BIOMECHANICAL ANALYSIS OF SIT-TO-STAND MOTION IN CHILDREN WITH BACKPACK LOAD

by

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ABSTRACT

BIOMECHANICAL ANALYSIS OF SIT-TO-STAND MOTION IN CHILDREN WITH BACKPACK LOAD

Sit-to-stand (STS) motion is a highly coordinated and energy demanding task of daily activities. The primary objective of this study was to investigate the effects of back load on the sagittal plane kinematics and kinetics of STS motion in healthy children. The secondary objectives were to determine the limbs which may be more prone to damage and to suggest a critical value of back load relative to the body weight. Fifteen healthy children (8 males, 7 females, mean age 9.6 ± 1.2) participated in the study to perform STS motion in three conditions: (1) with no back load (2) with a back load of 10% of the body weight (BW) and (3) with a back load of 20% of the BW. The motion was performed using a fixed bench height at a self-selected speed. Kinematic and kinetic data were collected via a 6-camera motion analysis system and 2 force plates.

The present results led us to four major conclusions reflecting the effects of back load on the STS motion: (1) The neuromuscular system is concluded to adjust the durations of the individual phases rather than that of total STS in order to adapt the motion to the new mechanical conditions. (2) Subjects followed a “trunk flexion strategy” in the loaded cases by shifting the new center-of-mass both forward and downward presumably to ease the control of the motion and to reduce the risk of falling. (3) Different back load levels affect different joints. Increasing the load to 20% BW caused significantly higher ankle moment. On the other hand, even 10% BW load produced significantly higher knee moment. However, no major effect of back load was shown on hip moment and power. (4). Back loading causes higher forces and increases the range of eccentric activity of gastrocnemius and soleus muscles by leading to much higher angles of ankle dorsiflexion. Therefore, the calf muscles and the achilles tendon, were concluded to be the most prone elements of the muscle-tendon complexes of the lower extremity to damage while performing STS motion with back load.

Keywords: Sit-to-Stand Motion, Backpack Load, Children, Motion Analysis

ÖZET

ÇOCUKLARDA SIRT ÇANTASI YÜKÜ İLE AYAĞA KALKMA HAREKETİNİN BİYOMEKANİK ANALİZİ

Ayağa kalkma hareketi günlük hareketler içinde en çok nöromasküler koordinasyon ve enerji gerektirenlerinden biridir. Bu çalışmanın birincil hedefi bilimsel olarak izin verilen aralıktaki sırt çantası yükünün 10 yaşındaki çocukların ayağa kalkma hareketinde alt ekstremitelerin yanal düzlem kinematik ve kinetiğine olan etkilerini incelemektir. İkincil amaçlar ise hasar görme riskindeki uzuvların belirlenmesi ve bunlara tehlike arz edebilecek yük miktarının belirlenmesidir. Deney kurgusunda, ortalama olarak 9.6 ± 1.2 yaşındaki 15 sağlıklı denekten 3 değişik koşulda ayağa kalkma hareketini uygulaması istenmiştir. Bu koşullar: (1) Sırt çantası olmadan (2) Vücut ağırlığının 10%'u kadar yüklenmiş sırt çantası ile (3) Vücut ağırlığının 20%'si kadar yüklenmiş sırt çantası ile. Ayağa kalkma hareketi okul sırası boyutlarına uygun sabit yükseklikteki bir oturak ile deneklerin tercih ettikleri kalkma hızında gerçekleştirilmiştir. Kinematik ve kinetik ölçümler 6 kamera ve 2 kuvvet ölçen plaka ile alınmıştır.

Sonuçlar analiz edildiğinde sırt çantası yükünün ayağa kalkma hareketine etkisi ile ilgili 4 dikkate değer sonuca ulaşıldı: (1) Nöromasküler sistem toplam kalkma süresinde bir değişiklik yapmasa da her bir fazın sürelerini ayarlayarak yüklü ayağa kalkma hareketini gerçekleştirir. (2) Denekler yüklü kalkmayı kolaylaştırmak için kalkarken ileri eğilme stratejisini izlediler. Bu da yeri arkaya kaymış olan ağırlık merkezini öne ve ileri çekerek denge kontrolünü kolaylaştırdı ve düşmeyi zorlaştırdı. (3) Değişik yük miktarları değişik eklemlerde etkili oldu. Bilekte 20%'lik, dizde ise 10%'luk yükler dikkate değer moment artışlarına sebep oldu. Kalçada ise bir farklılık gözlenmedi. (4) Yüklü kalkma sırasında artan yük ve ayak bileği dorsifleksiyonun gastrocnemius ve soleus kaslarının hem aktivitesini hem eksantirikliğini artırdığından hareketle ve iki kasın ve de bağlı bulundukları aşil tendonunun zorlanacağı söylenebilir.

Anahtar Kelimeler: Ayağa Kalkma Hareketi, Sırt Çantası Yükü, Çocuklar, Hareket Analizi

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$\mathbf{r}_i$	Position vector trajectory i
$\mathbf{r}_{i/j}$	Relative position vector i/j
$X_i(t)$	Trajectory of position vector i on X coordinate
$\mathbf{v}_i$	Velocity vector trajectory of point i
$\mathbf{v}_{i/j}$	Relative velocity vector trajectory of point i with respect to point j
$\Delta\mathbf{t}$	Time step size
ω_i	Angular velocity of point i
m	mass
M_i	Moment around i axis
a_i	Linear acceleration of point i
α_i	Angular acceleration of point i
F_i	Force vector in I direction
I_{ij}	ijth component of the moment-of- inertia tensor
P	Power

LIST OF ABBREVIATIONS

AFO	Ankle-foot orthosis
ANOVA	Analysis of variance
BW	Body weight
COM	Center of mass
EMG	Electromyography
FIR	Finite impulse response
GRF	Ground reaction forces
GLM	Gluteus Maximus
GM	Gastrocnemius
HMS	Hamstrings
LPS	Lumbar Paraspinal muscles
LSM	Link-segment model
QUA	Quadriceps
S.D.	Standard deviation
SCM	Sternocleidomastroid
SOL	Soleus
STS	Sit-to-Stand
TA	Tibialis anterior
TRA	Trapezius

1. INTRODUCTION

Sit-to-stand (STS) motion is one of the most frequently executed motions of daily life. However, it is not a simple task. During rising from the seated position, the body weight is transferred from a relatively stable position with a wide support base to an instable position with a much narrower support base. Therefore, above all it is a mechanically demanding motion [1]. Moreover, a highly coordinated neuromuscular control is necessary [2] in order to regulate the horizontal and vertical transfer of momentum during STS motion: the muscles of particularly the lower extremities undergo alternating activation patterns, in order to supply the joints with the required moments to realize the different phases of the motion.

Regarding both the measurement techniques applied and the implications of the results acquired, analysis of the STS motion is not as accomplished as the gait analysis. However, in several previous studies performed on adults, this motion has been addressed. The different phases of STS motion were addressed in order to analyze the STS motion [3]. Lankhorst et al. analyzed the biomechanical and electromyographic differences between two strategies of STS motion [4]. Using electromyography, Valls-Sole et al. aimed at determining the strategy dependence of activation patterns and differences between dominant muscles of STS [5]. In addition to such studies with no external load acting on the subjects` body, sitting and standing postures were analyzed when lifting a weight [6]. Strategies of arm motion during weight lifting were addressed [7]. The differences of STS motion between adults and children were analyzed [8]. The effects of obesity [9], seat height [10, 11], pregnancy [12], and position of foot [13] on STS motion were studied.

Standard STS motion was also studied on spastic children [14] as well as healthy children [8]. Studies were performed to compare the STS motion of healthy and spastic children to determine the effects of ankle foot orthosis (AFO) on the spastic children's STS motion [15].

It has been shown that on the average an elementary school student carries a backpack load of 14 to 17% of his body weight in United States [16] and 22% of his body weight in Italy [17], where one third of the students were reported to carry more than 30% of their body weight once a week. Heavy backpack use has been shown to be correlated with back strain, altered gait, bad posture, low back pain and sore muscles in the childhood [18, 19]. Heavy backpacks conceivably overload the growing joints and ligaments in young students, initiating back strain [20-22], which may lead to deformity and even pathology.

Recent studies have addressed the motion of children with back loading (applied with the use of a backpack including a predetermined weight) in activities such as walking [23], prolonged walking [24], and stair ascending as well as descending [25]. However, the biomechanical analysis of STS motion of children with back loading has not been performed.

The goal of this study is to investigate the effects of back load on kinematics and kinetics of STS motion in children. For this purpose primary school students were tested in three conditions: (1) with no back load (2) with a back load of 10% of the body weight and (3) with a back load of 20% of the body weight. The primary specific purpose of this study is to determine the joints, muscles, and tendons which may be more prone to damage and to suggest a critical value of back load relative to the body weight.

An interesting result of the earlier studies in which the STS motion of healthy children was compared to that of adults [8, 26] was that the children showed deficiency in coordination. In this study it was hypothesized that such lack of coordination is likely to increase with back loading and it may lead to instability in the joints of the lower extremity. The secondary specific purpose of this study is to test this hypothesis.

2. THEORY OF MOTION ANALYSIS AND DATA ACQUISITION

2.1 Link-Segment Model (LSM)

Link segment model is an idealization of a non-rigid body onto a rigid frame. On a rigid body, the distance between any two points remains constant regardless of the motion. However, on a non-rigid body distance between 2 points may differ. In LSM each segment of the body is defined using 3 marked points: at two nodes apart and in the middle of the limb. Note that these markers do not lie on a line, but form a plane on the limb and the lengths in between are assumed to be constant (Figure 2.1). Those planes are used to determine:

- The center of rotation of each joint,
- The linear and rotational position of each limb.

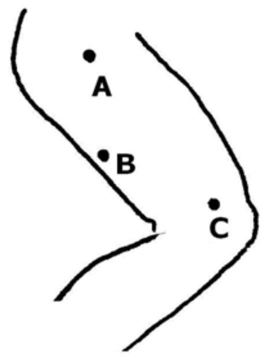

Figure 2.1 Example of marked points on a limb

LSM of the body has the following assumptions:

- Joints are frictionless pin-joints
- Segments are rigid with their mass concentrated at their center of mass
- There is no co-contraction of agonist and antagonist muscles.
- Precise positions of joints and segments can be obtained by tracking a fixed point on the skin located on the joint or segment of interest.

2.2 3D Reconstruction of Segment Positions

The components of each position vector for each point are recorded throughout the desired motion with respect to a fixed reference coordinate at a fixed sampling rate e.g. 100Hz.

$$r_A = X_A(t)\hat{I} + Y_A(t)\hat{J} + Z_A(t)\hat{K} \tag{2.1}$$
$$r_B = X_B(t)\hat{I} + Y_B(t)\hat{J} + Z_B(t)\hat{K} \tag{2.2}$$
$$r_C = X_C(t)\hat{I} + Y_C(t)\hat{J} + Z_C(t)\hat{K} \tag{2.3}$$

At each frame of the motion, using the position vectors (Figure 2.2), the relative position vectors are obtained from C to A, from C to B by

$$r_{A/C} = r_A - r_C \tag{2.4}$$
$$r_{B/C} = r_B - r_C \tag{2.5}$$

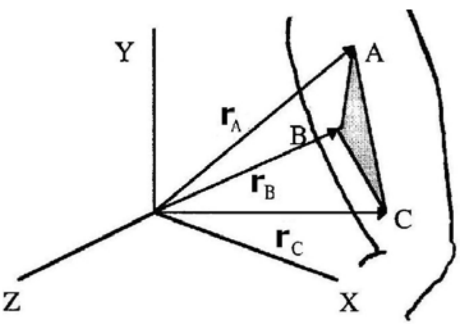

Figure 2.2 Position vectors of markers on a limb: Position vector of each point is expressed in terms of a fixed laboratory reference coordinate.

Assuming $r_{A/C}$ is in k and ($r_{b/c}$ x $r_{a/c}$) in j direction, î direction is found to be as their cross products, j x k. It is the final coordinate direction for the body segment (Figure 2.3).

Therefore, not only the linear both also the rotational positional information is obtained in 3 dimensional coordinate frame.

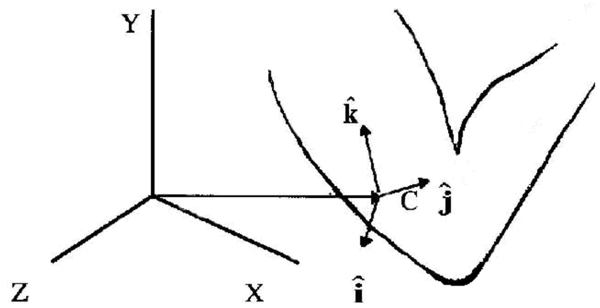

Figure 2.3 Segmental coordinate system

2.3 Joint Kinematics

In human motion analysis, the position and orientation of one body segment relative to the adjoining one should be known. That's, relative angles, lengths, velocities, and accelerations should be obtained. Using the methods in the previous section and constructing each limb on each other, a 3D body frame is obtained (Figure 2.4).

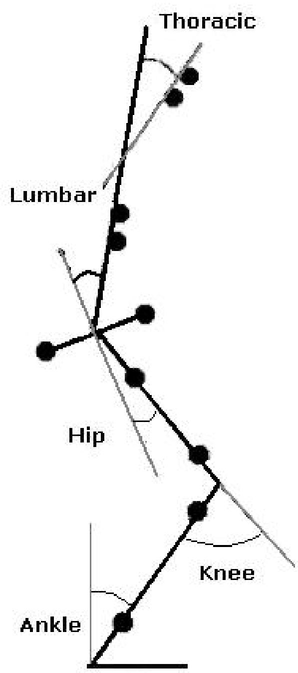

Figure 2.4 Idealized 3D frame

The relative position of one body segment to another is easier to obtain but more difficult to interpret clinically. The difficulty is in the sequence of principal rotations to obtain the desired positions. The sequences of rotations are not reversible, thus lead to different orientations. Different laboratory or motion analysis system manufacturers define the sequence of rotations in different orders, resulting in different joint angles. Care should be taken when comparison of data from one laboratory to another is inevitable in the interpretation of published data.

To ultimately calculate the moments acting on a joint, the center of the joint must be defined. It is the center of rotation. In some joints, it is hard to define a single joint center since it moves with successive movement. The geometric center of the joint can be defined as the midpoint between the femoral condyles. Although this is certainly not the kinematic or rotational joint center, it will provide a reproducible reference point for the analysis of the joint moments. It is possible to define a kinematic joint center using an instantaneous center of rotation for sagittal plane analysis or an instantaneous helical axis for general 3-D analysis.

Considering two points on a body segment, the relative position vector $r_{B/A}$ can only change orientation in space as the body segment rotates (Figure 2.5). This is how the rotation of the body is tracked. Points A and B have absolute linear velocities in space that are not equal. The linear velocity of a point is defined as the time rate of change of the position vector to that point and the velocity is defined as the derivative of the position vector with respect to time.

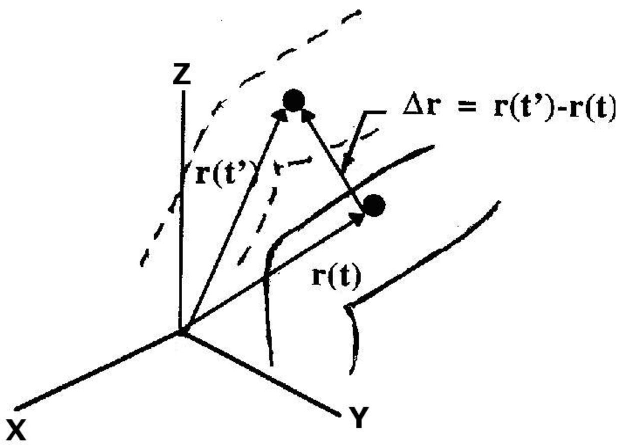

Figure 2.5 Trajectory of a position vector

Velocity of each point and relative velocity is obtained via differentiation:

$$\vec{v} = \lim_{\Delta t \to 0} \frac{\Delta \vec{r}}{\Delta t} = \frac{d\vec{r}}{dt} \tag{2.6}$$

$$\vec{v}_{B/A} = \vec{v}_B - \vec{v}_A \tag{2.7}$$

The following relation gives the angular velocity of the segment:

$$\vec{v}_{B/A} = \omega x \vec{r}_{B/A} \tag{2.8}$$

The angular velocity of the joint is the relative angular velocity of the body segment distal to the joint relative to the proximal segment. Therefore, for example, the angular velocity of the knee is:

$$\omega_{Knee} = \omega_{Tibia} - \omega_{Femur} \tag{2.9}$$

The angular velocity of the joint will have components in three directions and these components will be equal to the rate of flexion/extension, abduction/adduction, and internal/external rotation.

Analytical differentiation cannot be used in above calculations. Therefore, the data must be differentiated numerically. The velocity and acceleration are obtained by numerical differentiation of these data and are thus subject to increased noise in the calculation of the velocity and acceleration.

To completely describe the motion, the data needs to be differentiated for a second time to obtain acceleration. Therefore, the position data are filtered using digital filters to smooth the data and to obtain reliable differentiation.

Therefore; position, angular velocity, and angular acceleration data are obtained.

2.4 Joint Kinetics

In most engineering studies of dynamics, a direct dynamics problem arises, that is, the forces are known and are used to develop the differential equations of motion. These equations are usually nonlinear in nature and numerical solutions are sought. The solution involves integration of the equation; a process, which by its nature, smoothes out noise in the force data. On the other hand, most biodynamics problems involve the inverse dynamics solution, wherein the motion of the body is known and the accelerations, forces, and moments are obtained by differentiation of the position-time curves.

2.4.1 Ground Reaction Forces

The joint forces may be obtained considering free-body diagrams of each segment by solving from the most distal segment proximally (Figure 2.6). Each segment is modeled as a rigid body and as isolated from the other segments such that the reaction forces and moment still act on it.

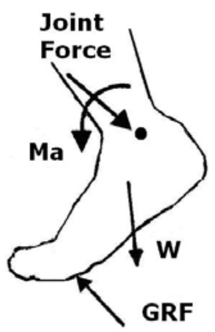

Figure 2.6 Forces on the foot: In calculations forces are obtained by solving from distal to proximal part.

2.4.2 Moments

It is aimed at the determination of the amount and the timing of force or moment generated. These quantities can be derived from the kinematics using the Newton-Euler equations:

$$\sum F_x = ma_x \qquad \sum M_x = I_{xx}a_x - \omega_y \omega_z (I_{yy} - I_{zz}) \tag{2.10}$$

$$\sum F_y = ma_y \qquad \sum M_y = I_{yy}a_y - \omega_z \omega_x (I_{zz} - I_{yy}) \tag{2.11}$$

$$\sum F_z = ma_z \qquad \sum M_z = I_{zz}a_z - \omega_x \omega_y (I_{xx} - I_{zz}) \tag{2.12}$$

The right-hand sides of Newton-Euler equations are known from motion analysis data and the forces and moments at the proximal joint of the segment are determined by solution of these equations simultaneously. One-by-one for each segment, these equations are solved based on each limb's free-body diagram from distal to proximal parts (Figure 2.7). Position, velocity, and acceleration are used together with masses and inertias to solve for the forces and moments at the joints.

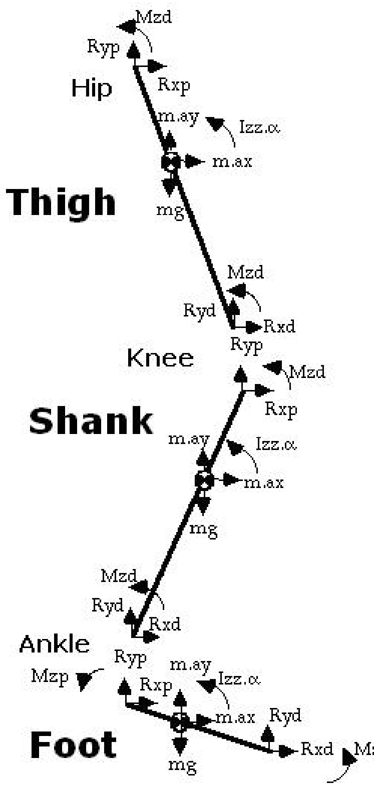

Figure 2.7 Free-body diagrams of lower extremities

Anthropometric data such as mass, mass moment of inertia of each segment, and the positions of their centers of mass are required to be known. These are determined mostly from published cadaver studies, though there now exist automatic techniques for their estimation in vivo.

2.4.3. Power

The indirect analysis of power which is the product of the moment and the angular velocity of the object, is a new concept in clinical biomechanics:

$$P = M\omega \tag{2.13}$$

When examining the muscle activity, it is evident that the muscle is doing concentric contraction, or positive work if the moment and angular velocity are in the same direction, and the muscle is doing eccentric contraction (lengthening), or doing negative work is the moment and angular velocity are in opposite directions.

Therefore, motion analysis yields the joint forces and moments via measuring external forces, segmental displacements, and knowledge of masses and inertias. However, these are net forces, moments and powers. Not for an individual muscle, ligament etc. The LSM models the net forces of different muscles as a single equivalent muscle. In the reality, motion of each joint is controlled by both agonistic and antagonistic muscles all the time.

2.5 Instrumentation and Filtering

2.5.1 Instrumentation

The image is retrieved via the video system, 3D position of each marker is measured and expressed as coordinates in terms of a fixed laboratory reference system. Orientations may be different but the method of analysis is the same for all systems (Figure 2.8).

During the motion, the force the foot applies to the ground is measured by a force plate (Figure 2.9) that is mounted securely in the floor such that its surface is leveled with the floor. The force plate has an instrument center that is below the floor and the resultant force and moment about this instrument center is measured. These data are sampled at a specific rate: usually at 1000 Hz. The resultant force and moment are expressed in an equivalent force system composed of the resultant force acting at a specific point on the surface of the force plate and a torque about the vertical axis.

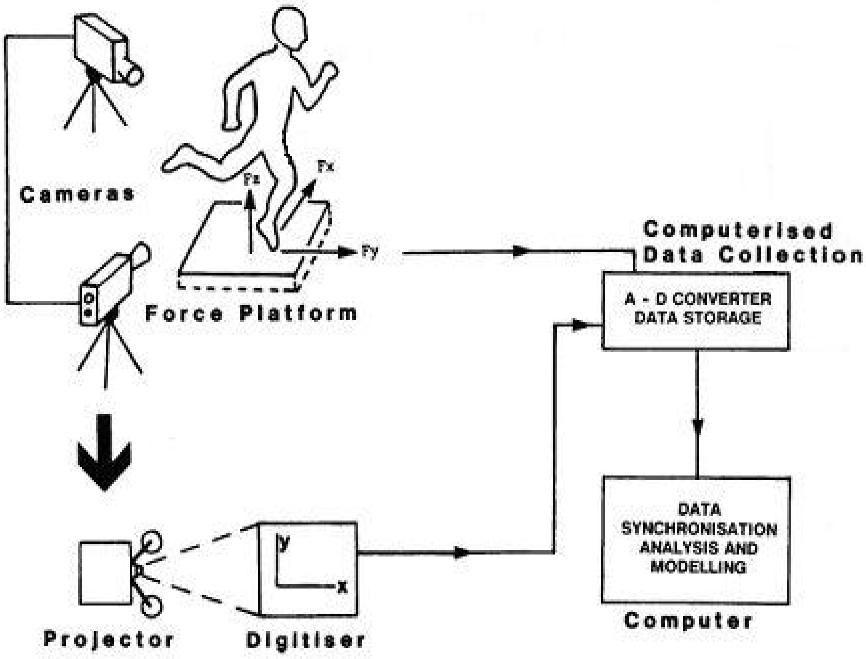

Figure 2.8 A typical data acquisition system in motion biomechanics

The signal obtained from the force platform is then amplified, passed through a signal conditioner, which converts it to a digital representation, and recorded. This signal is then amplified, passed to an analogue-to-digital converter, and the digital information which represents the characteristics of the ground reaction forces is recorded by the computer.

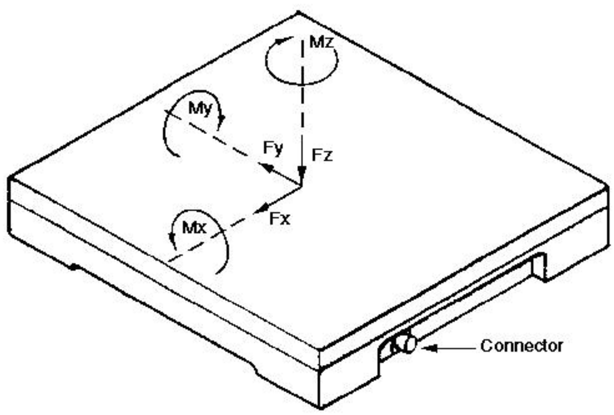

Figure 2.9 Force plate measures 3D forces and moments
(adapted from http://physiotherapy.curtin.edu.au/home/facilities/equipment/plates.cfm)

The process of recording body movement and production of segmental angular positions, velocities and accelerations represents a different system, while the attachment of electrodes to selected muscles, and the subsequent amplification, A-D conversion and recording of the EMG signals comprises a third which is not used in the present research.

2.5.2 Data Filtering

The sampled data is always contaminated with noise. The noise content must be minimized prior to differentiation because differentiation magnifies the signal and hence the noise. Following data collection, additional measures should be taken to minimize random error. Random noise is usually "white noise," characterized by high frequency content. The motion signal is usually limited to a band of low frequencies. Therefore, a *low-pass filter* should be used to remove the high frequency components and retain those of the low frequency. Systematic noise (such as the change in position of skin-fixed markers due to tissue deformation) is a greater problem. It may include both high and low frequency components. However, only the high frequency components will be filtered out by a low-pass filter. No filters can distinguish between a low-frequency systematic error signal and the movement signal if nothing is known about the nature of the error signal.

There are many types of filters applied to have an optimal cut-off frequency: classic Butterworth, Fourier series, Kalman, cubic, and finite impulse response (FIR) filters. Filter equations are frequently recursive. Current values depend on the previous values, which introduces a phase lag into the signal. Therefore, these filters are applied in both forward and reverse directions to remove the phase lag. A filter applied once in the forward and once in the reverse direction is termed a second order Butterworth filter.

It is seen clearly in Figure 2.10 that the data differentiated are too noisy to be reliable. To completely describe the motion, the data needs to be differentiated for a second time to obtain acceleration. Therefore, the position data are filtered using digital filters to smooth the data and to obtain reliable differentiation.

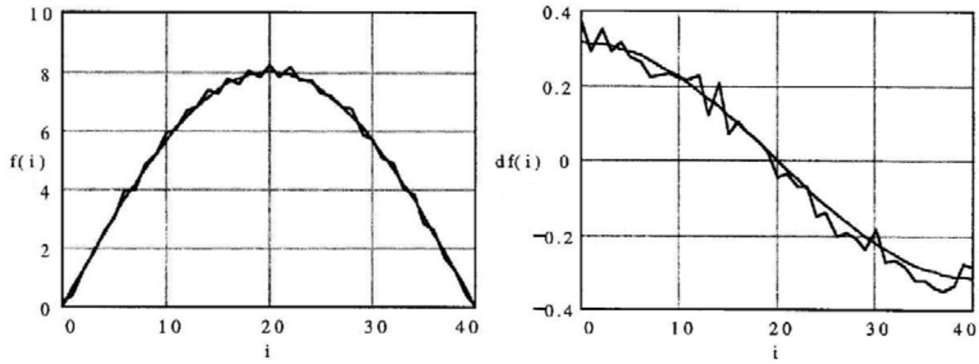

Figure 2.10 Noisy position and velocity data

3. MATERIALS & METHODS

3.1 Subjects

Fifteen normally developed third and forth grade primary school children (eight males and seven females) participated in the present study (see Table 3.1 for their averaged characteristics). All participants were able to hear and understand clearly the instructions given. After a full explanation of the purpose and methodology of the experiments, informed consent was obtained from the parents of the subjects.

Table 3.1
General and anthropometric characteristics of the subjects (N=15)

	Mean ± S.D.	Range
Age (Years)	9.61 ± 1.24	6.58-11.60
Mass (kg)	34.2 ± 5.1	28-42
Height (m)	1.38 ± 0.09	1.24-1.50
Pelvis width (cm)	18.80 ± 1.93	15-22
Hip width (cm)	9.20 ± 1.25	6.5-11
Knee width (cm)	8.77 ± 1.10	7-11
Ankle width (cm)	5.97 ± 0.44	5.5-6.5
Total leg length (cm)	72.70 ± 5.75	6.25-8.20

3.2 Procedures

A motion analysis system (Elite Eliclinic, BTS, Milan, Italy) with six infrared cameras was used to for the sit-to-stand motion measurements. Reflective markers were placed onto the segments and joints according to Davis marker placement protocol (see

Figure 3.1 for the lower extremities). The detailed marker positions are given in Appendix A.

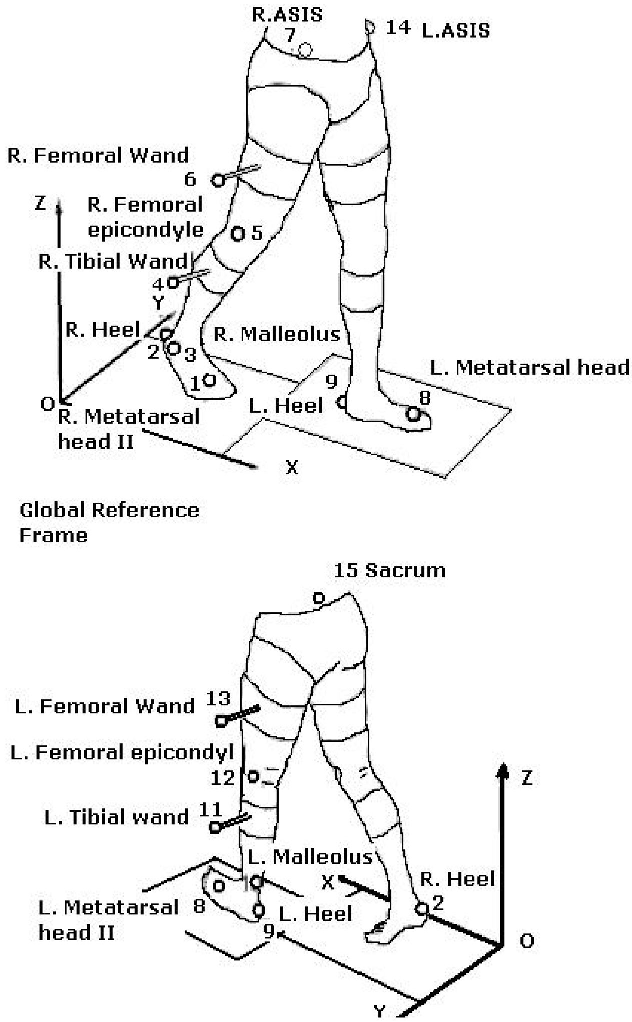

Figure 3.1 Davis Marker Placement Protocol for Lower Extremities
(adapted from http://www.lifemodeler.com/LM_Manual/A_motion.htm)

Two force plates (Kistler Instrumente AG, Winterthur, Switzerland) were used to measure ground reaction forces (GRF) in 3 directions. To analyze the motion in the sagittal plane, the GRF in the vertical and fore-aft shear directions were used. GRF components were normalized with respect to the body weight.

The subjects were asked to wear tight shorts and they were bare feet (Figure 3.2). They were seated on a backless seat which was selected to have a fixed height of 43 cm. Such seat height represents the standard school benches in primary schools. Each of the subjects' feet was positioned on a separate force plate with no initial angle limitations. This allowed the subjects to adjust their STS motion strategies as a function of the back load. The distance between the feet was set to equal the subjects' shoulder width. The subjects were asked to maintain the upright position and asked to look straight forward to a point 4 meters ahead and 2 meters high. Their hands were crossed on the chest holding shoulders in order to exclude the inertial effects of arm position.

After explaining the procedure and subsequent to a test motion, STS motion was performed at the subject's own preference of speed. 2 seconds after capture initiation, the subject was instructed to stand up. Care was taken to maintain the same voice level for the instructions in each trial. Upon completion of standing up, the subject was kept waiting for 2 more seconds before the instruction to finalize the task was given. The subjects performed 8 sit-to-stand trials and the smoothest 3 of them were selected for data analysis.

The measurement procedure was repeated in three different conditions:

(1) With no back load (referred to as *unloaded case*)

(2) With a backpack containing 10% of the subjects' body weight (referred to as *10% load case*)

(3) With a backpack containing 20% of the subjects' body weight (referred to as *20% load case*)

Note that for the 10% and 20% loaded cases, the subjects wore both strands of the backpacks.

3.3 Definition of STS Phases Used

The STS motion may offer alternative patterns without skipping critical points. Five critical points (T0 to T4) were defined to characterize the phases of STS motion:

T0, initial point of trunk flexion

T1, the point where buttocks are lifted

T2, the point of maximal hip flexion

T3, the point of maximal ankle dorsiflexion

T4, the point at which hip rise was finalized.

Figure 3.2 A subject with 20% body weight load.

Bounded by these critical points, four phases (Phase I to Phase IV) of STS motion were defined:

Phase I, forward lean of the trunk

Phase II, hip lifting off to maximal hip flexion

Phase III, maximal hip flexion to maximal ankle dorsiflexion

Phase IV, maximal ankle dorsiflexion to end of hip rise (standing up).

In addition, the time interval from lift-off to end of hip rise (i.e., T1 to T4) is referred to as the *on-the-ground-STS motion*. This duration is emphasized in data analysis since most of the energy demanding activities take place after the subject lifts off the seat.

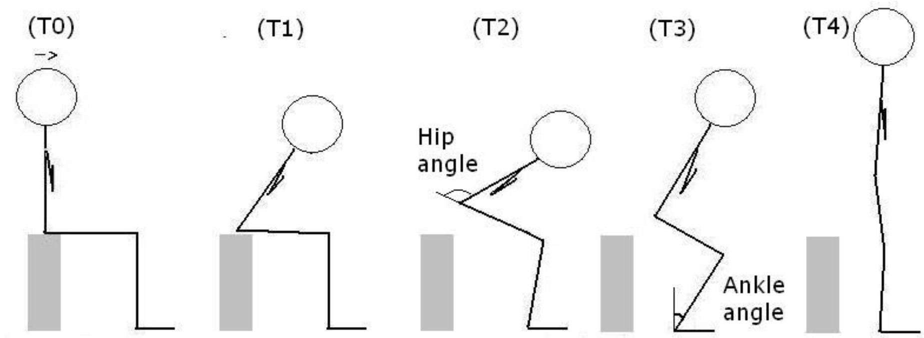

Figure 3.3 Critical points of STS motion.

3.4 Data Analysis and Statistics

The total duration of time from T0 to T4 was converted into a 100% scale: referred to as *% STS*. Kinematic and kinetic data were averaged for each percent of the STS motion. Averaged data were plotted and presented in a family of curves allowing a comparison for the dynamics of the STS motion for unloaded, 10% loaded, and 20% loaded cases.

It should be noted that the plotted curves are informative about the general mechanism of the STS motion and more importantly on the effects varying back loading. However, because the extremum points of different measurements do not necessarily occur at the same percent of the STS motion, such presentation of data causes an underestimation for the analysis of the kinematic and kinetic determinants of the motion. Therefore, an extremum analysis was performed and the results obtained (including a statistical analysis) were shown in tables.

For testing the statistical significance of differences in kinematic and kinetic data, one-way analysis of variance with repeated measures was used. Differences were considered significant at $p < 0.05$. Otherwise, the p value is indicated in text.

4. RESULTS

4.1 Phase Durations of STS Motion with Backpack Load

The durations of four phases of unloaded case was compared to those of the 10% and 20% loaded cases (Table 4.1). Phase I (forward lean of the trunk) is a remarkably long part of STS motion (e.g. for the unloaded case this phase takes 0.46 sec, 36% of STS). However, it is not a highly energy demanding phase: it is rather a preparation phase for the succeeding phases. Although for the 10% loaded case, Phase I lasts significantly shorter than the unloaded case the duration differences between unloaded and loaded cases are considered minor.

For the unloaded case Phase II (hip lifting off to maximum hip flexion) lasts about 0.17 sec (13% of STS motion). Statistical analysis shows that the duration of this phase shortens only to a limited extent as the back load increases ($p_{1\text{-}3} = 0.07$, $p_{2\text{-}3} = 0.10$).

Table 4.1
Durations of Phases in STS Motion for Different Back Loads

	Unloaded	10% Loaded	20% Loaded
Phase I (sec)	0.46 ± 0.12	0.42 ± 0.07**	0.44 ± 0.10
Phase II (sec)	0.17 ± 0.08	0.16 ± 0.07	0.14 ± 0.07*†
Phase III (sec)	0.14 ± 0.13	0.16 ± 0.08	0.20 ± 0.08**††
Phase IV (sec)	0.51 ± 0.18	0.53 ± 0.11**	0.57 ± 0.18**
Total STS (sec)	1.28 ± 0.26	1.29 ± 0.21	1.35 ± 0.22†
On-the-ground STS (sec)	0.83 ± 0.21	0.86 ± 0.16	0.93 ± 0.21**†

Values are in seconds (mean ± SD)
* $0.05 < P \leq 0.10$ compared to the unloaded case. ** $P \leq 0.05$ compared to the unloaded case.
† $0.05 < P \leq 0.10$ compared to the 10% load case. †† $P \leq 0.05$ compared to the 10% load case.

Phase III (maximum hip flexion to maximum ankle dorsiflexion) takes approximately 0.14 sec (11% of STS motion) for the unloaded case. A remarkable result is that the duration of this phase was significantly longer for the 20% load case.

Phase IV (maximal ankle dorsiflexion to end of hip rise) is the longest phase of STS motion (e.g. for the unloaded case this phase takes 0.51 sec, 40% of STS). The duration of this phase is significantly longer for both of the loaded cases. However, the duration difference between the 10% load and 20% load cases is not significant.

On-the-ground part of the motion (beginning of Phase II to end of Phase IV) includes the most demanding activities: This part takes significantly longer for the 20% load case compared to the unloaded case ($p_{1-3} < 0.01$).

In conclusion, the total duration of STS motion remains almost the same independent of the back load. However, the different phases of the motion show sizable duration variations for different back loading.

4.2 Kinematics of STS Motion with Backpack Load

Figure 4.1 shows the averaged angular displacements of the ankle in the sagittal plane. The general pattern of ankle dorsiflexion angle variation during the motion is similar for all the cases. However, significantly more pronounced ankle dorsiflexion was measured as the back load of the subjects increased.

Table 4.2 shows the distinction more clearly as a result of an extremum analysis: (1) Initial angle of the ankle joint was significantly higher for 20% load case compared to that of unloaded and 10% load cases. (2) The maximum value of the ankle dorsiflexion angle is substantially different in each of the cases. Note however that there is no significant difference in the ankle dorsiflexion angles at the end of the extension phase.

Knee flexion pattern during STS motion follows almost an identical path independent of the back load: firstly it remains nearly constant followed by a substantial decrease in knee flexion angle (Figure 4.2). A small peaking behavior for the 20% load case in the first third of the motion is an exception for that.

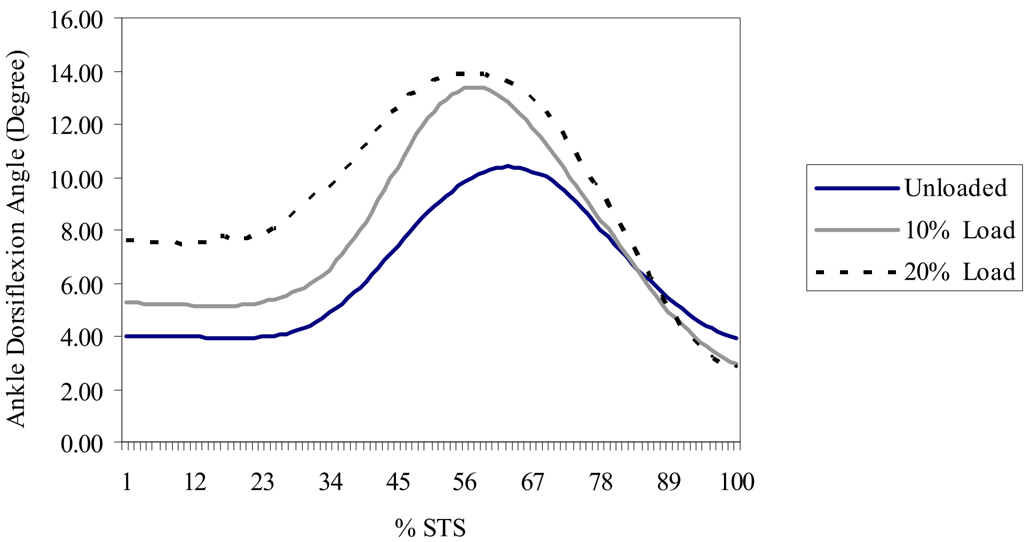

Figure 4.1 Ankle dorsiflexion angle in STS motion for different back loads

However, the final knee flexion angle does show sizable and significant differences (Table 4.3): (1) compared to the unloaded case, the final knee flexion of 10% load case is limited (2) compared to both the unloaded case and the 10% load case, 20%

Table 4.2
Extremum Values of Ankle Dorsiflexion Angle in STS Motion for Different Backpack Loads

	Unloaded	**10% Loaded**	**20% Loaded**
Initial Angle (^{O})	3.99 ± 7.10	5.26 ± 6.80	7.61 ± 7.06*†
Maximum Angle (^{O})	12.12 ± 5.73	14.85 ± 6.27*	17.08 ± 7.02*†
Final Angle (^{O})	3.91 ± 5.08	2.93 ± 4.76	2.81 ± 5.21

Values are in degrees (mean ± SD)
* $P \leq 0.05$ compared to the unloaded case.
† $P \leq 0.05$ compared to the 10% load case.

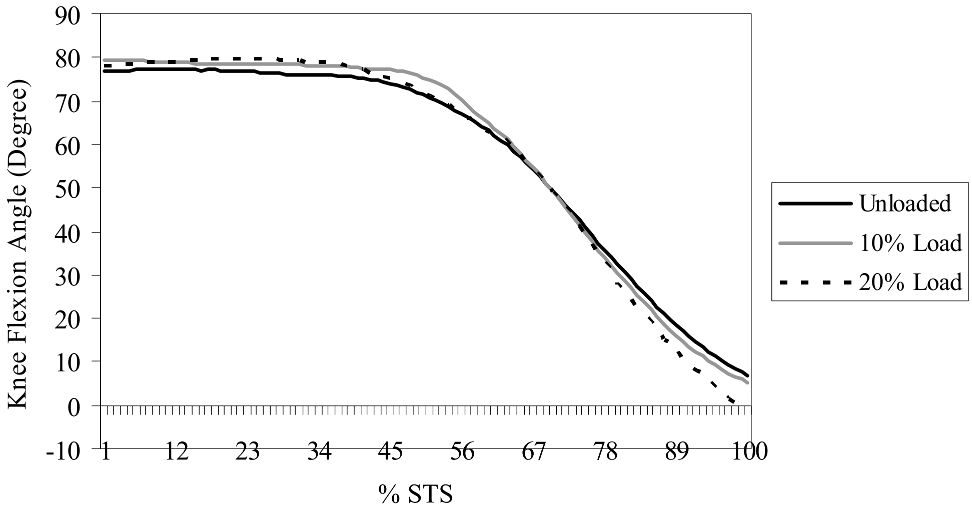

Figure 4.2 Knee flexion angle in STS motion for different back loads

load case shows a major reduction in knee flexion. The latter result shows that at the end of STS motion, the knee reaches full extension and even hyperextension at very high back loads.

Table 4.3
Extremum Values of Knee Flexion Angle in STS Motion for Different Back Loads

	Unloaded	**10% Loaded**	**20% Loaded**
Initial Angle (^{O})	76.93 ± 9.25	79.37 ± 8.99*	78.09 ± 9.15
Final Angle (^{O})	7.01 ± 9.52	5.29 ± 9.39*	-0.20 ± 9.56*†

Values are in degrees (mean ± SD)
* $P \leq 0.05$ compared to the unloaded case.
† $P \leq 0.05$ compared to the 10% load case.

Hip flexion angle first typically increases up to a maximum and then decreases until almost full extension (Figure 4.3). A remarkable result is that both 10% and 20% load cases show higher hip flexion angles compared to the unloaded case. This difference is much more pronounced at the first half of the STS motion. The peak hip flexion angle of the 20% load case occurs much earlier than the other cases. This is not only because of

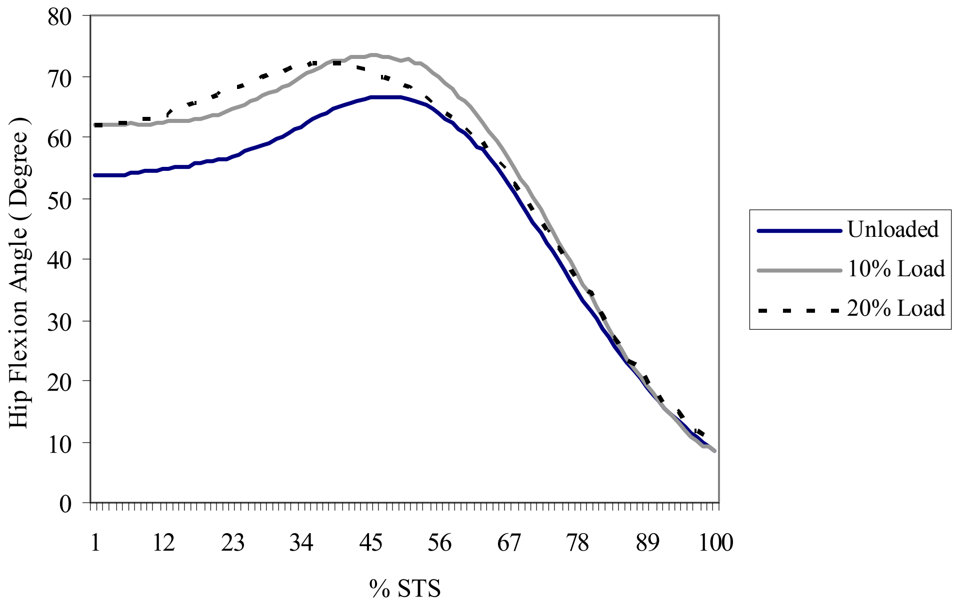

Figure 4.3 Hip flexion angle in STS motion for different back loads

the shortening of previous part of the STS motion but also due to lengthening of the latter part.

The extremum analysis allows reflecting more details (Table 4.4): (1) 10% and 20% load cases differ significantly from the unloaded case in terms of initial hip flexion angle and maximum hip flexion angle (for both angles p_{1-2}, $p_{1-3} \leq 0.01$). (2) However, no statistical difference is found among the three cases for the final hip angle.

Pelvic tilt in general initiates from a negative value, increase up to a positive maximum, then decreases to a certain positive value (Figure 4.4). Initial pelvic tilt angle and transitional tilt angle (at which maximum rate of pelvic tilt angle is observed) shows that there is a substantial difference between the unloaded and loaded cases (Table 4.5): (1) Initial angle of pelvic tilt decreases significantly (2) Both of the loaded cases attain positive transition angle values in contrast to the unloaded case. The angle differences between the unloaded case and the loaded cases are significant. However, no significant difference was found among the loaded cases.

Table 4.4
Extremum Values of Hip Flexion Angle in STS Motion for Different Back Loads

	Unloaded	10% Loaded	20% Loaded
Initial Angle (^{O})	53.77 ± 10.09	61.87 ± 8.19*	61.92 ± 7.92*
Maximum Angle (^{O})	70.64 ± 12.89	76.16 ± 11.88*	77.37 ± 9.87*
Final Angle (^{O})	8.60 ± 10.29	8.50 ± 10.70	10.76 ± 12.03

Values are in degrees (mean ± SD)
* $P \leq 0.05$ compared to the unloaded case.
† $P \leq 0.05$ compared to the 10% load case.

It is concluded that a back load up to 20% of the body weight does not change the shape of curves representing kinematic data. However, back load does cause sizable shifts to occur in these curves both in terms of angles and % of STS motion: (1) typically, initial joint angles tend to show a higher degree of flexion. For example, the trunk attains a more

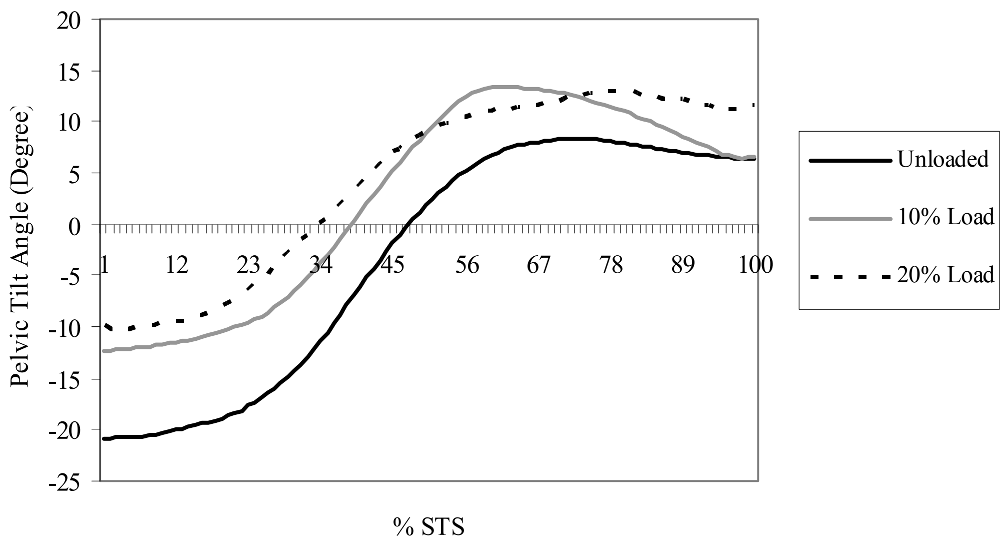

Figure 4.4 Pelvic tilt angle in STS motion for different back loads

forward initial position. (2) The flexion angles are higher before rising with the load (i.e., at beginning of Phase III), and (3) in the 20% load case, at the end of the STS motion the knee goes into hyperextension where cruciate ligaments (CL) produce forces in the favor of full extension.

Table 4.5
Discrete Values of Pelvic Tilt Angle in STS Motion for Different Back Loads

	Unloaded	**10% Loaded**	**20% Loaded**
Initial Angle (^{O})	-20.89 ± 7.54	-12.29 ± 7.72*	-9.76 ± 6.64*
Transition Angle (^{O})	-6.31 ± 6.84	2.33 ± 8.09*	1.16 ± 4.79*
Final Angle (^{O})	6.29 ± 5.80	6.49 ± 7.32	11.59 ± 9.13

Values are in degrees (mean ± SD).
* P ≤ 0.05 compared to the unloaded case.
† P ≤ 0.05 compared to the 10% load case.

4.3 Kinetics of STS Motion with Backpack Load

4.3.1 Ground Reaction Forces (GRF)

Figure 4.5 shows the vertical GRF of the subjects for the unloaded case as well as the 10% and 20% load cases. In general, in the first quadrant of the motion almost no GRF were recorded independent of the back load. During the on-the-ground part however, much higher GRF were measured as the back load of the subjects was increased.

Table 4.6 shows that the maximal vertical GRF of the 20% load case is significantly higher than that of the 10% load case and the unloaded case. Note also that for the 10% load case the maximal vertical GRF is significantly higher compared to that of unloaded case.

Note that in contrast to the plotted averaged GRF data (Figure 4.6), the extremum analysis shows that minimum value of the fore-aft shear force remains almost the same for all cases (Table 4.6).

It is concluded that the vertical GRF increases in proportion with increasing back load whereas fore-aft shear showed no significant changes.

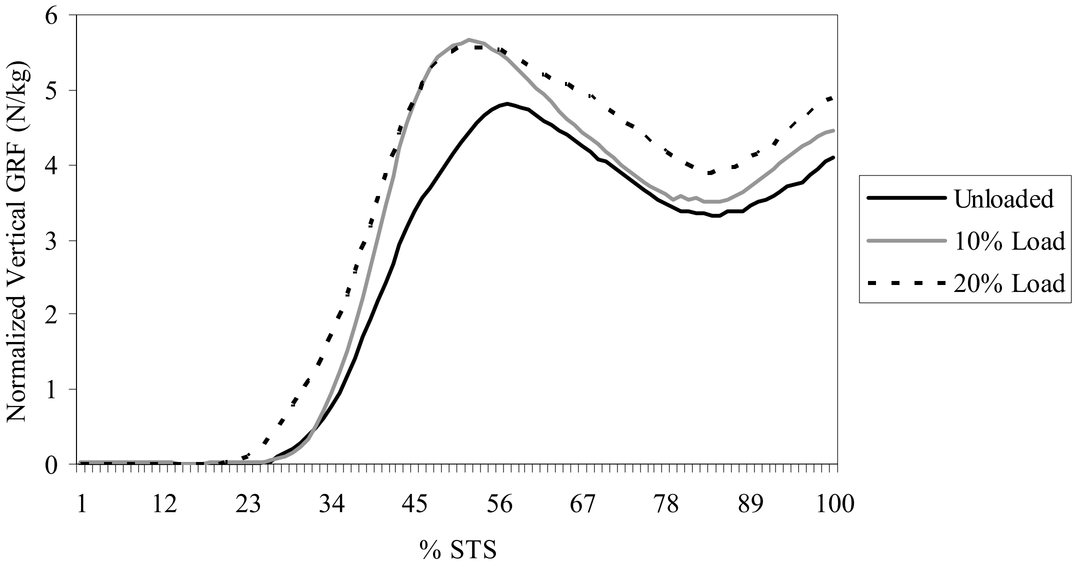

Figure 4.5 Normalized vertical ground reaction force for different back loads

4.3.2 Joint Moments and Powers

Ankle dorsiflexion moment shows a continuous increase as the vertical GRF attains values other than zero (Figure 4.7). Therefore, the ankle dorsiflexion moment has its maximal value at the standing position. Maximal ankle dorsiflexion moment of the 20% load case is significantly higher than that of the unloaded case (Table 4.7).

Figure 4.8 shows that the ankle power is approximately zero until the vertical GRF is nonzero. It reaches a peak in the last third of the motion and drops thereafter. Ankle

Table 4.6
Extremum Values of Normalized Ground Reaction Forces For One Leg for Different Back Loads

	Unloaded	**10% Loaded**	**20% Loaded**
Maximum Vertical GRF (N/kg)	5.73 ± 0.84	6.22 ± 0.88*	6.64 ± 1.01*†
Minimum Fore-Aft Shear Force (N/kg)	-0.65 ± 0.26	-0.63 ± 0.25	-0.62 ± 0.21

Values are in Newtons/kilogram (mean ± SD).
* $P \leq 0.05$ compared to the unloaded case.
† $P \leq 0.05$ compared to the 10% load case.

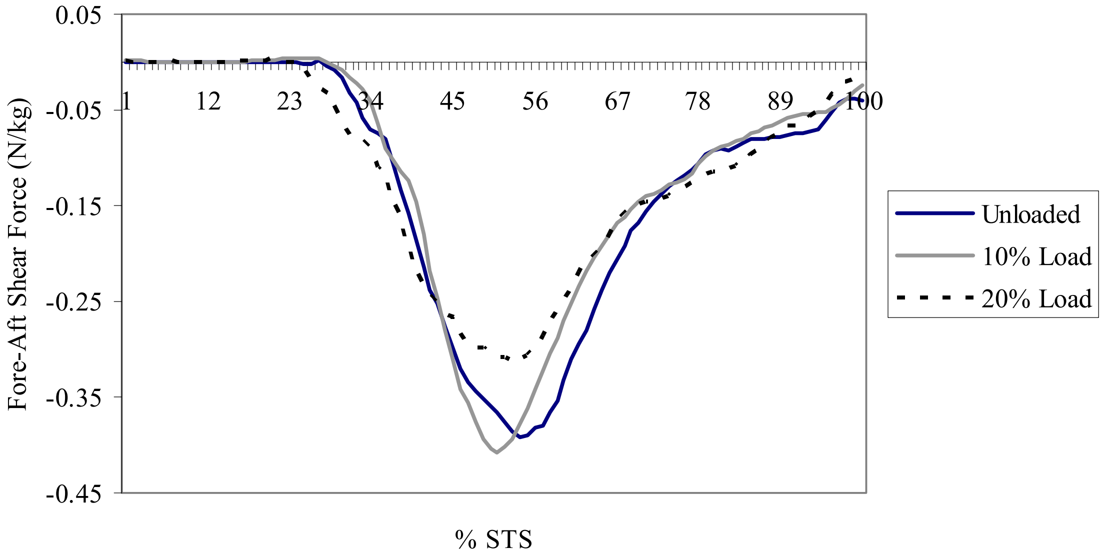

Figure 4.6 Fore-aft shear force for different back loads

power increases substantially as the back load increases. In terms of maximal ankle power, both 10% and 20% load cases show a significant difference compared to the unloaded case (Table 4.7).

In general, after Phase I, knee extension moment increases until a maximal value is reached (Figure 4.9). Subsequently, a reduction of the extension moment occurs (till

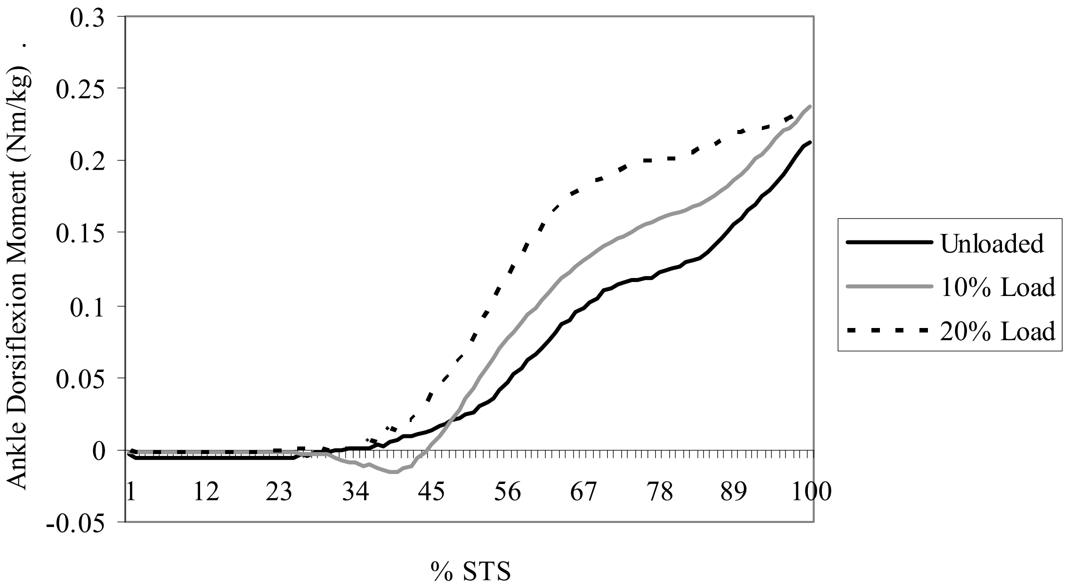

Figure 4.7 Normalized ankle dorsi-flexion moment for different back loads

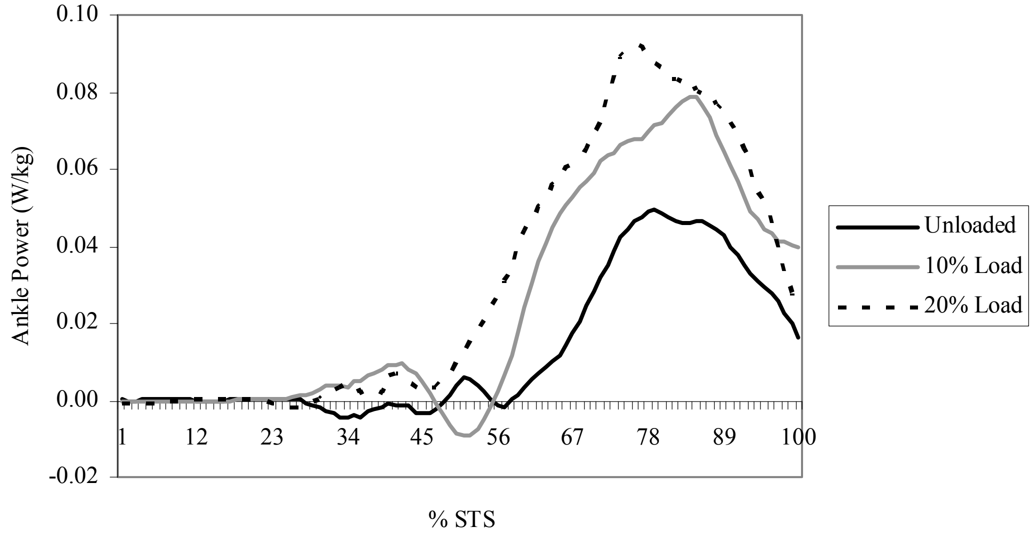

Figure 4.8 Normalized ankle power for different back loads

approximately 80% STS) which later turns into an increasing flexor moment. In cases with back load, greater knee extension moment is observed. The same is true for knee flexion moment.

Regarding the maximal knee moment the differences between the loaded and unloaded cases are significant. However, no significant difference is found among the 10% and 20% load cases (Table 4.7). Nevertheless, it is remarkable that even 10% load case causes an increase of approximately 25% in the maximal knee moment.

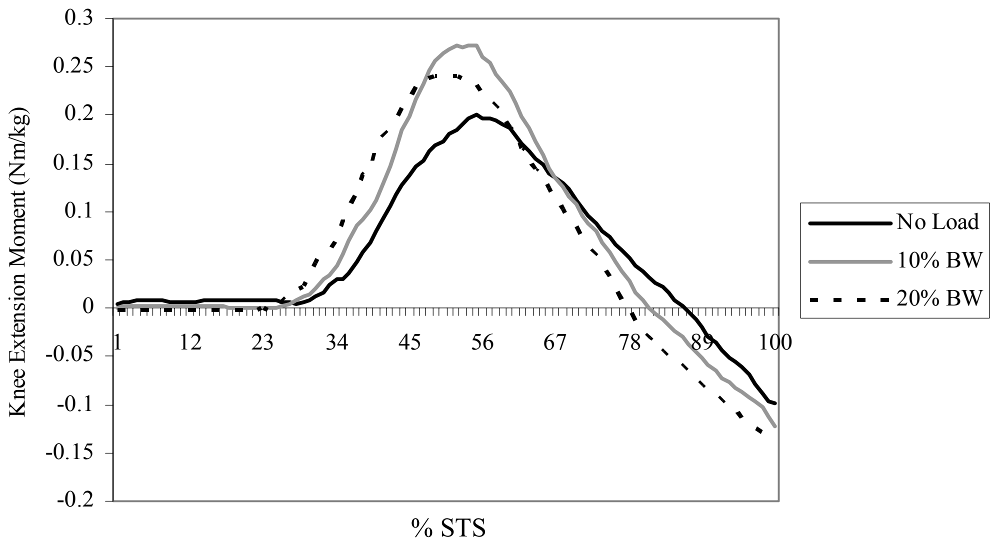

Figure 4.9 Normalized knee extension moment for different back loads

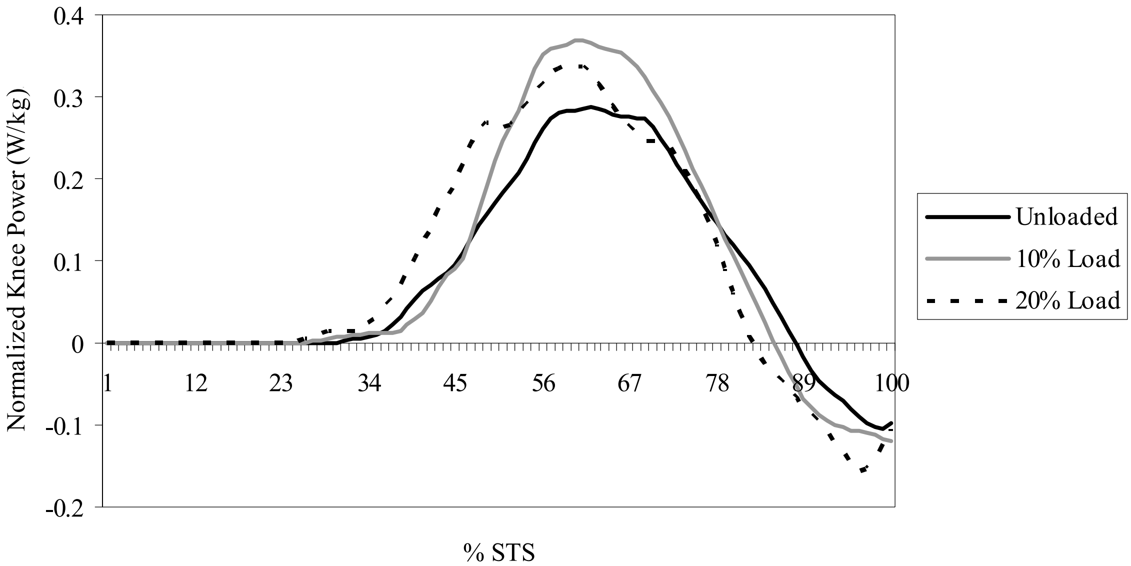

Figure 4.10 Normalized knee power for different back loads

Note that in general knee extension moment and power plots show great similarities (Figures 4.9 and 4.10). Both quantities (1) are constant during Phase I, (2) increase to a maximal value after which (3) they decrease to zero and at the later parts of STS motion (4) a flexor moment and power builds up. However, both 10% and 20% load cases show markedly higher moment and power values than those of the unloaded case.

Extremum analysis shows that in 10% and 20% load cases maximal knee power is significantly higher than that of the unloaded case (Table 4.7). However, no significant difference is found among the 10% and 20% load cases.

In general, hip starts STS motion with a flexion moment, decreases to zero moment at the first third of the motion. Then hip moment rises on the extension side up to a peak value, lowers slightly, and ends with an extension moment (Figure 4.11). All three cases have a similar hip moment profile especially during on-the-ground stage of the STS motion. Table 4.7 shows that the change in the maximal value of hip extension moment with respect to back load is not significant.

Hip power rises moderately in Phase I (forward lean of the trunk). In phase II (lift off – maximal hip flexion), it decreases slightly and achieve a minimum before a rise to

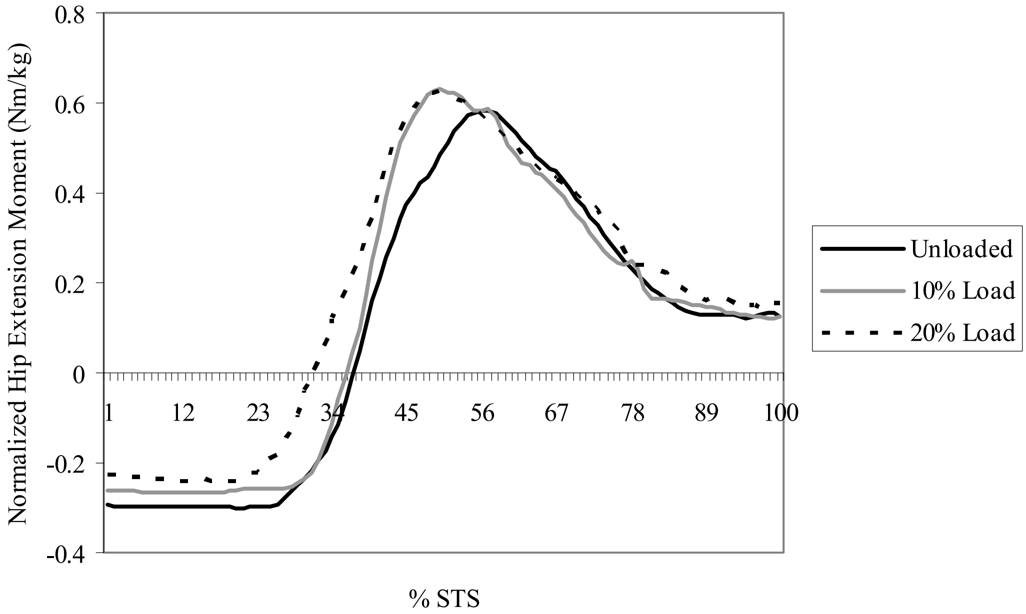

Figure 4.11 Normalized hip extension moment for different back loads

the maximal hip power occurs. In the cases with load, higher powers are observed in the second half of the motion (Figure 4.12). However, Table 4.7 shows that maximal hip power does not differ significantly regardless of the load.

In the most general sense, different backpack loading levels have impact on different joints. With respect to unloaded case, 10% load significantly increased the

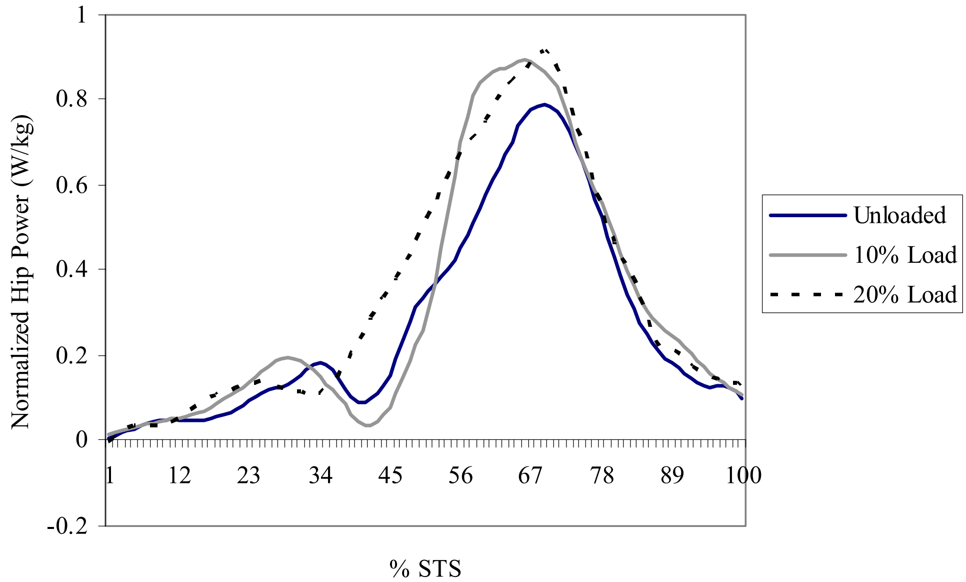

Figure 4.12 Normalized hip power for different back loads

observed maximum knee moment (21% increase) and maximum knee power (26% increase). Ankle power is increased by 42%. However, no change is observed in ankle moment (Table 4.7).

On the other hand, when the load was increased from 10% to 20% of body weight, the knee kinetic variables remained constant. Nevertheless, maximum ankle moment increased by 19% with respect to 10% load case (Table 4.7).

Table 4.7
Extremum Values of Ankle, Knee and Hip Moments as well as Powers for Different Back Loads

	Unloaded	**10% Load**	**20% Load**
Max. Ankle Moment (Nm/kg)	0.25 ± 0.15	0.27 ± 0.17	0.32 ± 0.22*
Max. Ankle Power (W/kg)	0.12 ± 0.10	0.17 ± 0.14*	0.20 ± 0.17*
Max. Knee Moment (Nm/kg)	0.28 ± 0.14	0.34 ± 0.21*	0.37 ± 0.19*
Max. Knee Power (W/kg)	0.54 ± 0.37	0.68 ± 0.46*	0.77 ± 0.57*
Max. Hip Moment (Nm/kg)	0.80 ± 0.24	0.79 ± 0.38	0.85 ± 0.37
Max. Hip Power (W/kg)	1.37 ± 0.85	1.44 ± 1.04	1.36 ± 0.96

Moment and power values are in Nm/kg or W/kg respectively (mean ± SD)
* $P \leq 0.05$ compared to the unloaded case.
† $P \leq 0.05$ compared to the 10% load case.

Despite important changes on the kinetic variables of ankle and knee; hip moment and power remained constant. It should be noted that hip moment and power values exhibited a high variance.

Therefore, different back load levels affect different joints of the lower extremity: Up to 10% backpack load the most prone joint is the knee. However, when the load is further increased to 20% of body weight, ankle is the most affected part of the lower extremity. None of the kinetic variables of the hip joint were affected by the backpack load.

5. DISCUSSION

5.1 Phase Definitions for the STS Motion

The STS motion is commonly defined as moving the body's center of mass upward from a sitting position to a standing position without losing balance [27]. Alternatively, it is defined as a transitional movement to the upright posture requiring movement of the center of mass (COM) from a stable position to a less stable position over extended lower extremities [11].

In addition to such nonexistence of a univocal definition for the entire STS motion, the definitions proposed for the different phases of the motion are not also unique. Such phases are described by a set of critical points of kinematic or kinetic variables [28-30]. A definition of these phases that is used frequently is the one provided by Schenkman et al [29]. It is marked by 4 events (Table 5.1). However, in recent reports, a new phase was introduced by dividing phase II into two events [11]. This new phase starts from initial knee extension and ends with maximum ankle dorsiflexion. Therefore, this definition includes 5 phases (see Table 5.1 for a comparison).

In the present study a 5-phase definition similar to Vander Linden et al., was employed however certain modifications were introduced:

(1) Presently, initial knee extension was replaced by maximal hip flexion. During the STS motion the knee is continuously extended from an initial flexed position: in the first half of the motion, the knee remains flexed with minor alterations whereas in the second half, it experiences a highly pronounced extension. Therefore, to make the distinction a cut-off point needs to be detected. However, no established definition of such cut-off point was proposed in the literature. Because of the low slope, the assignment of the cut-off point for the rate of knee extension makes it subjective and error sensitive (i.e.; leading to great variability with regard to artifacts) and criteria dependent. In contrast, maximal hip flexion is an extremum point that occurs in each trial and can be

Table 5.1
Previously and presently used phase definitions

	Stam et al. (2002)	Lee et al. (2004)	Present Research
Phases	**Phase I** (flexion-momentum phase): starts with forward lean of the trunk and ends just before the buttocks are lifted off the chair.	**Phase I** is the same as Phase I of *Stam et al.*	**Phase I** is the same as Phase I of *Stam et al.*
	Phase II (momentum-transfer phase) begins as the buttocks are lifted and ends with maximal ankle dorsiflexion.	**Phase II**: begins as the buttocks are lifted and ends with maximal hip flexion.	**Phase II**: begins as the buttocks are lifted and ends with maximal hip flexion.
		Phase III: starts from initial point of steep knee extension and ends with maximum ankle dorsiflexion	**Phase III:** starts from maximal hip flexion and ends with maximum ankle dorsiflexion
	Phase III (extension phase): initiated just after maximum ankle dorsiflexion and ends when the hips first cease to extend; including leg and trunk extension.	**Phase IV** is the same as Phase III of *Stam et al.*	**Phase IV** is the same as Phase III of *Stam et al.*
	Phase IV (stabilization phase): begins after hip extension is reached and ends when all motion associated with stabilization is completed.	**Phase V** is the same as Phase IV of *Stam et al.*	**Phase V** is the same as Phase IV of *Stam et al.*

detected more precisely.

(2) Extension phase (phase IV) involving vertical transfer of body mass is characterized by knee and hip extension. Its endpoint was previously defined as the full extension of the hip and knee [15, 29]. On the other hand, an additional stabilization phase was characterized by a swinging motion which introduces short range flexion-extension motions. This leads to longer extension phase although the end of rise is readily achieved. Therefore, presently the end of phase IV is defined as the end of rise of hip joint.

(3) In earlier studies, the end of STS motion was chosen to be the cease of motion of markers [15, 29]. However, this period includes a stabilization phase, end point of which is not clearly defined in the literature. Since stable standing involves lateral and anterior-posterior sway, it is difficult to define the end point. Some of the studies using Schenkman's protocol typically report the first 3 phases [31, 32]. In the present study, especially for the 20% load case, subjects were observed to face a considerable difficulty in achieving stabilization. Although this may be considered as an interesting effect of back loading, in our view, analysis of the movement pattern in the stabilization phase turns into a standing posture problem, outside the general pattern of STS motion analysis. Therefore, such stabilization phase is beyond the goal of the present study and for that reason it was not included in our analysis. However, new studies on the effects of back loading analyzed after a sudden deceleration at the end of STS motion are indicated.

5.2 Biomechanical Implications of the Present Results

The present results reported for children with no back load showed that on the average a healthy child of age 10 completes the STS motion (up to the stabilization phase) in 1.28 seconds. This is in agreement with the findings of van der Linden et al. [11]: STS motion of healthy children (9 years old on the average) was studied in order to investigate the effect of bench height (and to compare the results with those of children with cerebral palsy). The results of these authors showed that the total duration of STS motion approximates 1.24 seconds for healthy children. Moreover, the kinematic data obtained presently and those reported by Van der Linden et al show the same patterns. Above all such comparable results show that our present results obtained for the unloaded children are in agreement with earlier findings. This is important since these results are used presently as control measurements to determine the effects of back load on STS motion.

A major result shown presently was that the total duration of STS motion does not change significantly for the range of back load studied. However, the results of this study show that the durations of the individual phases do change as a function of back load suggesting strongly that the neuromuscular system employs different strategies to accommodate the altered loading conditions in order to complete the STS motion within the same period of time. It should be noted that no previous study on STS motion of children with back load is available, results of which could be compared to our present results. However, interestingly, Van der Linden et al. [11] showed for healthy children with no back load that the total duration of STS motion did not differ after the experiments were performed using a high or a low bench. Nevertheless, the duration of the extension phase was affected by the bench height: It increased significantly as the STS motion is performed on the low bench. Similar to our findings, this shows adjusted individual phases to complete the STS motion within the same period of time. We propose the following explanation for the similar results. If the subjects are healthy, they are capable of not reflecting the effects of altered mechanical conditions to the total duration of STS motion. Therefore, this determinant of STS motion alone can be used as a test to identify pathology in the neuromuscular system. New studies are indicated to test the limits of such mechanical perturbation which would still not impose an increased total duration of STS motion.

On the other hand the effects of back load can be better understood if the temporal parameters of STS motion are analyzed i.e., if the durations of individual phase of the motion are considered together with kinematic and kinetic data. Such effects are discussed below for each phase separately.

Phase I (forward lean of the trunk) Increasing the load did not affect the phase duration. This is due to the fact that both the initial position and the maximal forward flexion of the trunk are slightly forward maintaining the angular displacement. In other words, in the loaded cases, hip is initially more flexed to balance the new center of mass (COM) of the body on the seat. However, the end point of the hip flexion within phase I also shifts. As the velocity is conserved and mass increases, momentum is increased.

Phase II (hip lifting off to maximal hip flexion) Only a limited shortening occurs in the phase duration as a result of back load. This is because for example in the 20% load case the hip angle (flexion) in the beginning of this phase is already close to its maximum. Regarding the other joints, the ankle is rapidly performing dorsiflexion whereas knee undergoes only a limited extension. It should be noted that the activity of ankle dorsiflexors, especially tibialis anterior (TA) muscle, dominates this phase: as the back load increases TA muscle performs contracts to cause dorsiflexion for a wider range of ankle joint.

Phase III (maximal hip flexion to maximal ankle dorsiflexion) The phase duration increases substantially (by 36%) for the 20% load case. Within this phase the initial and maximal ankle dorsiflexion angles increase equally. This helps the body to move the COM (1) forward to counter balance the effect of the back load shifting the COM backwards and (2) downwards to lower the COM in order to reduce the risk of falling. Therefore, both strategies enhance the mechanical stabilization of the body. On the other hand the strategy of increasing the initial and maximal ankle dorsiflexion equally allows keeping the range of angular motion the same for both loaded and unloaded cases. However, the range of motion for the hip joint is increased substantially for example for the loaded condition. Therefore, we conclude that the main reason for the phase duration to increase is the altered motion of the hip joint as a function of back load. It should be noted that our kinematic and kinetic data suggests that the hip extensor muscles are predominantly active in this phase. Because the events of this phase prepare the body for the vertical displacement, it can be considered as the *load acceptance phase* in analogy to the gait cycle.

Phase IV (maximal ankle dorsiflexion to end of hip rise) The phase duration increases significantly with back loading. A major reason for this is that the range of motion the ankle joint covers increases substantially with increased back load. The substantially higher maximal angle of dorsiflexion measured at the end of the preceding phase for the loaded cases causes the triceps surae muscles to attain higher lengths (distal lengthening). In accordance with the nature of the STS motion, the ankle joint attends its natural position at the end of this phase independent of the back load. Therefore, an increase in the back load results in a wider length range to be covered by the triceps surae muscles as a higher muscle length was attained in the beginning of the last phase of STS motion. The general notion is that skeletal muscles generate their optimal forces at neutral positions of the joints [33]. This would mean that STS motion with back load causes these muscles to function in an unfavorable length range for force exertion. However, Maganaris et al. showed using dynamometric measurements in human and in vivo that both medial and lateral gastrocnemius muscles exert much higher forces in the dorsiflexion to neutral position joint range compared to neutral position to plantar flexion joint range [34]. These findings suggest that the increased angle of maximal dorsiflexion is a part of the strategy adopted by the loaded subjects to overcome the higher demand for ankle moment.

The kinematic analysis shows that the joint angles give different responses to back loading. At the onset of the motion, hip flexion angle and pelvic tilt angle is greater in loaded subjects. This is explained as an effort to maintain the position of the COM of the upper body on the hip-seat interface by means of forward flexion of the trunk. In the mid-STS, both the maximal ankle dorsiflexion and maximal hip flexion angles were shown to be greater for the loaded subjects to shift the new COM both to a lower and forward position. Relative changes in these two directions reduce the moment arm of the COM providing a decreased need for power generation and conceivably ease the coordination of muscular activity. Therefore, mid-STS responses to loading can be considered as the maintenance of COM and balance strategy. It should be noted that at the end of phase IV of 20% load case, some trials even showed knee hyperextension (-0.20 ± 9.56 degrees) where cruciate ligaments produce forces in the favor of full extension.

Kinematic results of the present study show that a back load within the range studied does not change the shapes of the curves. An increase from 10% to 20% load does not change the extremal values of the hip joint. However, hip represents a difference between unloaded and loaded trials. On the other hand, ankle and knee responded to load increments. Kinematical compensation mechanism works almost the same for all cases: Reduction of moment arm of COM, increasing the stability and ability to control are the main goals of the strategies of the joints performing STS motion.

Peak vertical ground reaction forces (GRF) were found presently to increase with back loading. However, as the load is increased, the change of peak vertical GRF between each load increment reduces. This is because the dynamic component of the vertical GRF reduces as the STS motion is slowed down.

In terms of kinetics, the body exhibits different strategies as the backpack load increase. As the load increases, the hip moment and power remain statistically indifferent with a high variance which may be due to constant bench height. The same bench can be classified as low or high bench depending on the knee height of the subject. Previous research showed that changing the seat height affects the maximum moment needed at the hip and knee. The differences can be as large as 50% to 60% [35-38]. Knee exhibits a significant increase both in maximal values of moment and power when the load is increased to 10% of body weight. However, this is not the case when the load is further increased to 20%. Therefore, it is concluded that loads even low as 10% cause high knee moments (21% increase). On the other hand, the picture is different for the ankle joint. Moment produced is conserved when the load is increased to 10%, but it increases significantly as compared to unloaded case when the load is 20%. To say, 20% load produces high ankle moments when compared to that of unloaded case (28% increase). Therefore, the complex mechanism of STS motion seems to work in a way to employ different joints on different load increments. Although forward flexion of the trunk strategy was shown to reduce the knee moments [4], loading effects increase the moments faced.

It is concluded that children performing STS motion with their daily backpacks (i.e., a load up to approximately 20% of their body weight) achieve net knee and ankle moments of 30% and powers 50-70% higher than the unloaded case. However, no significant differences in joint kinetics were found between 10% and 20% load cases except for the ankle moment. The increased duration of phases III and IV is expected to prevent the impact effects on the GRF and therefore on the peak moments and powers. Slowness in the late STS of 20% load case both supplies the extra energy needed from the SOL muscle which is a slow contracting but powerful muscle and avoids the higher moments and powers induced by the high impact forces on the ground. A remarkable conclusion is that even 10% load may be as problematic as the 20% load for the knee joint. This suggests that the back load limit should be below 10% of the body weight and especially for children with pathology in the muscles and tendons of the knee joints should avoid performing such a motion with high back loads. It is also concluded that, as the back load is increased, the joints experience wider range of angles. Therefore, the muscles controlling the STS motion function in wider length ranges of force exertion which suggests that a higher state of neuromuscular coordination is necessary in order to perform the STS motion with back load.

5.3 Clinical Implications of STS Motion with Backpack Load

Doorenbosch et al. and Valls-Sole et al. showed that the dominant muscles of STS motion while performing different strategies are: Tibialis anterior (TA), Gastrocnemius (GM), Soleus (SOL), Quadriceps (QUA), Hamstrings (HMS), Gluteus Maximus (GLM), Lumbar Paraspinal muscles (LPS), Sternocleidomastroid (SCM) and Trapezius (TRA) [4,5].

Using electromyography, STS motion was analyzed after performed using different strategies: standard, feet forward, knees move first, flexion of the trunk, head supported, trunk straight [5]. These authors showed that TA and SCM muscles are activated on the onset of movement whereas HMS, QUA, LPS muscles were activated in a sequence invariably in all conditions and SOL muscle was the last activated muscle which remained active simultaneously with QUA and HMS muscles during standing. Valls-Sole et al. concluded that LPS, QUA, and HMS were the main muscles controlling the STS motion.

The strategy the loaded subjects of the present study chose to apply during the STS motion may be referred to as a “trunk flexion strategy”. This is because the maximal hip flexion angle for the loaded cases is higher than that of the unloaded case. In previous studies, it was shown that such trunk flexion strategy utilizes the biarticular muscles around the knee [4]. Connected to the thigh and the heel, GM is a biarticular muscle. It is one of the braking muscles of late STS. In table 4.1 and 4.2 it was shown that until phase IV (starting from maximal ankle dorsiflexion), GM muscle is lengthened both distally and proximally. It is concluded that up to phase IV, GM is performing eccentric contraction, which is known make the muscle more prone to injury [39].

However, the trunk flexion strategy was shown also to utilize SOL muscle to a great extent [4]. SOL muscle is a powerful muscle with a slow contraction. It was also shown to have higher activity especially at the first 3 phases in the unloaded trunk flexion strategy than that of the standard STS motion [4,5]. Therefore, it is expected to have much higher activity in the loaded subjects, preventing them from falling forward. In the present research, the trajectory of joint angles show that in the first 3 phases of the standard STS motion, SOL muscle typically undergoes eccentric contraction due to distal lengthening. In addition to the SOL muscle the remainder of triceps surae muscles (GM and GL muscles) also contract eccentrically in the first 3 phases of the STS motion. Load causes these muscles to lengthen further leading to a more pronounced eccentric activity. Both muscles are expected to show a higher activity if the subject has a back load within the selected strategy. Therefore, their combined tendon (Achilles tendon) is likely to be loaded more with back loading and conceivably at unphysiological ranges if the back load to body weight proportion is increased.

It is concluded that, back loading causes employment of a trunk flexion strategy by the subjects and selecting such a strategy causes the triceps surae muscles and the achilles tendon to be more prone to damage among the muscle-tendon complexes of the lower extremities. On the other hand, a necessity of highly coordinated activity of the *LPS, QUA, and HMS* muscles is also indicated if the subjects have back loading during STS motion.

According to Howell and Barad [40], the knee of an average person is able to hyperextend up to on the average 10 degrees which is also considered as an indicator for hypermobility in Beighton Joint Mobility Index [41]. Kennedy and Grainger [42] reported the effect of extensive hyperextension on the posterior cruciate ligament (PCL) rupture. However, no such result was reported regarding the anterior cruciate ligament (ACL). It was shown that the posterior capsule is torn at a hyperextension angle of 30 degrees, which takes place before the rupture of PCL. However, Finerman et al. [43] showed that forced passive hyperextension of the knee (5 degrees of hyperextension angle and 10.0 Nm of hyperextension moment) imposes such high forces onto the ACL that the mean force on PCL is 23% of the mean force on the ACL. Moreover, the tibial torque and valgus-varus moment exert much higher forces on the ACL when the knee is in full extension rather than flexion. Because the PCL is stronger than the ACL, it can be concluded that hyperextension is unlikely to damage PCL without concomitant rupture of the ACL and the combination of factors causes the PCL to damage much less frequently than the ACL. Consequently, with increasing knee hyperextension, the ACL is torn first at an unknown angle, followed by the rupture of PCL and posterior capsule at a hyperextension angle of 30 degrees [44].

ACL injuries are seen children. In multiple studies, complete ACL tear was reported in skeletally immature children [45-49]. One of the factors that DeLee [50] has identified as a cause of children’s knee injuries is the greater number of children participating in organized sports. Operative treatment is usually postponed until physeal closure in order to avoid bone growth problems. However, the motions experienced during sports and daily life in such condition may cause even greater risks to the knee.

According to our findings, in the unloaded condition, the knee remains in flexion to a certain degree at the end of rise. However, in the 20% load condition, it shows hyperextension. Moreover, the knee is expected to achieve even greater hyperextension angles in the stabilization period. On the other hand, for the latter case the final knee moment increased considerably. Both factors cause more than two-fold increase on the forces acting on ACL [41]. Therefore, we conclude that increasing the back load may damage the ACL during STS motion especially for children with non-operated ACL problems.

5.4 Different Conditions to Study STS Motion

In the present study the unknown effects of back loading on the STS motion of children were investigated. The experiments were performed using a seat of constant height and the subjects were allowed to select their own speed in order to address a natural motion pattern. However, in the literature various other experimental conditions were applied to test their effects on STS motion. Such conditions and their possible implications for the present results are discussed below:

Seat height: Presently, the seat height was take as to be constant regardless of the knee height of each subject. However, it has been shown that changing the seat height leads to different kinematic and kinetic mechanisms for the STS motion [35-38]. The differences for hip and knee moments between high and low benches can be up to 60%, having a greater influence on the moments at the knee [35-38]. Therefore, the seat height affects biomechanical demands or leads to altered strategy. Nevertheless, in the present study, the seat height was taken as the height of a standard primary school bench representing the environment of the subjects involved. Such an experiment set-up is informative about the real kinematics and kinetics of the motion being performed.

Speed: The speed employed within the experiment was self-selected by the subjects. Our aim was to study a natural motion pattern and this approach served well for this purpose. It is found in earlier studies that the speed of motion is also a control parameter in order to adjust the maximal moments faced within the STS motion [51, 52]. As the speed of STS motion increases, the hip, knee, and ankle joints were shown to also increase in flexion, extension and dorsiflexion angles respectively [53]. In future studies, the effect of different speeds of STS motion on the kinematics and kinetics of subjects with back loading needs to be investigated. This can be done by standardized instructions or using a metronome to unify the speed of the motion and therefore to increase the comparability of the results [4, 27, 54].

Foot positioning: As the load increases, the initial ankle, knee, and hip flexion angles increase, the duration of the motion decreases. One reason for those is the posterior positioning of the feet or similarly anterior positioning of the hip on the seat. The same findings were shown by Shepherd and colleagues [55] on the condition of posterior feet positioning. The same study shows that with the posterior placement of the feet, hip flexion and hip flexion speed were lowered, whereas anterior placement of the feet increased the pre-extension phase [55]. These are in accordance with our results. The effect of posterior and anterior foot placement on STS motion with load can be studied in the future.

Trunk positioning: Initiating the STS motion with a trunk position other than vertical changes the kinematic and kinetic variables of the motion. During STS motion the momentum generated by the upper body is used during the extension phase [29]. Therefore, duration of the extension phase [56] and the total STS motion [5] are prolonged when the trunk initially is more flexed. Our findings are in parallel with the elongation of the extension phase, in contrast, the total STS duration remained the same with such a strategy. All these findings are in parallel to our research. Using the maximum flexion strategy, 27% lower knee moments were found. Therefore, it is obvious that more initial hip flexion does not mean only the adjustment of the new COM at the initial position, but it allows the subject to reduce the knee moment in further stages.

Arm motion: As in this case, the study of the STS movement is often done with constraints on the use of the arms. Restriction of the arms leads to a different ankle angular displacement, with a higher mean standard deviation with respect to the arms-free case. This suggests that more adjustment of the strategy of rising is needed at the ankle joint during restricted arm motion [57]. In our case, the arms were crossed on the chest, holding shoulders to minimize the effects of upper extremities and make the trunk act like a rigid body. The effects of arm motions can be addressed in the future after a classification of arm motions has been made.

Terminal constraint: The terminal constraint is the required body position or activity at the end of the STS movement [58]. The STS motion studied was aimed at still standing at the end. However, sit-to-walk and sit-to-run patterns, which are commonly seen in many occasions, can also be investigated in future.

Armrests: The seat used in the research had no armrests. Such an approach results in higher moments and higher neuromuscular coordination in the lower extremity. However, using armrests results in lower values at peak extension moments of knee and hip (approximately 50%) [59-61]. The effects of armrests on the STS with load can be analyzed in the future.

Backrests: The seats used in the experiment were backless to make the markers more visible to cameras. In general, backrests are usually used to standardize the initial position of the trunk. However, the focus of the research is on the strategies, extreme angles, moments and powers faced within the STS motion.

5.5 Conclusion

It is concluded that different loading levels affect different joints of the body during STS motion. For the 10% load case the highly affected joint is the knee joint. However, a further increase in the back load leads to a more pronounced effect in the ankle joint kinetics. As their activity affects the mechanics of both knee and ankle joints and as loaded STS motion causes them to contract eccentrically and at much higher lengths than in the unloaded case the gastrocnemius muscles are more prone to damage. Accordingly, transmitting much higher muscle forces than in the unloaded STS motion, the Achilles' tendon is indicated to carry a risk for damage.

As the back load is increased, the muscles experience a wider range motion and higher forces. This means that as the backpack load is increased, higher neuromuscular control is to be applied. Therefore, lack of neuromuscular coordination which may be the case for children with for example cerebral palsy is likely to worsen the loaded STS performance.

APPENDIX A. MARKER POSITIONS

Table A.1
Marker Positions of the Lower Extremity according to Davis Marker Placement Protocol

Abbr. Name	Full Name	Detailed Instructions
SACR	**Sacrum**	**Keep the center of marker to be on the reversed direction of "i" vector of pelvis LCS**
R, LASIS	**Ant. Sup. Iliac spine**	**Keep the center of marker to be on the direction of "i" vector of pelvis LCS**
R, LGTRO	**Greater Trochanter**	**On the greater trochanter of femur**
R, LTHI_W	**Thigh wand**	**A stick marker on the lower lateral thigh**
R, LTHI_L	**Thigh lateral**	**On the lateral thigh**
R, LTHI_A	**Thigh anterior**	**On the anterior thigh**
R, LLCON	**Lateral Epicondyle**	**On the center of lateral epicondyle of femur**
R, LMCON	**Medial Epicondyle**	**Only for static trial. On the center of medial epicondyle of femur. Can be found more easily if the subject's knee flexed a little**
R, LTIB_W	**Tibial wand**	**A stick marker on the upper lateral surface of lower leg. Around upper 1/3 point of fibular head to lateral malleolus, where is no tibial torsion component**
R, LTTUB	**Tibial tuberosity**	**On the tibial tuberosity**
R, LFH	**Fibular Head**	**On the fibular head. 2/3cm inferior to fibular head would be better to avoid merging with lateral condyle marker**
R, LLMAL	**Lateral Malleolus**	**Center of lateral malleolus**
R, LMMAL	**Medial Malleolus**	**Center of medial malleolus**
R, LMT	**Head of 2nd Metatarsal**	**On the second MT head**
R, LHEEL	**Heel**	**On heel**
R, LANKLE	**Ankle (estimated)**	**Estimated position of ankle joint**
R, LKNEE	**Knee (estimated)**	**Estimated position of knee joint**
R, LHIP	**Hip (estimated)**	**Estimated position of hip joint**

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