only means that the simulation can be used to predict changes in the joint forces, joint moments and ground reactions, as a result of proposed changes to a backpack's dynamic characteristics.

Whole body inverse dynamics combined with force plate measurements provides a redundancy of data, which can be used to improve the estimates of joint loads (Kuo, 1998). However, in this study, the aim is to predict the changes in the ground reactions and the joint loads as a result of hypothetical changes in a backpack's suspension characteristics. Therefore, force plate measurements are only relevant as a means of validating the gait modelling, prior to using it as a predictive tool.

The equations of motion of the i^{th} body segment, in the sagittal plane, can be expressed as

$$\begin{cases} m_i \cdot \vec{a}_i = \sum_{j=1}^{n_i} \vec{F}_{ji} + \vec{F}_{ei} + m_i \cdot \vec{g} \\ I_i \cdot \alpha_i = \sum_{j=1}^{n_i} M_{ji} + M_{ei} + \sum_{k=1}^{n_i} M_{ki} \end{cases} \qquad (1a, 1b)$$

where $\vec{F}_{ei}$ and M_{ei} are the resultant external force and moment acting on the segment (e.g. ground or backpack reactions). $\vec{F}_{ji}$ and M_{ji} are the resultant force and the net muscle

moment at the j[th] joint. M_{ki} is the moment caused by the resultant joint force at the k[th] joint. The segment has n_i joints connecting it to other segments.

By combining the equations of motion of all the body segments, the sums of all the external forces and moments can be expressed as

$$\begin{cases} \sum_{i=1}^{l} \vec{F}_{ei} = \sum_{i=1}^{n} m_i \cdot (\vec{a}_i - \vec{g}) \\ \sum_{i=1}^{m} M_{ei} = \sum_{i=1}^{n} I_i \cdot \alpha_i - \sum_{i=1}^{n} \sum_{k=1}^{n_i} M_{ki} \end{cases} \quad (2a, 2b)$$

where *l* and *m* are the number of external forces and moments, and n is the number of body segments.

When walking with a backpack, the major external forces and moments acting on the human body, other than gravity, are the ground reactions and the pack interface force and moment. Therefore, Equations (2) can be rewritten as follows:

$$\begin{cases} \sum \vec{F}_{gi} = \sum_{i=1}^{n} m_i \cdot (\vec{a}_i - \vec{g}) - \vec{F}_p \\ \sum M_{gi} = \sum_{i=1}^{n} I_i \cdot \alpha_i - \sum_{i=1}^{n} \sum_{k=1}^{n_i} M_{ki} - M_p \end{cases} \quad (3a, 3b)$$

where $\vec{F}_{gi}$ and M_{gi} are the ground reactions, and $\vec{F}_p$ and M_p are the pack force and moment.

Once the pack forces and moments have been calculated (see backpack modelling section), the sum of the ground reaction forces can be determined from the motions and inertial properties of the body segments, using Equation (3). Therefore, during the swing phase, the ground reaction force acting on the single supporting foot can be obtained directly. This fact

has previously been used to derive the vertical ground reaction force during running, which has no double support phase (Bobert, Schamhardt and Nigg, 1991). However, during the double support phase of walking, when both legs and the ground surface form a closed-loop, the problem of determining the ground reaction forces under each foot becomes indeterminate. In order to solve this redundant problem, some simple linear relationships, based on empirical data, have been used to model the transfer of the ground reactions from one foot to the other during the double support phase. The transfer of the three ground reaction components are modelled as follows:

- The ratio of the vertical ground reaction force on the heel-strike foot to the sum of the vertical forces on both feet varies linearly during the double support phase.
- The ratio of the horizontal ground reaction force to the vertical ground reaction force on the toe-off foot varies linearly during the double support phase.
- The ratio of the centre of pressure position for the heel-strike foot to the sum of the centre of pressure positions for both feet varies linearly during the double support phase.

From Figure 2, it can be seen that the linear transfer assumptions are in good agreement with published ground reaction measurements (Winter, 1990). Symmetry of the right and left limbs has been assumed.

During gait simulation, the equations and linear transfer assumptions described above are used in the following sequence of calculations:

1. The pack forces and moments, acting on the torso, are calculated as described in the backpack modelling section;
2. During the swing phase, the ground reaction force acting on the supporting foot is obtained directly from Equation (3a);

3. During the double stance phase, the sum of the ground reaction forces on both feet is calculated using Equation (3a); then the ground reaction forces on each foot are calculated from the linear transfer relationships;
4. Starting from the one or two supporting feet and working up, segment by segment, Equation (1a) is used to calculate the resultant force at each joint.
5. During the swing phase, the ground reaction moment acting on the supporting foot is obtained directly from Equation (3b);
6. During the double stance phase, the sum of the ground reaction moments on both feet is calculated using Equation (3b); then the ground reaction moments on each foot are calculated from the linear transfer relationship for the centres of pressure;
7. Starting from the one or two supporting feet and working up, segment by segment, Equation (1b) is used to calculate the net muscle moment at each joint.

This rather complex sequence of calculations is a result of the fact that the ground reactions are calculated rather than measured, the latter being the case in the traditional application of inverse dynamics. Also, because the joint forces are inputs to Equation (3b), they must first be calculated using Equation (1a), segment by segment, before the ground reaction moments can be calculated.

Gait measurement

To provide kinematic input data for gait simulation, three-dimensional gait measurements have been conducted to capture whole body motion while carrying a backpack. Two healthy male subjects were selected from a population of postgraduate students. Prior to participation, the subjects provided informed consent in accordance with the policies of Salford University's Ethical Advisory Committee. The subjects walked in bare feet, inside a gait

laboratory, while motion data was collected at 120Hz using a 6-camera Vicon motion analysis system (Oxford Metrics Limited, Oxford, UK). Two experimental conditions were used, walking with no load at a self-selected velocity, and walking with a backpack load of 10kg at a self-selected velocity. Each experimental condition was measured six times to ensure that a repeatable data set for a complete walking cycle was obtained.

For each subject, the movements of the 13 major body segments, defined in the preceding section, were recorded. Specially designed plastic plates, each carrying four reflective markers, were attached to each body segment. A helmet was used to carry the four markers on the head. An elastic hip belt was used to firmly locate the plastic plate carrying the four markers on the pelvis. The plastic plates and the helmet eliminate the relative motion between the cluster of four markers on a segment, thus increasing the accuracy of the recorded motion data.

To describe the segment positions and orientations in a standardised way, anatomical landmarks and a bone-embedded anatomical reference systems are defined for each major body segment. These landmarks and reference frames are based mainly on the recommendations of Cappozzo et al., 1995, and Van der Helm and Pronk, 1995, with small adaptations to suit the special requirements of this study. Before the walking trials, a set of calibration procedures was used to locate the anatomical landmarks using the calibrated anatomical system technique, or CAST (Cappozzo et al., 1995).

In this study, the shoulder joint centre is defined to be the functional humerothoracic joint centre, which is the effective centre of rotation between the upper arm and the trunk. As movement of the shoulder involves compound motions of the humerus, scapula and clavicle,

it is unlikely that the centre of rotation is located at the centre of the humeral head. Therefore, a functional approach (Cappozo, 1984; Leardini et al, 1999) has been used to establish the humerothoracic joint centre, as well as the hip joint centre. A closed-form algorithm is employed to estimate the joint's centre of rotation (Gamage and Lasenby, 2002), which does not require manual adjustment of optimisation parameters. The positions of other joint centres were determined directly from anatomical landmarks, for example, the knee joint centre coincides with the midpoint between the lateral epicondyle and medial epicondyle.

The raw output data from the Vicon Workstation software was passed to SMAS (Salford Motion Analysis Software), a MATLAB based software package for three-dimensional motion analysis, which can perform kinematic and kinetic analyses for general articulated multi-body systems. Missing frames are dealt with by a fill-gap procedure and the maximum gap that can be filled is 15 consecutive frames. If, after applying the fill-gap procedure, there were still missing data, then that trial was discarded. After fill-gap processing, the data were filtered using a low pass fourth-order Butterworth digital filter with a cut-off frequency of 6 Hz. The timings of foot contact events (heel strike and toe off) were determined using foot marker kinematics (Mickelborough et al., 2000).

The use of the CAST technique and functional joint centre location, for the shoulder and hip, provides an accurate and effective approach for whole body gait measurement in three-dimensions. The derived sagittal gait kinematics are more accurate than those obtained using the traditional 2-D measurement method (Winter 1990; Hong and Cheung 2003).

Backpack modelling

Modelling the interaction between pack and torso is particularly difficult because of the nonlinear properties and redundancy of pack suspension systems, the difficulty of measuring the pack interface forces, and the complex relative motion between pack and body. This problem is compounded by gait and posture changes made by the human nervous system in response to the backpack's effect on physiological factors, such as joint and muscle forces, skin pressure, and fatigue. Because of these difficulties, there has been limited research activity in the area of backpack modelling (Pelot et al., 2000). For the work reported here, a dynamic model has been developed, utilising a non-linear pack suspension equation, which describes a backpack's dynamic response to trunk motions.

To describe backpack kinematics in the sagittal plane, two moving coordinate systems are defined, the backpack system $x_p o_p y_p$ and the trunk system $x_t o_t y_t$ (see Figure 3). It has been assumed that, under normal conditions, the shoulder straps and waist belt prevent rotation of the backpack relative to the trunk, in the sagittal plane, and translation along the x_p axis relative to the trunk. In this case, if internal deformation of the backpack is neglected, the pack can be modelled as a rigid body that can slide along the back, but is otherwise constrained to move with the trunk. Thus, the pack kinematics can be described as follows:

$$\begin{cases} \ddot{x}_p = \ddot{x}_t \cdot \sin\theta_t - \ddot{y}_t \cdot \cos\theta_t - \alpha_t \cdot (d_y + u) + \omega_t^2 \cdot d_x - 2\omega_t \cdot \dot{u} \\ \ddot{y}_p = \ddot{x}_t \cdot \cos\theta_t + \ddot{y}_t \cdot \sin\theta_t - \alpha_t \cdot d_x - \omega_t^2 \cdot (d_y + u) + \ddot{u} \\ \alpha_p = \alpha_t \end{cases} \quad \text{(4a, 4b, 4c)}$$

where $\ddot{x}_p$, $\ddot{y}_p$ and α_p are the linear and angular accelerations of the pack; $\ddot{x}_t$ and $\ddot{y}_t$ are the linear accelerations of the torso mass centre; θ_t, ω_t, and α_t are the angular displacement, velocity and acceleration of the torso; and u, $\dot{u}$ and $\ddot{u}$ are the displacement, velocity and acceleration of the pack suspension system (i.e. relative motion in the y_p direction).

By considering the forces and moments acting on the pack and applying the Newton-Euler equations, the pack dynamics can be described as follows:

$$\begin{cases} m_p \cdot \ddot{x}_p = F_{xp} + m_p \cdot g \cdot \cos\theta_t \\ m_p \cdot \ddot{y}_p = F_{yp} - m_p \cdot g \cdot \sin\theta_t \\ I_p \cdot \alpha_p = M_{zp} \end{cases} \quad \text{(5a, 5b, 5c)}$$

where F_{xp} and F_{yp} are the normal and tangential pack interface forces; and M_{zp} is the pack interface moment about the pack mass centre.

The pack moment can be obtained directly from Equations (4c) and (5c), and the angular acceleration of the torso. The pack force in the normal direction can be obtained by combining Equations (4a) and (5a), which leads to

$$F_{xp} = m_p \cdot (\ddot{x}_t \cdot \sin\theta_t - \ddot{y}_t \cdot \cos\theta_t - \alpha_t \cdot (d_y + u) + \omega_t^2 \cdot d_x - 2\omega_t \cdot \dot{u} - g \cdot \cos\theta_t) \quad (6)$$

So, given the motion of the torso and the motion of the pack relative to the torso (u and $\dot{u}$), F_{xp} can be calculated.

A non-linear suspension model is used to describe the relationship between tangential pack force and relative pack motion. Elastic, damping and inertial effects are allowed for by including three cubic polynomials in u, $\dot{u}$ and $\ddot{u}$ respectively. The cubic polynomials enable non-linear characteristics to be modelled. Thus, the pack suspension model can be written as

$$\begin{aligned} F_{yp} = {} & c_3 \cdot \ddot{u}^3 + sign(\ddot{u}) \cdot c_2 \cdot \ddot{u}^2 + c_1 \cdot \ddot{u} + \\ & b_3 \cdot \dot{u}^3 + sign(\dot{u}) \cdot b_2 \cdot \dot{u}^2 + b_1 \cdot \dot{u} + a_3 \cdot u^3 + sign(u) \cdot a_2 \cdot u^2 + a_1 \cdot u \end{aligned} \quad (7)$$

where a_1, a_2, a_3, b_1, b_2, b_3, c_1, c_2, c_3 are the constant suspension parameters, which depend on the type of pack, how it is loaded, and the adjustment of shoulder straps and waist belt.

For a particular backpack, the parameters of the suspension model can be identified from dynamic test data, obtained using the hydraulically driven load carriage test-rig shown in Figure 4 (Gretton and Howard, 2000). Harmonic analysis techniques have been used to identify the suspension parameters from the measured motion and force data (Gretton, 2003).

Substituting Equations (4b) and (7) for $\ddot{y}_p$ and F_{yp} in Equation (5b) leads to the following non-linear differential equation:

$$F_{yp} + m_p(\omega_t{}^2 u - \ddot{u}) = m_p(\ddot{x}_t \cos\theta_t + \ddot{y}_t \sin\theta_t - \alpha_t d_x - \omega_t{}^2 d_y + g\sin\theta_t) \tag{8}$$

Therefore, given the measured data describing torso motion ($\ddot{x}_t, \ddot{y}_t, \theta_t, \omega_t, \alpha_t$), a numerical integration algorithm can be used to solve Equation (8) and thereby obtain the relative pack motion (u , $\dot{u}$ and $\ddot{u}$). In this study, a 4th order Runge-Kutta algorithm has been employed. Because the numerical integration time step is normally smaller than the gait measurement interval, cubic interpolation is used to provide torso motion data at the necessary frequency. The initial values of pack suspension displacement and velocity are set to zero, and the simulation runs for repeated gait cycles until a steady state pack motion cycle is achieved.

Thus, given the measured torso motions, the relative pack motion can be determined from Equation (8), and then the pack forces and moment can be calculated from Equations (6), (7) and (5c). These can be described as an equivalent force system acting at the torso's mass centre, as follows:

$$\begin{cases} F_{x\_pack} = -(F_{xp} \cdot \sin\theta_t + F_{yp} \cdot \cos\theta_t) \\ F_{y\_pack} = F_{xp} \cdot \cos\theta_t - F_{yp} \cdot \sin\theta_t \\ M_{z\_pack} = F_{x\_pack} \cdot (d_y + u) + F_{y\_pack} \cdot d_x - M_{zp} \end{cases} \tag{9}$$

This force system is then applied to the torso in the whole body gait model. In this way, the combined simulation of gait and pack dynamics is achieved, which can be used to investigate the biomechanical effects of different backpack suspension dynamics.

Results

The combined dynamic simulation of the human-pack system, using the methods described above, has been implemented in the MATLAB programming environment. The kinematic input data for the whole body gait model was obtained in the gait laboratory, using the methods described earlier. The simulation results described here were produced using the data for just one of the two gait laboratory subjects (a healthy male of age 30 and weight 75kg).

In the results presented here, a simple pack suspension model has been considered, which includes only linear elastic and linear damping components. Initial values for stiffness and damping coefficient were estimated from load carriage test-rig data for a military backpack carrying a 10kg load. Figure 5 shows the simulation results, over one gait cycle, for relative pack displacement and for the pack interface forces and moment acting on the torso. The horizontal pack force varies around a mean value of approximately zero; whether it is tensile or compressive being largely dependent on the angular acceleration of the torso. The vertical pack force is compressive and fluctuates around a mean value, which is equal to the weight of the backpack. The pack moment acting on the torso is counter clockwise over the whole gait cycle due to the pack's position, posterior to the trunk. Figures 6 to 8 show the calculated ground reactions and the joint loads at the ankle, knee and hip over one gait cycle. The results

with a 10kg backpack load are compared with the unloaded case. The horizontal, as well as vertical, ground reactions and lower limb joint forces are increased, which is consistent with experimental studies (Kinoshita, 1985; Harman et al, 1992).

The effects of different backpack suspension characteristics on locomotion energetics was assessed by varying the stiffness and damping coefficient in the suspension model. Mechanical energy expenditure (MEE) has been used to represent the energy expended in walking. MEE is calculated by integrating the absolute values of the joint powers, at all the major joints, over the whole gait cycle (Aleshinsky 1986a, 1986b; Zatsiorsky, 2002). Figure 9 shows the calculated MEE over one gait cycle with different pack suspension characteristics. It can be seen that the mechanical energy expenditure decreases with decreasing stiffness and increasing damping ratio, however, the differences are negligible.

The effects of different backpack suspension characteristics on pack interface forces and moment and on joint loads were also investigated (Figures 10-11). It was found that decreasing the suspension stiffness significantly reduces the peak values of vertical pack force, acting on the torso, which has important implications for the skin pressures under the shoulder straps and waist belt, and for the risk of injury when heavy loads are carried. The effect on horizontal force and moment is smaller. The peak values of the vertical forces at the lower limb joints are also moderated, however the effect is much smaller because the backpack load is relatively small compared with body weight. With larger backpack loads, the advantages of lowering suspension stiffness would be more pronounced.

Discussion

A methodology has been introduced for studying the biomechanics of load carriage and, in particular, the effects of a backpack's suspension characteristics on human locomotion. A combination of computational modelling and gait measurement has been adopted, where gait measurement is used to determine the effect of those factors not being investigated (e.g. load, load distribution, and gender). The resulting gait kinematics are then used as input data for the computational modelling, which predicts the effects of different backpack suspension characteristics on joint forces and moments, ground reactions and mechanical work. In this way, the possible effects of different suspension characteristics on locomotion efficiency, comfort and the risk of injury can be investigated. This approach neglects any changes in gait patterns as a result of changing pack suspension characteristics, which are extremely difficult to predict by computational means because of the complexity of the musculoskeletal system and the lack of understanding of human motor control (Yamaguchi, 1990; Pandy, 2001).

Although the skin pressures under the shoulder straps and hip belt have been measured (Holewijn, 1990; Martin and Hooper, 2000), it has not yet been possible to measure the resultant pack forces and moment acting on the torso because of the complexity of the interface between pack and body. In this study, the net effect of the pack interface is represented by a non-linear suspension model, which relates the resultant pack forces and moment to the torso motions, and can be identified from load carriage test-rig experiments. Combined with a whole body, inverse dynamics model of gait, this approach allows the ground reactions and the loads on the musculoskeletal system to be predicted for hypothetical pack suspension characteristics. Moreover, a wide range of suspension characteristics can be studied by simulation, where an experimental approach would not be feasible.

The simulation results show that the linear suspension characteristics investigated (stiffness and damping) have a negligible effect on the mechanical energy expenditure (MEE). This agrees with the experimental results for load carriage with springy poles (Kram, 1991), which showed that adopting a very compliant suspension had no obvious effect on metabolic energy cost (oxygen consumption rate). However, it should be noted that the mechanical energy expenditure (MEE) is calculated by integrating the absolute values of the joint powers, at all the major joints, over the whole gait cycle (Aleshinsky 1986a, 1986b). Therefore, when used as a measure of the total mechanical work done by the muscles, MEE assumes single-joint muscles and no co-contractions (Prilutsky et al., 1996). However, if the energy transfer and recovery associated with biarticular muscles and elastic elements, such as tendons, is considered, MEE may well be an overestimate of the mechanical muscle work.

Aside from energetics, decreasing backpack suspension stiffness has some very important biomechanical advantages. The fluctuation in the vertical force acting on the torso is significantly reduced. Moreover, the peak values of ground reaction forces and lower limb joint loads are also reduced. Therefore, a soft pack suspension could reduce the risk of tissue and nerve damage (rucksack palsy), under shoulder straps and hip belts, and also of back and lower limb injuries. This could be particularly relevant when heavy backpack loads are carried, as it has been found that some peak joint forces increase disproportionately with increasing pack load (Goh et al., 1998).

However, a more compliant pack suspension will result in larger pack motions relative to the torso, which would affect the user's balance and agility. It may be possible to overcome this problem with the right combination of stiffness and damping, where the damping is chosen to

allow a compliant suspension without excessive relative motion. Future investigations will also examine the effects of non-linear suspension characteristics.

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Figures and Tables

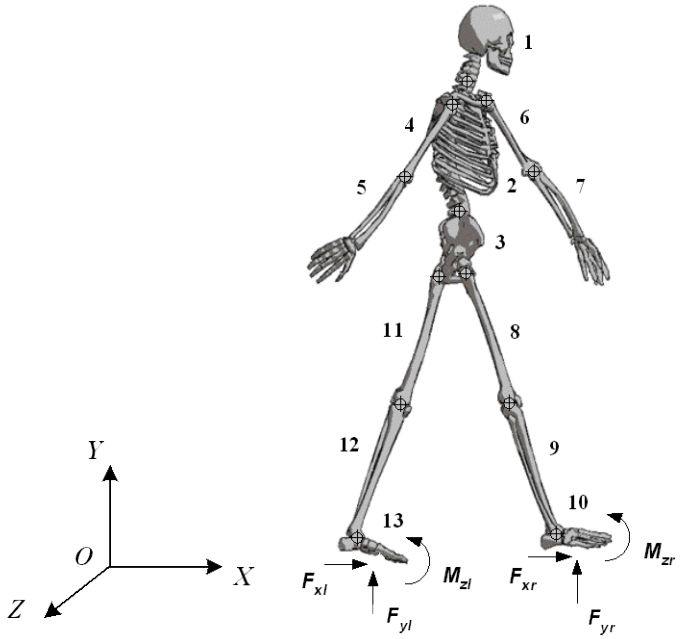

Figure 1 The whole body model with 13 segments and 12 connecting joints

Table 1 Anthropometric data for whole body model

Segment	Proximal endpoint	Distal endpoint	Mass (%)	Mass centre position (%)	Radius of gyration (%)
Head	C7	Vertex	6.94	50.02	30.3
Trunk	Omphalion	C7	32.29*	49.85*	33.74*
Pelvis	Midpoint hip joint centres	Omphalion	11.17	61.15	61.5
Humerus	Shoulder joint centre	Elbow joint centre	2.71	57.72	28.5
Forearm	Elbow joint centre	Wrist joint centre	2.23*	67.5*	43.88*
Thigh	Hip joint centre	Knee joint centre	14.16	40.95	32.9
Shank	Knee joint centre	Ankle joint centre	4.33	43.95	25.1
Foot	HEEL	2nd Metatarsal	1.37	44.15	25.7

* Adjusted values based on original data.

Masses are percentages of body mass, and mass centre positions (from the proximal end) and radii of gyration are percentages of segment length.

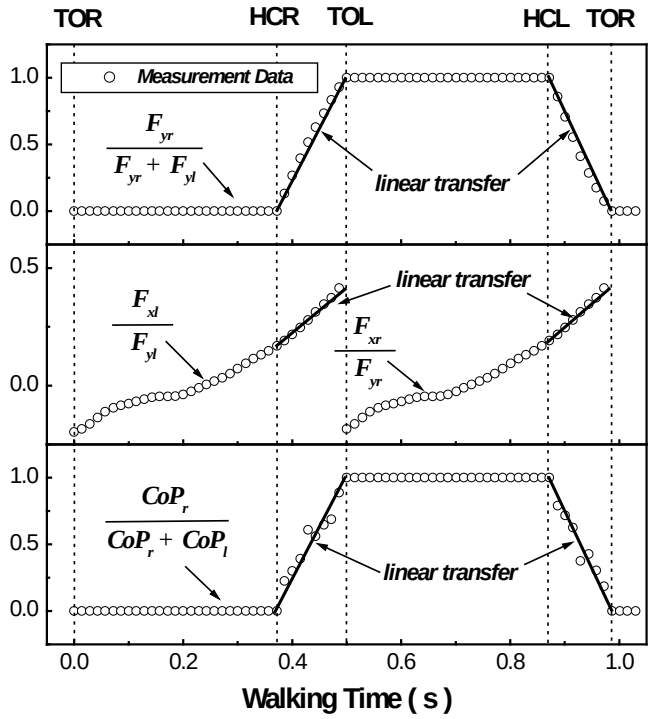

Figure 2 Calculated transfer ratios (solid line), based on linear assumptions, compared with measurement data from Winter (1990).

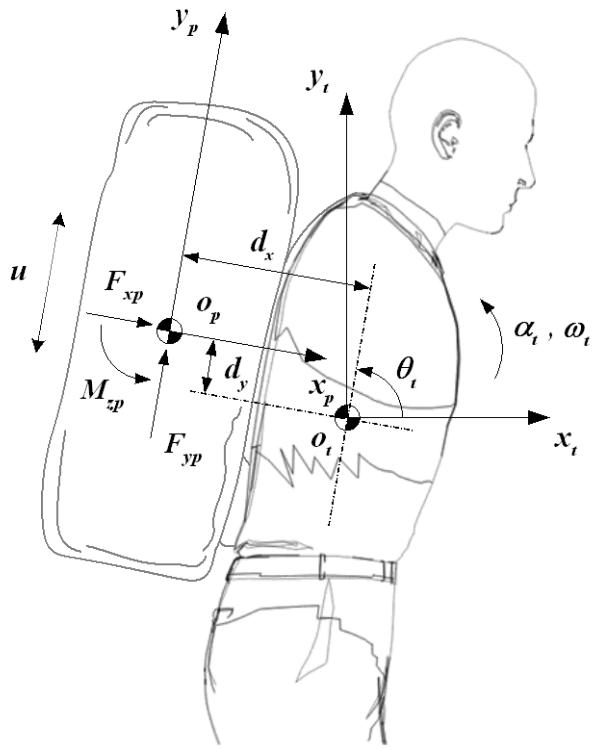

Figure 3 Pack interface forces and moment, and the pack and trunk local coordinate systems

Figure 4 Dynamic load carriage test-rig

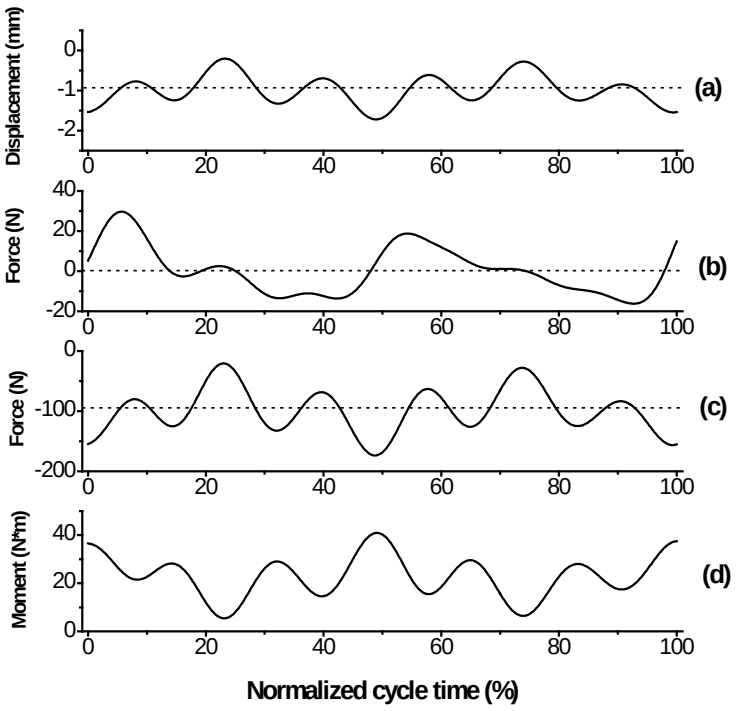

Figure 5 Calculated time history of relative pack motion (a), vertical pack force (b), horizontal pack force (c) and pack moment (d) over one gait cycle.

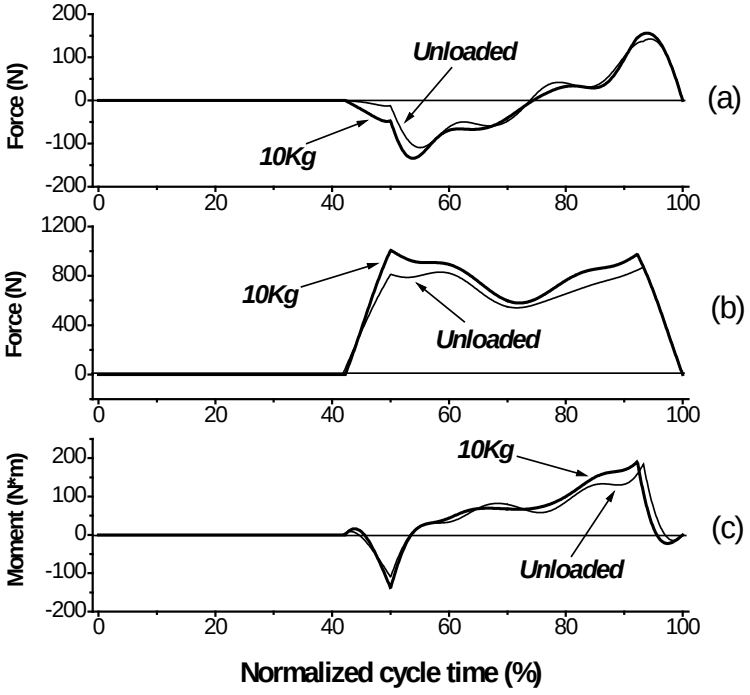

Figure 6 Comparison of calculated horizontal ground force (a), vertical ground force (b) and ground reaction moment (c) over one gait cycle.

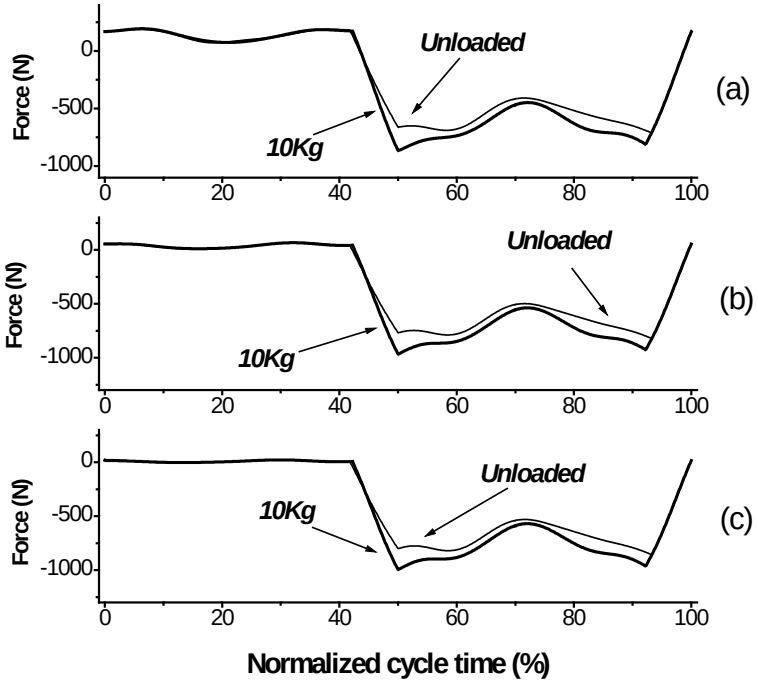

Figure 7 Calculated vertical joint loading at hip (a), knee (b) and ankle (c) over one gait cycle.

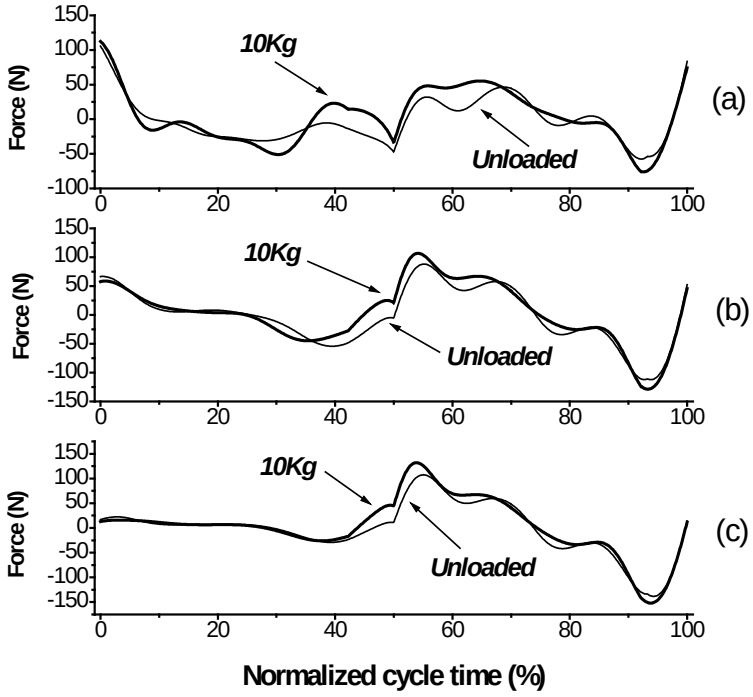

Figure 8 Calculated shear joint force at hip (a), knee (b) and ankle (c) over one gait cycle.

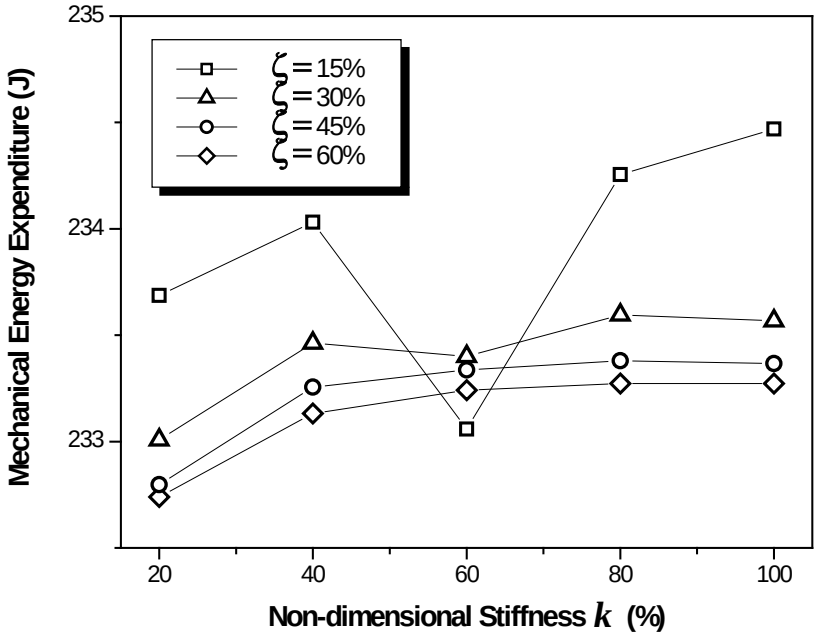

Figure 9 Calculated mechanical energy consumption over one gait cycle with different pack suspension characteristics

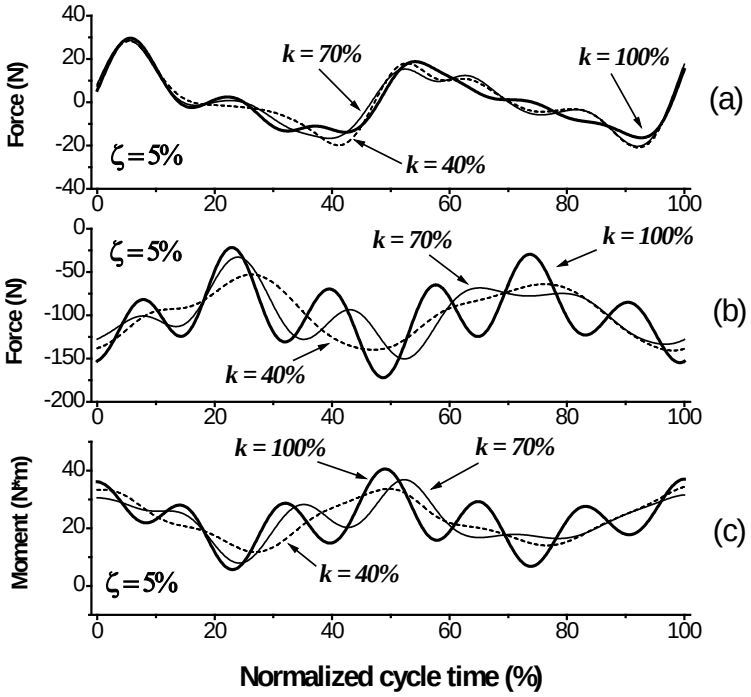

Figure 10 Calculated time history of horizontal force (a), vertical force (b) and resultant moment (c) exerted on torso by pack over one gait cycle with different pack suspension stiffness.

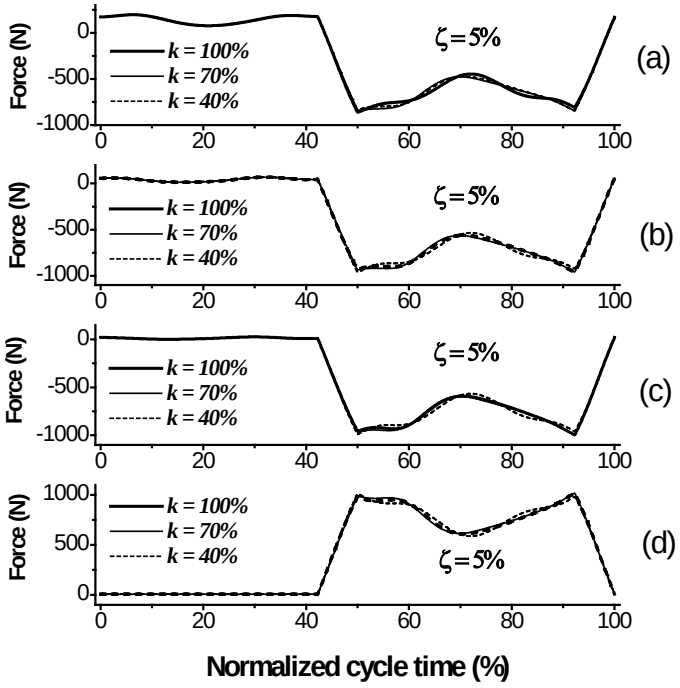

Figure 11 Calculated vertical force at hip (a), knee (b), ankle (c) and ground (d) over one gait cycle with different pack suspension stiffness.

Figure Captions

Figure 1 The whole body model with 13 segments and 12 connecting joints. The hand and forearm are considered to be one segment, as the relative motion of the hand with respect to the forearm is small. The global coordinate system is defined thus: X-axis lies in the sagittal plane and points in the direction of forward progress, Y-axis also lies in the sagittal plane and points upwards, and Z-axis lies in the frontal plane and points to the right with respect to the direction of forward progression

Figure 2 Calculated transfer ratios (solid line), based on linear assumptions, compared with measurement data from Winter (1990). F_{xr}, F_{yr}, F_{xl} and F_{yl} are the horizontal and vertical ground forces at the right and left foot. CoP_r and CoP_l are centres of pressure for right and left foot. CoP is defined as ground reaction moment about the ankle joint divided by vertical ground force M_z/F_y. In the double support phase from right heel contact (HCR) to left toe off (TOL), the vertical force transfer ratio r_{t_fy} increases from 0 to 1, the horizontal force transfer ratio r_{t_fx} increases from $r_{t\_fx}^{(HC)}$ to $r_{t\_fx}^{(TO)}$, while the CoP transfer ratio r_{t_cop} increases from 0 to 1.

Figure 3 Pack interface forces and moment, and the pack and trunk local coordinate systems. The origins, o_t and o_p, are located at the mass centres of the trunk and the backpack respectively. The directions of the x_t and y_t axes coincide with the X and Y axes of the global coordinate system. The y_p axis is parallel with the trunk's longitudinal axis (the line connecting the waist joint to the neck joint of the whole body gait model), with the x_p axis pointing towards the trunk.

Figure 4 Dynamic load carriage test-rig. The hydraulic ram drives the mannequin up and down at different frequencies and amplitudes. Accelerometers measure the motion of the backpack and the mannequin, and a load cell measures the dynamic force propelling the mannequin.

Figure 5 Calculated time history of relative pack motion (a), vertical pack force (b), horizontal pack force (c) and pack moment (d) over one gait cycle. The pack load was 10 Kg and the walking speed was 1.12 m/s.

Figure 6 Comparison of calculated horizontal ground force (a), vertical ground force (b) and ground reaction moment (c) over one gait cycle. The thick line is the loaded case (10 Kg) at a walking speed of 1.12 m/s. The thin line is the unloaded case at a walking speed of 1.17 m/s.

Figure 7 Calculated vertical joint loading at hip (a), knee (b) and ankle (c) over one gait cycle. The thick line is the loaded case (10 Kg) at a walking speed of 1.12 m/s. The thin line is the unloaded case at a walking speed of 1.17 m/s.

Figure 8 Calculated shear joint force at hip (a), knee (b) and ankle (c) over one gait cycle. The thick line is the loaded case (10 Kg) at a walking speed of 1.12 m/s. The thin line is the unloaded case at a walking speed of 1.17 m/s.

Figure 9 Calculated mechanical energy consumption over one gait cycle with different pack suspension characteristics. The backpack was loaded at 10 Kg and the walking speed was

1.12 m/s. The non-dimensional pack stiffness, *k*, is defined as the ratio of the simulated pack stiffness *K* to the test pack stiffness K_0. The non-dimensional damping ratio, ζ, is defined as $c/2\sqrt{m_P K}$, where *c* is the pack suspension's damping coefficient

Figure 10 Calculated time history of horizontal force (a), vertical force (b) and resultant moment (c) exerted on torso by pack over one gait cycle with different pack suspension stiffness. The backpack was loaded at 10 Kg and the walking speed was 1.12 m/s.

Figure 11 Calculated vertical force at hip (a), knee (b), ankle (c) and ground (d) over one gait cycle with different pack suspension stiffness. The backpack was loaded at 10 Kg and the walking speed was 1.12 m/s.