Final Performance Report

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Sudden Loading and Fatigue Effects on the Human Spine

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16. Abstract (Limit: 200 words) The effects of fatigue and load magnitude on the in-vivo torso response of human subjects to a sudden load were analyzed in six individuals. Parameter identification of the human spine was performed within a sudden loading, short duration, high intensity fatigue paradigm. The findings indicated that fatigue is an important factor under conditions of sudden loading. The fatigued subject does not respond in the same way as he does when not tired. It is suggested that a higher degree of injury would be expected under conditions of fatigue. Most of the time the worker is not at the point of fatigue, and therefore performance of certain tasks is usually practiced under conditions which are better than those prevalent under conditions of fatigue. Two reports are being produced from the data gathered in this study. The first will examine the electromyographic responses of the individual and compare them to the dynamics of the torso. The second study will mathematically study the compression levels expected as a result of earlier work. The authors suggest that their most important finding was the ability to mathematically model a complex response of the torso to a sudden load with a simple second order linear model. The study demonstrated that short duration, high intensity muscular fatigue changes the way in which the torso responds to a sudden load.

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Significant Findings

Grant # 1 R03 OH02821-01: Sudden Loading and Fatigue Effects on the Human Spine.

The significant findings of this research can be grouped into three basic areas: 1) measurement techniques, 2) mathematical modeling and 3) *in vivo* subject results. The Torso Motion Measurement System (ToMMS) and the sudden loading paradigm was successful in predicting mass moments of inertia. The estimates for the mechanical analog were very close to those predicted mathematically. The torso results were lower than those predicted mathematically, as expected, because the mathematical estimates relied upon the rigid body assumption while the videotaped results showed the subjects' torsos violating the rigid body assumption (the entire head, arms and torso (HAT) structure was not accelerated uniformly). The ToMMS also accurately measured the sagittal-plane kinematics of the torso resulting from a sudden load. It is robust enough to be recommended for future studies involving non-sagittal-plane kinematics with minor modifications.

The most exciting finding of this research was the ability to mathematically model a complex response of the torso to a sudden load with a simple second order linear model. Identification attempts were made with model orders of four and higher. The fourth order models (chosen to account for non-uniform acceleration of the HAT) did not have any additional predictive power over the second order models. When model orders higher than four were tried the problem became numerically ill-conditioned. The results are consistent with previous single-joint modeling efforts which indicate that the body actively controls many parameters so that the damping ratio remains quite consistent across a variety of inputs and conditions. However, these results showed that the damping ratio remained constant, at approximately 0.1, whereas the single-joint research indicates that the damping ratio remains between 0.2 and 0.4.

There are some questions to be resolved concerning adding non-linear elements to the second order model (e.g., modeling the change occurring approximately 500 milliseconds into the response), but, in general, it performed quite well. Even with a very strong subject effect ($p < 0.0001$) significant differences ($p < 0.05$) could be determined between the two controlled parameters of force input level (15% and 20% of the subject's body weight) and muscular fatigue (before and after fatiguing exercises) across a wide variety of measured and mathematically derived parameters.

Usefulness

Grant # 1 R03 OH02821-01: Sudden Loading and Fatigue Effects on the Human Spine.

More accurately modeling the parameters of torso motion enables researchers to refine their predictions of the torso's abilities and constraints. Being able to predict the motion of a system such as the torso allows researchers to make other predictions of how other variables will behave under similar circumstances. For example, minimum compression levels in the spinal column can be determined given that motion. Then the effects of additional factors on spinal column compression, such as levels of co-contraction or differing moment arms of actuating muscles, can be determined in light of knowing the minimum compression levels.

Additionally this study showed that short-duration, high-intensity muscular fatigue does change the way in which the torso responds to a sudden loading. Future research may be able to use this information when determining work-rest cycles or probabilities of injury.

Final Report

This research analyzed the *in vivo* torso response of human subjects to a sudden load. The effects of fatigue and load magnitude were examined with respect to that response.

The paper entitled, "Sagittal plane dynamics of the spine *in vivo*: Parameter identification under sudden loading and fatigue conditions," follows this section. It models the torso response mathematically and shows how those models change under changing load magnitudes and fatigue.

Based on the findings of this research it can be said that fatigue plays a important role under sudden loading conditions. The response of the fatigued subject is not the same as when unfatigued. If the fatiguenecessary response is complex and rarely encountered, then a higher probability of injury would be expected to be associated with it. This is from control theory which predicts different strategies for different control situations and a greater likelihood for instability when delays are introduced into the system. Additionally, a worker spends most of their time performing, and therefore practicing, the response under the unfatigued condition. Under a fatigued condition the margin of error is smaller and the human is less practiced, therefore the statement about a higher probable injury rate is justified.

After examining the data two more papers are being researched and written using these experimental findings. The first examines the electromyographic (EMG) responses and compares them to the dynamics of the torso. The second mathematically examines the compression levels which would be expected as a result of the previous two studies. The abstracts are included and copies of both papers will be filed as soon as they are submitted to the proposed journals.

Equipment Inventory

Dynamic spinal loading - The effects of sudden loading and muscular fatigue Grant Award Number 1 R03 OH02821-01

One National Instruments DMA-2800 Digital Memory Access board was purchased and received on 2 October 1991. The cost of the item was $1345.50, including the necessary RTSI cable. It was paid for entirely out of this grant is in good condition after the experiments have been finished. It is located in the Ergonomics Laboratory in the Department of Mechanical Engineering at the University of Utah.

Sagittal plane dynamics of the spine *in vivo* Part I: Parameter identification under sudden loading and fatigue conditions.

by

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ABSTRACT

Parameter identification of the human spine in vivo was performed within a sudden loading/short-duration, high-intensity fatigue paradigm. Mass moment of inertia estimates showed consistency across all subjects. The parameters of a second-order linear system had as much explanatory power as a third- or fourth-order linear models. There are definite changes in control methodology before and after fatigue onset as well as a definite change during each trial's response, based upon the results of this study.

Six subjects were restrained such that sagittal plane motion was limited to that above the L5/S1 joint. Two square pulse magnitudes (15 and 20% of body weight) were applied to the center of mass of the head, arm, torso (HAT) over a 250 millisecond period during and after which the resultent motions were recorded (for a total of 4000 milliseconds). Five trials of each magnitude, randomly determined and randomly presented, were given preceeding a set of torso muscle fatiguing exercises. After the fatiguing exercises the five trials of each magnitude, randomly determined and randomly presented, were again given

for a total of 20 trials per subject. Mass moments of inertia were estimated from the initial motion conditions. Several parametric models were then computed and compared. The parameters were estimated using standard least-squares system identification techniques. Similar response patterns were observed across all subjects, although there were significant differences between subjects ($p < 0.0001$). There were also significant differences in the magnitude of impulse ($p < 0.05$), the response timing ($p < 0.05$) and pre- and post-fatigue ($p < 0.05$).

INTRODUCTION

Why Study the Low-Back?

Low back pain will afflict 8 out of 10 persons in their lifetime. Jensen (1988) discovered that 25% of the workers compensation claims, in the United States, were for back injuries. $14 billion was estimated to have been spent on treatment and compensation for low back pain in 1976 (Snook, 1987). This figure only included the direct costs. Newer figures from Andersson *et al.* (1991) place Snook's estimate between $28.6 and $56 billion. Prevalence and cost alone justify intense research into low-back injury mechanisms.

The development and validation of static models of the low-back has lead to prevention strategies in occupational, clinical, rehabilitation and home situations (e.g., see Chaffin and Andersson, 1984). Static models, however, are quite limited in their usefulness under dynamic conditions.

When an existing structure is examined under static conditions stability is assumed. Proceeding to dynamic analysis from static requires a definition of stability both physiologically and mathematically. Frymoyer and Krag's (1986) definition of spinal instability, "[It] is a condition in which the anatomic structures are disrupted, so that loads that are normally tolerated result in excessive or abnormal spinal motions, displacements, or strains, causing the development of progressive deformities," is sufficient for the physiologic definition. The mathematical definition is implicit in the control theory and system identification approach taken herein.

Most low-back injuries are chronic in development as suggested in Frymoyer and Krag's description above, but the onset of pain typically manifests itself as a result of an acute episode. This can occur when someone looses the control of their balance and quickly recovers their posture or when some variable a person is controlling, such as the velocity of a lifted object suddenly changes (i.e., the object starts falling and they quickly react to recover their control over the object). Manning, *et al.* (1984) reported that 12% of all back injuries in a British automobile plant were attributed to, "sudden, unexpected load[ing]." Since the body does not react correctly or fast enough in these situations it indicates the control mechanism is not sufficient for preventing injury.

Recent experiments have examined the preparatory responses of humans when subjected to a sudden unexpected load (Marras, *et al.*, 1987). They demonstrated that the duration of the muscle response increased as the unexpectedness increased, the muscle force (measured indirectly through electromyograms) was highest under the unexpected condition and the muscle forces under sudden loading were all much higher than under the static conditions.

Dynamic Posture Modeling

Dynamic upright posture requires coordinated functioning between many structures and systems. Posture analysis, therefore, has been used for modeling and prediction for many years. The majority of posture studies performed in recent years have examined standing posture (see Johansson and Magnusson, 1991 or Agarwal and Gottlieb 1982, 1984a, 1984b).

Mathematical modeling of dynamic posture earnestly began in the 1960s (Vukobratović and Juričič, 1969). Vukobratović and Stepanenko (1972) used an inverted pendulum model as the basis for examining human gait. Dynamic posture greatly affects the kinematics and kinetics of gait. Two kinds of stability were proposed, internal synergy (recovery from small perturbations) and external synergy (recovery from large perturbations via kinematic and kinetic changes). A minimization between the state parameters and the desired parameters was performed mathematically. Chow and Jacobson (1971 and 1972) constrained the inverted pendulum to just the torso for closer examination. Hemami, *et al.* (1973) expanded Vukobratović and Stepanenko's model by expanding the number of types of nonlinear feedback and showing the model's range of inherent stability. Further refinement and simulation was done by Hemami and Golliday (1977) and Hemami and Katbab (1982).

Katbab (1989) added muscle models to the previous mathematical models making them more human-like. Using muscle model parameters from various *in vitro* and *in vivo* studies, he showed that he could easily maintain the torso in an upright posture.

Raibert (1986), among other researchers, has utilized the basic ideas of these algorithms for controlling walking and hopping robots. These have also been used for system identification of standing posture, but none of these models have been used, to date, for system identification of just the low-back *in vivo*. Johansson, *et al.* (1988) and Maki (1986) both showed that identifying whole body posture parameters using linear models produced reliable results.

Most of these models are variants of a linear (or linearized) second-order model. This structure indicates that there are 4 parameters which need to be estimated. The model contains, in mechanical terms, the input ($X(s)$), the output($Y(s)$), the inertia (I), viscous damping (B), the spring constant (K) and the static gain (G, with $G = 1$ in the simplest formulation). In Laplace form, the model is,

$$\frac{Y(s)}{X(s)} = \frac{G}{Is^2 + Bs + K} \tag{1}$$

in the simplest case, or more generically, substituting $G = \frac{1}{I}$ and rearranging for

$$\frac{Y(s)}{X(s)} = \frac{G}{s^2 + \frac{B}{I}s + \frac{K}{I}}. \tag{2}$$

The natural frequency, ω_n, and damping ratio, ζ, can be calculated by matching terms of the denominator of Eq. (2) to the denominator of the second-order characteristic equation, $s^2 + 2\zeta\omega_n s + {\omega_n}^2$.

Estimating the Mass Moment of Inertia

Humans have been estimating their mass moment of inertia for almost as long as the field of dynamics has been in existence, the methods, however, have become much more refined in recent years. There are three major methods for measuring the mass moment of inertia of human segments (Roebuck, *et al.*, 1975): 1) mathematical estimations describing the distribution of the mass-density product, 2) physical models of the segement based on the mass-density distribution and analyzed by the compound or torsional pendulum techniques and 3) experimental methods using rotational acceleration, or the so-called quick-release method, where a force is placed on a segment a known distance from its point of rotation and is then removed suddenly while the resulting acceleration is measured. The inertia is then computed from this information and, when necessary, incorporating the parallel axis theorem. The method used for this experiment is a modification of the quick-release and is developed in more detail in the methods section.

Some of the earliest and most widely adopted results are those by Dempster (1955) who measured volume, mass, density, location of mass center and mass moments of inertia of cadaver segments. Drillis and Contini (1966) made detailed measurements on live subjects including the radius of gyration of different segments and then compared their results to those from cadaver studies and purely mathematical methods. Their data is also widely utilized.

More recently the mathematical estimation methods have improved as the computing and imagining power have increased. Recent studies by Clauser, *et al.* (1969), Hatze (1980), Yeadon (1990) and Hinrichs (1990), for instance, demonstate the effectiveness of the computational methods for determining mass moments of inertia. Mungiole and Martin (1990) used magnetic resonance imagining combined with density estimates for different structures on the lower extremities of athletic subjects. They found their method overestimates the mass moment of inertias when compared to cadaver and earlier *in vivo* data. This might be expected as the earlier studies were not as strictly controlled for the subject's athletic level and both age and activity levels were not matched when cadavers were used.

The major disadvantage to the mathematical methods is that they do not typically or effectively account for the human subject not being a rigid body. Modeling humans is difficult under the best conditions. Experimental measures are, therefore, still preferable for mass moment of inertia estimations.

Estimating Other Model Parameters

Sudden loading implies an impulse, pulse or some other well defined input to a system. Input-output modeling of systems is best handled using the techniques and methodologies of system identification (e.g., Marmarelis and Marmarelis, 1978, Sinha and Kuszta, 1983, Ljung, 1987, Kearney and Hunter, 1990, Ljung, 1991). One of two modeling categories are used in system identification. Those two are: 1) parametric, or model-based methods, and 2) non-parametric, or frequency response-based methods. This research employs the former as there have been many theoretical parametric models of the low-back developed in recent years (see previous section).

Kearney and Hunter (1990) provide a most comprehesive overview of research into single-joint parameter estimation in humans. They listed several interesting aspects of identifying the model parameters of single-joints. Second-order linear models are quite powerful and provide much information. They and other researchers have shown that the damping ratio (ζ) is relatively constant across a number of different joints and through a wide range of forces about those joints. The body accomplishes this by adjusting the the elastic (K) and viscous (B) parameters such that ζ remains approximately constant between 0.2 and 0.4. Another, expected, result is that the systems transient response scales to the gain, while the response duration is proportional to the natural frequency (ω_n).

Results of several whole body posture studies, mentioned earlier, indicate that the results across multiple joints should be similar to those of single joints.

Muscular Fatigue

The origins and effects of fatigue, although quite prevalent and much studied, remain relatively unknown. The general effect of fatigue is one of a reduction in muscular strength. This may result in slower reaction times and thereby reactions which are insufficient for maintenance of stability. Control theory demonstrates that when delays occur in a control loop the probability of maintaining stability decreases. The resulting actions require more effort which, in turn, produces more fatigue and the result is, in all likelihood, a greater probability for injury.

The first problem encountered when studying fatigue is determing what variable(s) to study. Grandjean (1979) glumly noticed, "[Fatigue] denotes a loss of efficiency, and a disclination for any kind of effort, but it is not a single, definite state." Cameron (1973), in a historical review of the literature on fatigue, divides fatigue into acute and chronic where acute is measured in terms of performance measures and chronic is subjectively influenced. Many more concise variables have been proposed, for example: changes in electromyograms (De Luca, 1984), metabolic byproduct changes (Fitts, 1988), changes in microcapillary pressure in muscles undergoing isometric contractions (Styf, 1987), changes in drug concentrations (Gullestad, *et al.*, 1989) and changes in muscle force outputs (Chaffin, 1973). The last is the variable chosen as being representative of the fatigue process for this study.

The second problem is determining a definition for fatigue once the kind of fatigue has been chosen. Many definitions have been put forth. Kirkendall (1990) chose, "...the failure to maintain an expected power output." Cameron (1973) proposed that fatigue be measured by the time necessary to recover the function lost or returned to its initial level. The problem is further complicated under this definition because the function's baseline must be determined in a self-referential fashion – what does a reseacher use as a metric for determining the subject's unfatigued state so the researcher can make baseline measurements of the new fatigue variable? One must also keep in mind that the definition of fatigue is closely tied to what is being measured and how it is being measured.

Choosing changes in muscle output sidesteps the question as to whether the fatigue is produced by the central nervous system, the transmission system or the muscle itself. Changes in muscular output can be further broken down into decreases in force production (e.g., Grabiner and Jeziorowski, 1991, Chaffin, 1973, or Hitchcock, 1989) or changes or delays in response timing (e.g., Siler and Martin, 1991, Povin and Norman, 1992, Lance and Chaffin, 1969, Christakos and Windhorst, 1986, Kroll, 1973, Pagala *et al.*, 1984, Kearney and Hunter, 1990). The present experiment combined both methods. Force can be measured at a specified point in time or it can be averaged or summed over a specific time period. Muscular fatigue is generally defined as the percentage decrease in time for maintaining a given posture. A fatigued state was defined as the inability of the body to maintain a posture for a predetermined percentage of time based on a previous measure of the unfatigued state. The difference in the fatigued states was then used for determining changes in forces and application times of those forces occurring because of muscular fatigue.

The final question to be answered concerning fatigue is the method by which it is induced. The experimental paradigm used was a short-duration, high-intensity fatigue. Since the goal was the demonstration of an effect, it was felt that this was the most efficient method for producing that effect. There is not much information in the literature as to how the fatigue is affected, as defined above, by the rate at which it is acquired.

METHODS

Subjects

Six athletic male subjects were paid for participation in this study. Athletic was defined as participating in aerobic exercise three times or more per week in addition to participation in a sport at least once per week. Strict selection criteria resulted in a homogeneous group. It was hoped that such a criteria would reduce inter-subject variations. Relevant subject anthropometry and characteristics are in Table 1, below.

Table 1: Relevant subject data.

Subject No.	Age (years)	Weight (kg)	Height (m)	HAT Mass Location Height above L5/S1 (m)
1	25.25	81.64	1.734	0.358
2	24.00	64.86	1.721	0.355
3	24.75	60.78	1.683	0.347
4	24.75	65.77	1.727	0.356
5	30.58	90.71	1.886	0.389
6	25.58	61.23	1.702	0.351
mean	25.65	70.83	1.742	0.359
st.dev.	2.45	12.38	0.073	0.015

Each subject participated in the experiment over 6 separate days. The first day consisted of a question and answer session with one of the researchers in which the subject candidate was informed about the study. If they wished to continue, they then signed the University of Utah Institutional Review Board subject participation consent forms. Next the subject was scheduled for a physical exam. Each subject was screened for back injuries or disorders, other musculoskeletal problems and any vestibular or other balance related anomalies by Occupational Medicine M. D.s. Only after receiving confirmation that the subject was physically fit for participation were they scheduled.

Once the preliminaries were complete, the study ran for 4 consecutive days. The subject arrived on day one (the third day of participation) and was tested for both anterior and posterior time-to-fatigue. This was done using a modified Sorenson fatigue test with a 5 cm drop as the criterion for fatigue (Biering-Sorenson, 1984). The subject was supported on an physical examination bench by the legs and pelvis while leaving the head, arms (crossed over the chest) and torso hanging in space. This test consists of timing how long the subject can maintain the horizontal posture. It was performed first in a prone position and then repeated in a supine position. It was repeated identically on the second day. Only one trial of each was performed on these test days. After the fatigue test on the second day the Torso Motion Measurement System (ToMMS) was adjusted to that particular subject.

Gerber (1991) showed that the modified Sorenson test was strongly affected by a learning curve. If the first day's results were removed from the data the reliability coefficient went to 0.96 from 0.76. Therefore the second day's results were used as the baseline for the modified Sorenson test. Day three the subject rested. This increased the likelihood that fatigue from the previous two days (e.g., delayed onset muscle soreness, etc.) would be minimized. On the fourth consecutive day the experiment was performed.

The test day was comprised of subject preparation, unfatigued subject data collection, fatiguing the subject, testing the subject's fatigue level and data collection of the fatigued subject (all explained in more detail below). Preparation consisted of warmup exercises (20 repetitions of side bending with no weights, 5 repetitions of stretching anterior and posterior and 5 repetitions of right and left leg stretches), skin preparation and placement of surface electromyograph (EMG) electrodes, fitting of a cervical collar (covered foam rubber with velcro attachment), assignment of hearing protection (it served a dual purpose: filtering out the hydraulic power supply noise while reducing external disturbances) and adjustment of accelerometer and pulse application shoulder harnesses.

The unfatigued (and fatigued) subject data collection was done after the subject had been lowered into the ToMMS (see Fig. 1 and Fig. 2). Padding was added to the cast section of the ToMMS, where necessary, insuring minimal subject movement. When the subject was quite snugly fit, they were asked to move about and the researchers ensured that there was minimal translatory motion of the pelvis.

The instrumentation arms were attached to the inner harness and the cable from the hydraulic actuator was attached to the outer harness (see Fig. 2). The EMG and accelerometer cables were attached and all of the instruments were calibrated.

A safety checklist was then completed. This included checking that the subject was correctly positioned and connected as well as a general mechanical check of the instrumentation and hydraulics. Next the subject instructions were repeated, "Attempt to remain in an upright relaxed posture, " and several test pulses were presented. Last minute system checks were performed during this preview session.

Each subject was positioned such that they were standing comfortably erect with their forearms crossed in front of their torso just above their umbilicus (see Fig. 1 and Fig. 2). An absolute upright reference was not used as it was feared that the subjects would spend most of their time aligning themselves with the reference and thus altering their response to the pulse. Additionally, the timing between pulses would have been more variable because of the wait while the subject adjusted to the reference point.

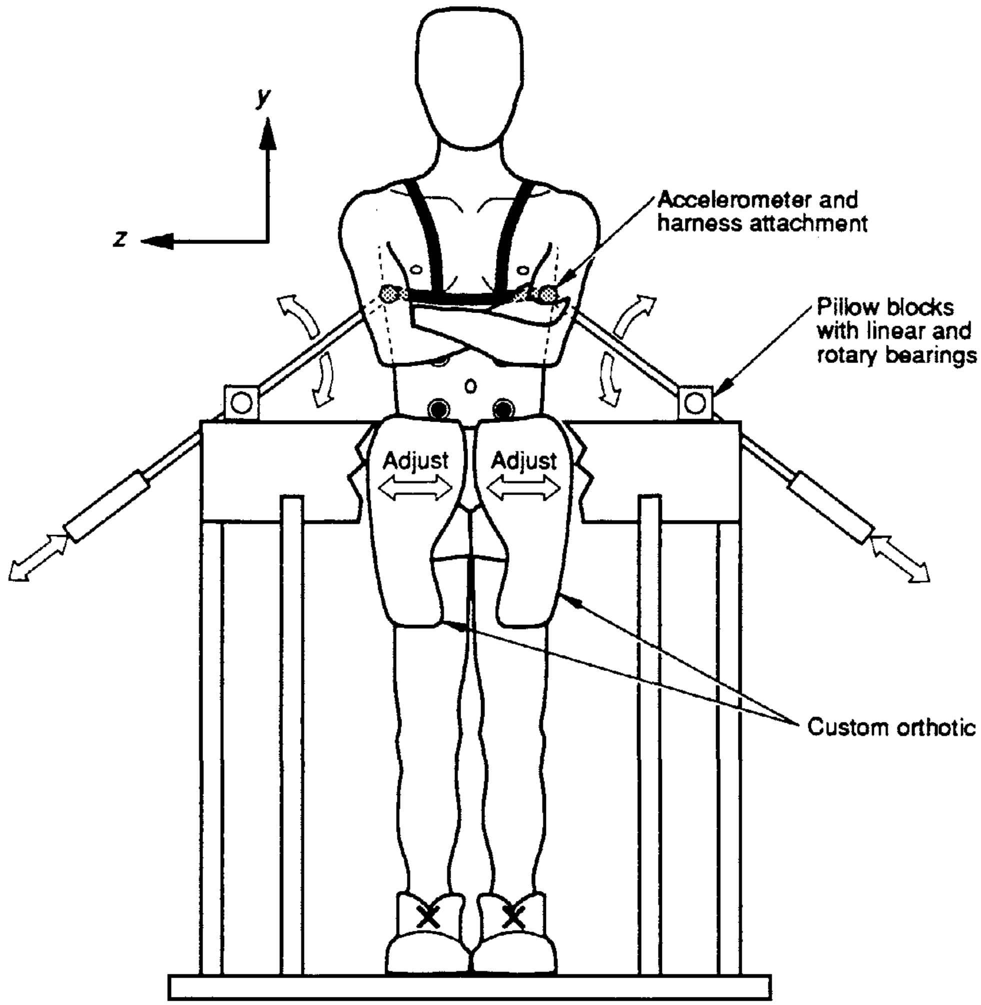

Figure 1: Front view of ToMMS (outline arrows indicate the measured degrees of freedom and the adjustability of the contraint).

The subject was presented with 10 trials consisting of a randomly determined mix of five pulses at an amplitude of 15% (low) of the subject's body weight and five pulses at 20% (high). Each trial consisted of a 90 second wait (to minimize fatigue) followed by a random 5-30 second wait (for between trial pattern interruption) during which time there was preload on the cable (approximately 5% of each subject's HAT weight). This preload

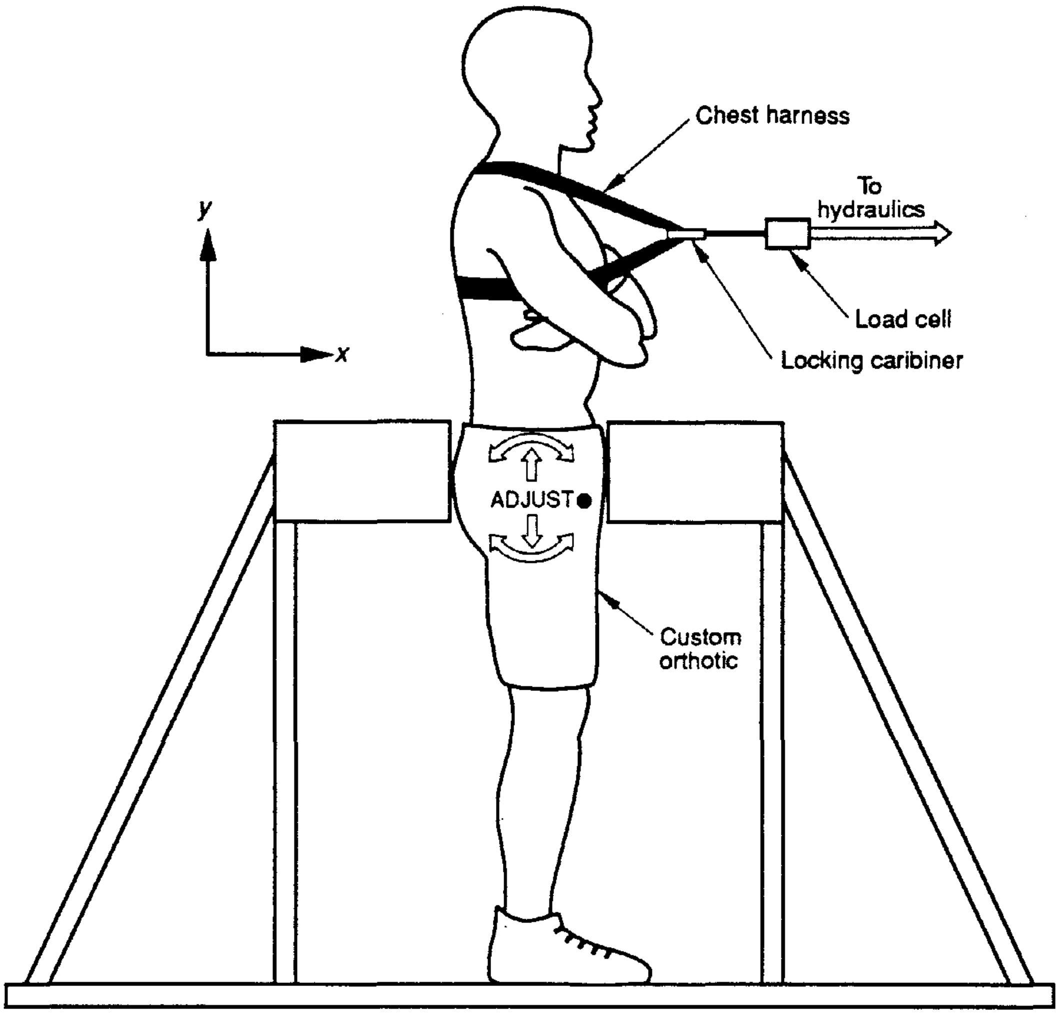

Figure 2: Side view of ToMMS (this also shows the connection to the hydraulic system).

minimized injury and provided the added benefit of somewhat stiffening the torso before the pulse was applied. After the 5-30 second random wait came the pulse for a duration of 250 milliseconds. It was followed by 3750 milliseconds with no force or preload on the cable. This was so the subject was unloaded when responding to the pulse. The slack in the cable was removed, after the quiesence, by the preload thus signifying the beginning of the next trial. Fig. 3 shows, schematically, the timing of events for each trial.

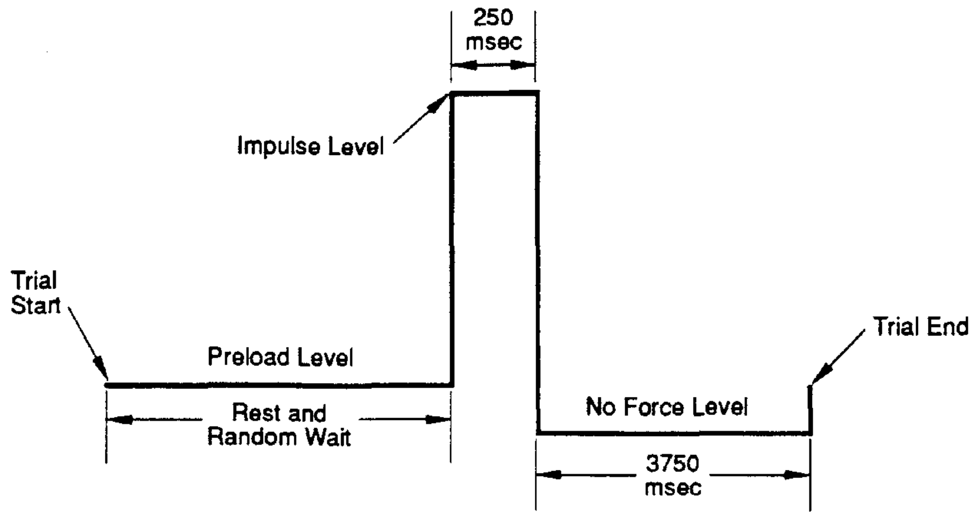

Figure 3: Schematic timing diagram (not to scale).

After the ten unfatigued trials the subject was removed from the ToMMS and was directed to the exercise area. Here they performed the short-duration, high-intensity fatigue exercises. These consisted of performing as many as 20 repetitions of standing torso rotations (about the body axis) with the legs approximately shoulder-width apart while holding $20kgf$ weights in each hand, 20 back extensions, 20 "stomach crunches" or stomach flexions, 20 oblique "crunches" on both the left and right sides. All extensions and crunches were performed on a Roman Chair (Ed's Athletic Supply, Salt Lake City, UT). Each set of 5×20 was referred to as a circuit. The curcuit was repeated once. The subjects were then tested performing the modified Sorenson Test. The original protocol called for additional repetitions of the circuit if any subject was able to maintain more than 50% of their unfatigued time, based on the second consecutive day's result of the modified Sorenson test. None of the six subjects were able to maintain more than 50% of that time while performing either the anterior or posterior parts of the test. After the fatigue test they were allowed up to a quart of water.

The fatiguing exercises took approximately 30-35 minutes. All subjects were placed into the ToMMS after 35 minutes and before 45 minutes had passed from the unfatigued trials. Subjective responses after the fatigued trials indicated that all subjects were "tired" during the fatigue test compared to during the unfatigued test.

Instrumentation

The ToMMS is a symmetric constraint and instrumentation system. It constrains the subject such that any torso rotations occur at or above the L5/S1 joint and it accurately measures those motions (see Fig. 1 and Fig. 2). The constraint is a partial plastic scoliosis cast (custom-built by Fit-Well Prosthetic and Orthotic Center, Salt Lake City, UT and further modified by one of the researchers) rigidly attached to a metal frame which, in turn, is bolted to a concrete floor. The natural frequencies of the system are well above this experiment's excitation frequencies or natural frequencies of the human spine *in vivo* (Pope, et al., 1987).

The input pulse was provided by a hydraulic power supply (HydroPak 5 hp, see Fig. 4 for a schematic of the experimental setup) which provided hydraulic fluid at 13.79×10^3 kPa (2000 psi) to a servo-valve (Raymond Atchley, model# 410 755) connected to a double acting actuator (Clippard Minimatic 7D-1). The servo-valve's action was directed by a controller (MTS 406.11 Controller) which was slaved to the data acquisition and control system (described later in this section).

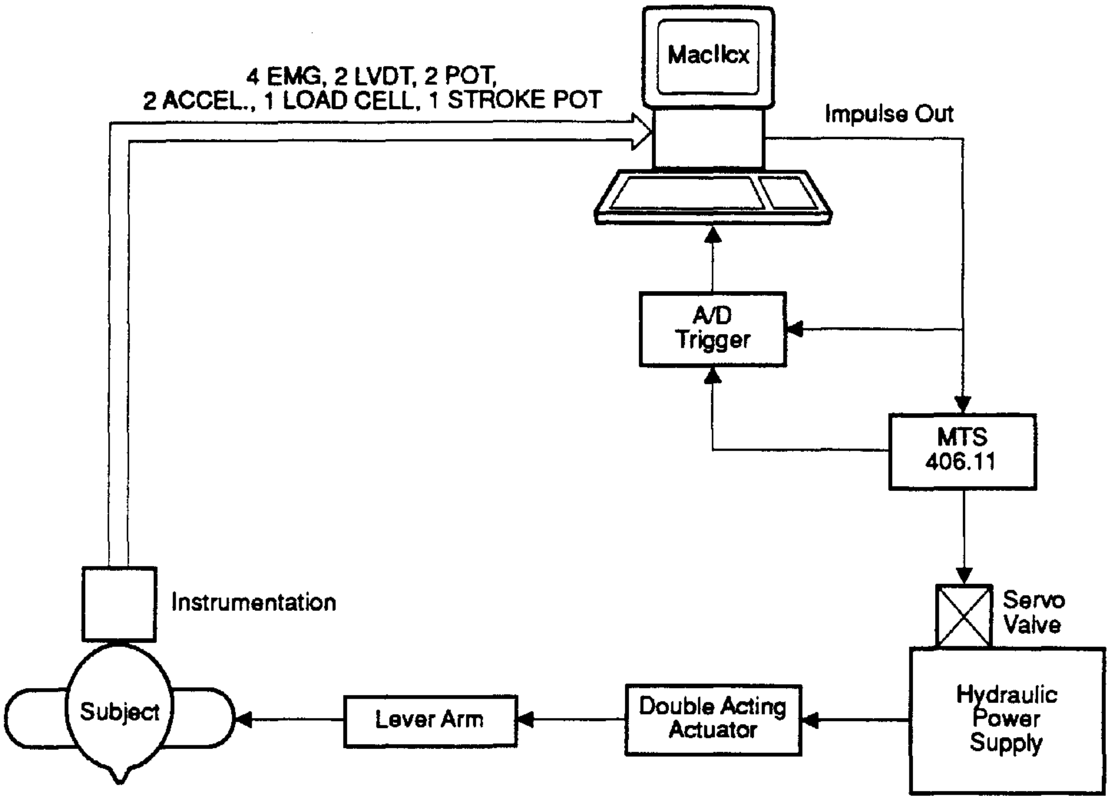

Figure 4: Schematic drawing of experimental setup.

The actuator was connected, via wire rope, to lever arm with a 10:1 reduction in mechanical advantage. This had multiple benefits. It first decreased the worst-case failure into approximately a 890 N pulse into the person. Secondly, it reduced the necessary travel of the actuator which allowed a faster system response. Finally, it allowed easy instrumentation from which to record the travel of the pulse cable.

The pulse cable incorporated an additional safety feature, its breaking strength was approximately 445 N. A load cell (Interface SM-250) was inline and allowed direct measurements of the force pulse profile applied to the subject. After the load cell was a locking caribiner connected to a climbing shoulder harness (see Fig. 2). The harness was adjusted so that the force applied was parallel to the ground and intersected the subject at the HAT center of mass.

A second harness on the subject was custom-made (Lone Peak Designs, Salt Lake City and one of the researchers). It had two ball ends of a ball joint attached to it (see Fig. 5). These were located at opposite sides of the harness (mirror-imaged w. r. t. the sagittal plane) and it could be adjusted such that each ball end was at the subjects center of mass for the HAT symmetric w. r. t. the frontal plane. Attached to each ball end was a $\pm 5g$ translational accelerometer (Entran model EGA-125). Ball sockets were attached to stiff, slender, polished metal rods. These were fed through a linear bearing and pillow block combination. Attached to each pillow block was a DC-DC Linear-Variable Differential Transducer (LVDT, Trans-Tek, DC-DC series 240 model 0246-000). Measurements and movements of the subject bending or extending their torso (±y, using the ISB orientations, ISB, 1992) were now possible, as were measurements of frontal plane motion and body roll or rotation about the long axis of the subject's torso (rotation about +x and +y, respectively). The pillow block was attached to a horizontal pillow block and bearing system which allowed sagittal plane rotation (about the x-axis) of the rod and frontal plane rotation about the z-axis as well as linear motion in anterior/posterior directions (±x). This last motion was fixed for each subject. The rotation in the sagittal plane was measured with a 10 KΩ rotary potentiometer on each pillow block. This instrumentation fixed the subject's hip and thigh while allowing the free movement of the torso in any direction without fear of damaging the measurement devices.

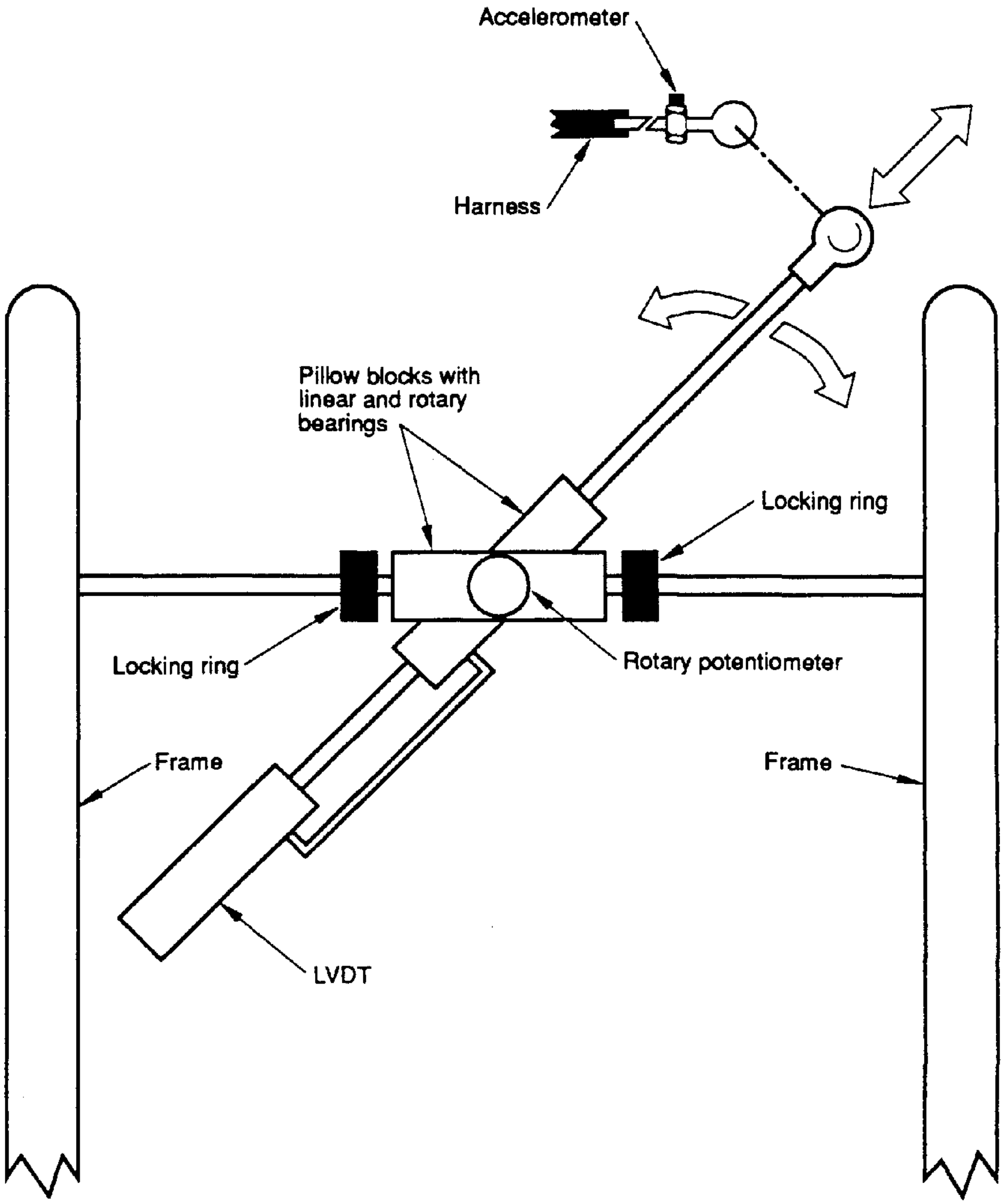

Figure 5: Instrumentation and attachment view of ToMMS (the outlined arrows indicate the degrees of freedom in this plane).

The mirror image design of the measurement system was utililized for several reasons. Differences between measurements are used when determining if the sagittal plane motion assumption was violated. If one system breaks during an experimental trial, the rest of the experiment can be run without interruption. This design also extends the ToMMS's usefullness for future experiments including those examining non-sagittal plane motions of the torso.

Mechanical Analog

A mechanical analog of the subject's HAT was used for checking the estimation ability of the ToMMS. The mechanical analog was scaled by a factor of 5 smaller due to equipment constraints. The results indicate that this may not have the adverse impact it was feared it might. It consisted of two rectangular weights supported on a wooden platform with its center of mass 0.3048 m above its point of rotation (total weight was 12.3 kg and the dimensions were $0.3556 \times 0.1524 \times 0.1778$ m) mounted on a slender metal rod (6.35×10^{-3} m in diameter). The rod was rigidly attached to the ToMMS metal frame. The orientation of the rectangular weights was with the longest dimension parallel to the $+z$ axis and the shortest was parallel to the $+y$ axis. The mass moment of inertia of the mechanical analog in the inverted pendulum configuration (assuming only rotation about the point of attachment) was 1.1725 $kg\ m^2$.

The same harnesses used for the subjects were used when connecting the mechanical analog to the ToMMS. The preload and pulses were determined using the same ratios as for the subjects insuring minimal scaling differences (see next section for details). The only dimension not scaled was the distance from the point of attachment to the center of mass of the weights. This was kept at 0.3048 m for mechanical considerations when using the ToMMS.

Data Collection & Reduction

The data from the rotary potentiometers, accelerometers, LVDTs, the stroke potentiometer and the load cell were all independently amplified by custom-built analog instrumentation amplifiers. Outputs from these amplifiers were fed into a National Instruments NB-MIO-16 a/d board inside an Apple Macintosh IIcx (see Fig. 4). This board and its companion board, National Instruments NB-DMA2800 dma controller, sent the pulse waveform to the hydraulic controller, triggered the sampling system and collected all 12 channels of data at 1250 samples per second. Because the boards could be configured for only one sampling rate all data was collected at the sampling rate of the EMGs. That sampling rate was 1250 Hz.

The data was recorded on disk in a binary format after each trial. There was no delay while the data was being recorded as the trial timer was independent of the data recording.

A FORTRAN program read the data from disk, calibrated each channel and then performed some initial smoothing. Any DC-offset was removed and a $3^{\underline{rd}}$ -order, low pass Butterworth filter, with a cutoff frequency of 50 Hz., was run forwards and backwards (bidirectional IIR filter) on the data. The bidirectional approach produced a filtered data set with no phase lags (Oppenheim and Schafer, 1989). The distance, r, to the subjects center of mass from the point of rotation about their L5/S1 joint then was adjusted using the LVDT data.

Finally the beginning and end of the pulse were determined. The algorithm used was a simple threshold where the pulse's start was determined to be the first point past a given magnitude level. The minimum threshold which found unambiguous starting points for all 120 trials was determined by a trial and error approach. The threshold chosen and used was when the force's input magnitude exceeded 10 N. The justification for doing this is first for consistency and second was to reduce any data contamination as a result of the preload. The drawback to this method is that approximately the first 100 milliseconds were not used for the parameter estimates. The inertia estimate routine used a different method which included data from the moment the pulse was sent out from the data acquisition and control computer.

The mass moment of inertia was calculated from

$$M = I\ddot{\Theta} \tag{3}$$

where M is the initial moment, I is the mass moment of inertia and $\ddot{\Theta}$ is the initial angular acceleration.

An expression for the moment in terms of the Force (recorded from the load cell), F, r, and an expression for the initial angular acceleration in terms of r and the initial linear acceleration (recorded from the accelerometers), a, were substituted into Eq. (3). This resulted in

$$Fr = I\frac{a}{r}. \tag{4}$$

Solving for I yields,

$$I = \frac{Fr^2}{a}. \tag{5}$$

Another FORTRAN program produced Eq. (5) as its output over the initial 0.75 seconds of each trial starting from when the pulse command was sent to the controller and the trigger. Determining which part of the curve to use was not easy. Several algoritms were attempted, including peak searches on second derivatives, before a piecewise linear curve-fit was utilized. First F, r and a data streams were filtered using a $5^{\underline{th}}$ -order, bidirectional Butterworth filter with a cutoff frequency of 10 Hz. The anthropometric information from Drillis and Contini (1966) and from Pheasant (1988) were adjusted to account for the L5/S1 joint being the point of rotation instead of the hip. This was done by assuming a constant density for the material of the torso above and below the L5/S1 joint. This assumption was used because the amount of movement of that section was very small compared to the rest of the body, it being closest to the point of rotation.

The first points of the output varied widely, but quickly formed a steep negative slope (see Fig. 6). A least-squares line was fit to these initial points. When the coefficient of determination (r^2) dropped below 0.98 another line was estimated. This was repeated until the data stream was complete. In a majority of cases (98 out of 120 trials, 6 subjects $\times$ 10 unfatigued + 10 fatigued trials) this resulted in three lines. The inflection point between the first and second line was chosen as the initial mass moment of inertia value.

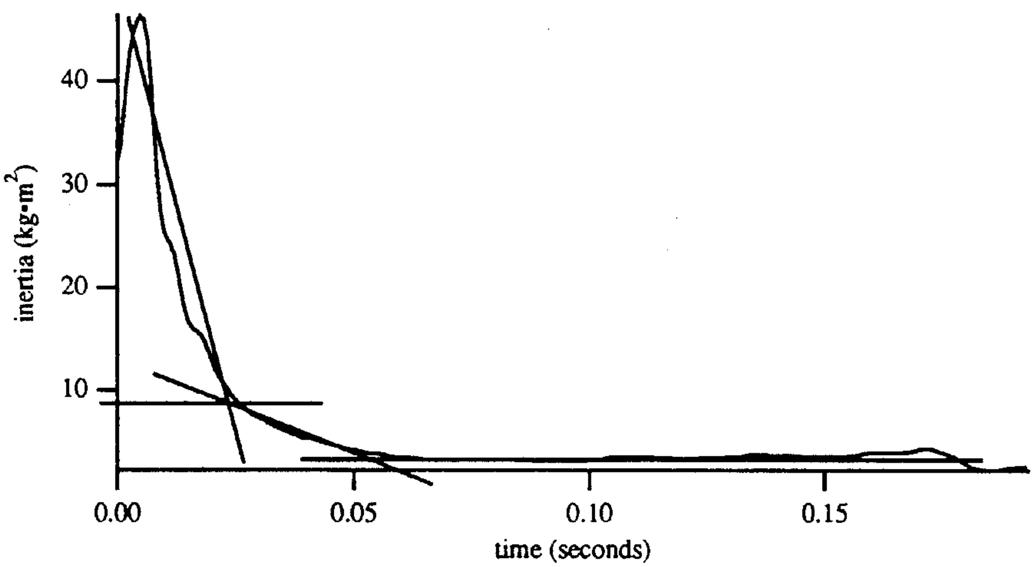

Figure 6: Output used for estimating the inertia showing the three calculated lines and their intersections.

All 120 cases were also estimated visually using a point information routine in Igor, a graphing and data analysis program (Igor, 1990). These results were close to those determined by the computer algorithm and the undetermined 22 cases were replaced with the corresponding visually determined data.

The calculated mass moment of inertia was computed using the cylinder method (Drillis and Contini 1966), anthropometric information from Drillis and Contini (1966) and Pheasant (1988) and anthropometric information taken from each subject. Adjustments were made to both data sets assuming that the rotations were occurring about the L5/S1 joint. The other assumption was the subject acted as a rigid body (i.e., all parts of the subject remained at identical positions from each other).

Single-joint Parameter Estimation

Single-joint parameter estimation was accomplished using MATLAB (1991) and these associated programs: 1) System Identification Toolbox, 2) Control System Toolbox, and 3) Signal Processing Toolbox. The pre-processed data was imported into MATLAB where additional filtering was performed on the data using a $5^{\underline{th}}$-order, bidirectional, Butterworth filter.

The system order was first estimated. A second order model was found usable (see Eq. (2)). Adding an order decreased the estimate's variance only slightly. Adding yet another order decreased the estimate's variance significantly. Therefore only the second and fourth order parametric models were estimated.

The least-squares approach, ARX in the System Identification Toolbox (Ljung, 1991), minimized the sum of the prediction errors. This method assumed the noise found in the system was white. An analysis of residuals showed this to be a good assumption.

RESULTS

Mass Moment of Inertia Estimation

The mass moment of inertia estimates were consistent. The rigid body assumption was violated because all of the body parts were not accelerated simultaneously. This violation was obvious enough to observe on the videotape record of the trials. The center of the subject's torso led while the shoulders and head lagged behind. This would imply modeling the system as a double inverted pendulum and is the reason the fourth-order linear modeling was performed. It may also be for this reason that the measured mass moments of inertia were expected to be smaller than the predicted mathematically. This expectation is affirmed in Fig. 7 which compares the mathematical calculation of the mass moment of inertia to the experimental measurements. There is an extra set of points on the graph. They are the higher group at $8.0 kg \cdot m^2$ and correspond to the results of a preliminary experiment where the preload was varied over 20 different trials. This was done using the same subject as for the regular testing and was used for verification purposes only. It was assumed that this did not effect the subject during the regular testing. These points were not included in the stated r^2 values. The difference might be intra-subject variability or the mathematical calculation may be sensitive to the level of the preload or it may just be noise.

As indicated by the r^2 values in Fig. 7, the results from the left and right instrumentation are good. The position instruments did not show such a deviation between sides. Possible instrumentation drift may account for discrepancies between the right and left instrumentation.

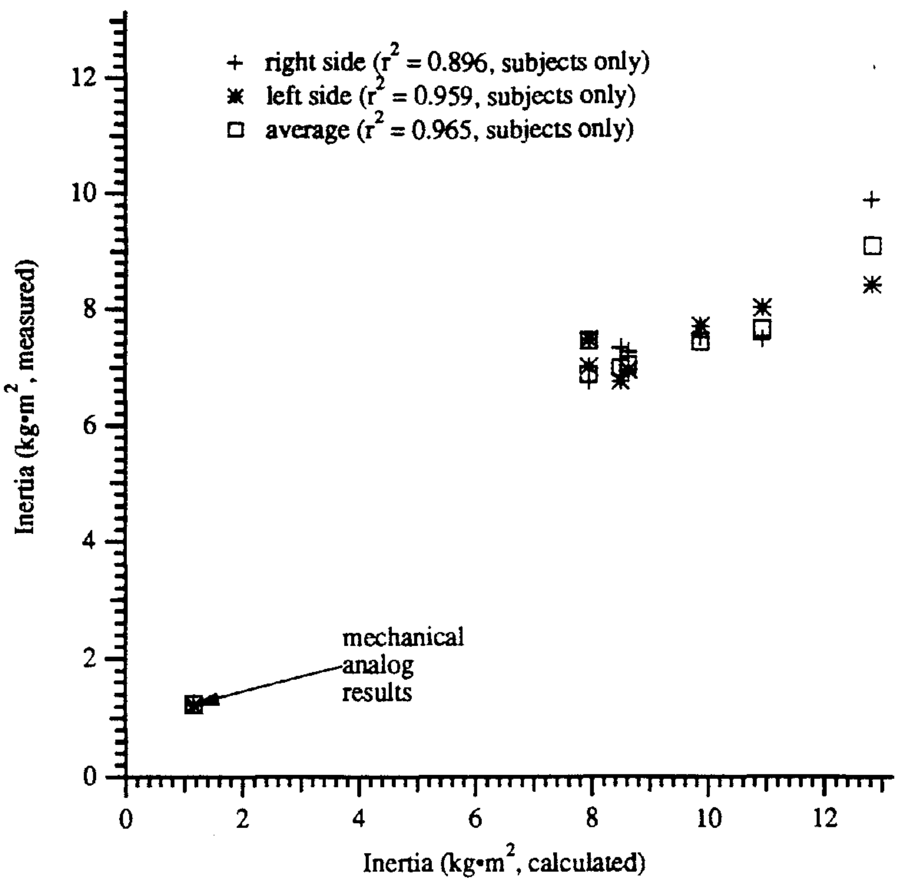

Figure 7: Inertia calculations vs. estimates (each point is an average of 20 trials except that the mechanical analog is averaged over 16 trials, also note the two points at approximately $8.0 kg \cdot m^2$ on the calculated axis, see text for explanation).

The averaged, measured, mass moment of inertia values were used throughout the rest of the analyses.

Second-order Modeling – Mechanical Analog

The parameter estimates of the mechanical analog system were quite good. Fig. 8 shows the average output trace and a simulated output trace using the parameters determined from a second order parametric estimation. The simulation's input was delayed 150 milliseconds. This delay was accounted for partially by the preload used. The preload was scaled to the mechanical analog's equivalent of the HAT weight. This preload level was not large enough to remove all of the slack from the pulse cable and there was a visually noticeable sag in the cable. It was felt that consistency in scaling was more important than the preload as the mechanical analog acted much more like a rigid body than did the subjects.

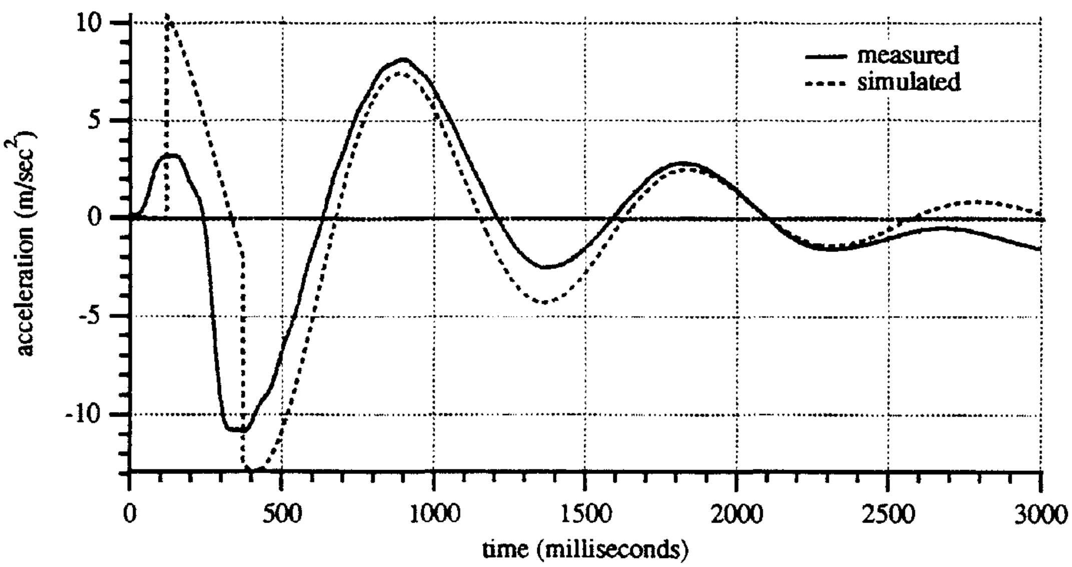

Figure 8: Analog results - measured and simulated.

The estimates for the mechanical analog compared favorably to those calculated. The estimate for ω_n was $6.6\frac{rad}{s}$ and for ζ was 0.172. The calculated values were $\omega_n = 7.5\frac{rad}{s}$ and the $\zeta = 0.2$. These were comparable to the the subject results of $\omega_n = 8.8155\frac{rad}{s}$ and the $\zeta = 0.1035$ (from Table 2, below) indicating that the mechanical analog was a good test even if it was scaled differently from the human subjects.

Simulating the output given the estimated second order parameters was difficult. The pulse delay combined with the violation of the rigid body assumption made for quite different traces. Once the delay was fixed the results were quite good (the average correlation coefficient was 0.9422 over the period from 500 to 3000 milliseconds). The simulation did not exactly match the actual output for several reasons: 1) the mechanical analog was really an inverted pendulum on a flexible rod resulting in linear estimations of a nonlinear system , 2) the analog's motion was not constrained to the sagittal plane only; it also had free movement in the frontal plane and 3) the the harness system and low preload made the rigid body assumption less valid. The high average correlation coefficient, however, does indicate that the ToMMS can identify (nearly) linear second order systems quite well.

Second-order Modeling – Subjects

Fig. 9 shows the average results, of the right and left acclerometers, aggregated across all subjects. The three major peaks are labeled as well as the zero crossing points used for the analyses.

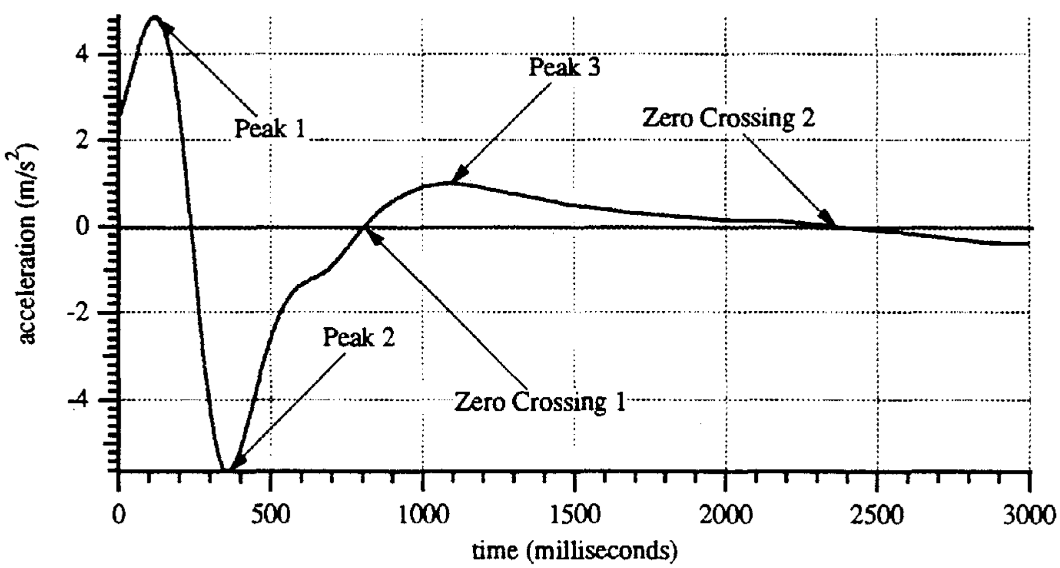

Figure 9: Acceleration outputs averaged over all subjects (note: time zero is where the input crossed a threshold of 10 N).

A linear, second-order model was fit to the data. The model differed slightly from the generic second order model, Eq. (2), in the Introduction. This was because the data utilized for the input, $X(s)$, was not position data, but acceleration data scaled to each subject's mass moment of inertia (Grabiner and Jeziorowski (1991) found that by scaling their data as a function of body mass or mass moment of inertia the r^2 results from their regressions increased by up to 50%). Substituting angular position, $\Theta(s)$, for the output, $Y(s)$, and torque, $T(s)$, for the input and then differentiating twice yielded,

$$\frac{\ddot{\Theta}(s)}{T(s)} = \frac{Gs^2}{s^2 + \frac{B}{I}s + \frac{K}{I}}. \tag{6}$$

Changing from rotational variables to translation variables (translational accelerometers), $T = Fr$ and $\ddot{\Theta} = \frac{a}{r}$ and substituting $G = \frac{r^2}{I}$ resulted in the estimated characteristic equantion,

$$\frac{a}{F(s)} = \frac{Gs^2}{s^2 + \frac{B}{I}s + \frac{K}{I}}. \tag{7}$$

Other methods (e.g., maximum likelihood estimation) were examined but the results were not consistent and a residual analysis indicated that the noise was indeed white thereby making the least-squares method the best estimator. The natural frequency and damping ratios were determined by utilizing the eigenvalues of the dominant poles from the z-transform (discrete-time) representations of these transfer functions (MATLAB, 1991). This is a different method than the method expressed in the introduction for continuous-time representations, but yields identical answers for nearly all conditions. The model was then increased to third and then fourth order. No significant predictive ability was gained with increasing model order. This reinforces Kearney and Hunter's (1990) claim that second-order approximations provide good descriptions of systems under minimally changing conditions.

Next, time-to-peak (PKTIM) and peak magnitudes (PKHGT) were calculated for the three major peaks of the response (see Fig. 9). The numbers following the variable name correspond to the peak locations in Fig. 9. Table 2 shows the grouped averages and their statistical significance across all subjects and trials of a full-factorial MANOVA ($SUBJ \times FOR \times FAT$, df. = 120) using SYSTAT (1992). Note that the SUBJ variable was significant at $p < 0.0001$, or better, under all conditions, except Zero Crossing 2 (ZCROSS2) which had no variable significant.

Table 2: Selected group averages and statistical results from the Eq. (7) estimations.

Parameter	Grand Average	fatigue level (FAT) (unfat.)	?	(fat.)	force level (FOR) (low)	?	(high)	Signif. Interaction
ω_n	8.8155	8.5010	*	9.1301	8.7167		8.9143	
ζ	0.1035	0.1102		0.0968	0.1011		0.1059	SUBJ×FAT†
PKTIM1	0.1189	0.1166	*	0.1211	0.1177		0.1200	SUBJ×FAT,* FAT†
PKHGT1	4.5220	4.2690	†	4.7751	3.6873	*	5.3567	
PKTIM2	0.3675	0.3654		0.3697	0.3654		0.3698	
PKHGT2	-5.2971	-5.4377		-5.1563	-4.6190	*	-5.9755	SUBJ×FAT,* SUBJ×FOR†
PKTIM3	1.0159	1.0067		1.0250	1.0165		1.0152	SUBJ×FAT*
PKHGT3	1.3771	1.6605	*	1.0937	1.3355		1.4187	SUBJ×FAT*
ZCROSS1	0.7167	0.6889	*	0.7445	0.7327		0.7007	SUBJ×FOR*
ZCROSS2	1.7864	1.7960		1.7767	1.8297		1.7429	

[ω_n ($\frac{rad}{s}$), ζ (dimensionless), PKTIM (seconds), PKHGT ($\frac{m}{s^2}$), ZCROSS (seconds)]

? – Significant between levels?
*denotes significance at the $p < 0.05$ level
†denotes near significance, between $0.05 \leq p \leq 0.06$

Fig. 10 and Fig. 11 show the different effects of these conditions across all subjects. The difference in force inputs (15 or 20% of bodyweight) was seen only in the magnitude of the subjects' responses (see Fig. 10). The high force input was 33% greater than the low force input ($\frac{20\% bodyweight}{15\% bodyweight}$). An interesting phenomena is shown by the high force response, on average, is 45.3% greater at Peak 1, 29.4% greater at Peak 2 and had declined to only

6.2% greater at Peak 3 (only the difference at Peak 3 was not significant, see Table 2). There were no differences in the arrival times of any of the peaks nor at Zero Crossing 1.

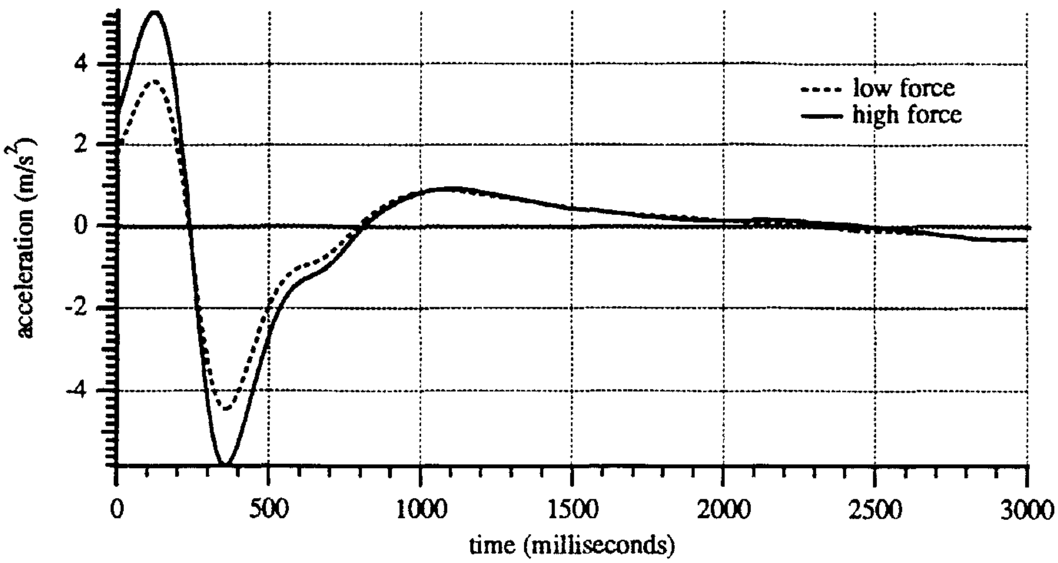

Figure 10: Acceleration outputs averaged as a function of force level over all subjects.

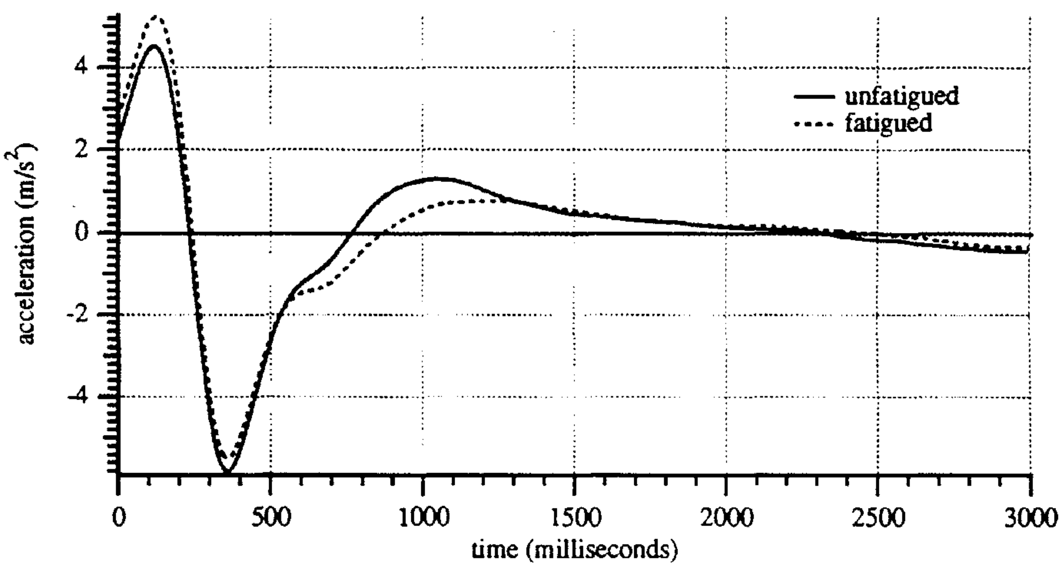

Figure 11: Acceleration outputs averaged as a function of fatigue state over all subjects.

The unfatigued vs. the fatigued plot shows different results (see Fig. 11 and Table 2). The estimate of ω_n and the timing of Zero Crossing 1 combined for a quicker response for the unfatigued case. The quicker response was not significant for the time-to-Peak 2 (PKTIM2), nor was the response magnitude significant at Peak 2 (PKHGT2). This may indicate that there is some "underlying" sudden response mechanism which is fatigue-resistant since that is the first parameter measured after the input pulse is removed.

Several studies (e.g., Lance and Chaffin, 1969) have indicated a reduction in the ζ as a function of increasing fatigue. Although it is not a significant difference, the trend is evident in Table 2 where, when grouped by fatigue, the average ζ for the unfatigued conditions was 0.1102 and decreased to 0.0968 under the fatigued condition.

Quasi-linear and several nonlinear simulations were performed. The large increases in model complexity did not yield any more useful information over the second order model.

CONCLUSIONS

Mass Moment of Inertia Estimation

The method used for mass moment of inertia prediction is good. However, it does need some refinement. It also needs a rational physiological explanation for choosing the first intersection point as the mass moment of inertia value. The method did underestimate the values computed from mathematical models, but this was expected because of the rigid body assumptions made in the mathematical models. It does provide repeatable information under these experimental conditions.

The ToMMS and associated software can accurately identify a given linear system. It is not known if it can recognize a known nonlinear system since no known nonlinear systems were tested.

Subject Differences

The hypothesis that it would be possible to select a homogeneous subject group using the criteria used in this experiment was not borne out by the results. Each subject's response was different ($p < 0.0001$ across all variables except Zero Crossing 2), but all experienced fatigue as measured in this study and as evidenced by the large number of FAT effects and $SUBJ \times FAT$ interactions (see Table 2) across nearly all measured variables.

Testing additional subjects in mild or moderate risk groups is probably possible without fear of injury. The subjective responses all indicated that the stress of the testing did not create any fear of injury. The fatigue conditions did not indicate loss of gross motor control and, therefore, would not expose the mild or moderate groups to a greater risk. The test is still not recommended for higher risk groups, such as those with low-back pain, those who exercise minimally or those with reduced or compromised cardiovascular or lung functions. More testing is necessary in the mild and moderate before any decision is made concerning those subjects at high risk.

Comparisons to Single-Joint Results

The results reported here, for the most part, agreed with those Kearney and Hunter (1990) report are consistent across parameter estimations of single-joints. The second-order linear models are powerful and do provide much information.

The damping ratio, ζ, remained constant throughout most conditions. The major difference was in the $SUBJ \times FAT$ interaction. The lowered ratio of 0.1 instead of 0.2-0.4 may be due to the nonlinearities observed in both the experiment and the results. More likely it is due to the torso needing to be more responsive to changing situations and the brain keeping it there. One method of producing faster responses accomplished by lowering the damping ratio (albeit at the expense of a longer transcient response). If the torso acts as some who have modeled it suggest (see Vukobratović and Stepaneneko, 1972, Chow and Jacobson, 1972, Hemami and Katbab, 1982), small perturbations about the upright position are damped out and the torso returns to the original position without additional muscular input. Since the torso is such an important component in posture, both in mass and position, this second idea seems the better explanation.

The transient response scaled to the gain of the input. However, the scaling was nonlinear as the gain at each peak became smaller while all returned to the original position at the same time. Perhaps there is a quasi-linear response as a function of time operating on the gain of the resonse, such as the saturation of muscle force which is indicative of fatigue's presence.

The response duration did not matching natural frequency changes as predicted. Several reasons for this are put forth. The duration effect was only prevalent in certain parts of the response, such as the slower time to Zero Crossing 1. Here there was an effect as a function of the response timing. The standard deviation may have blurred this effect as the standard deviation increased as a function of the time when the variable was measured. The largest was seen in the last variable, Zero Crossing 2. The nonlinear gain scaling may also have contributed. Perhaps the differences in ω_n were too small for the effect to manifest itself or that the nonlinear effects predominate in the response.

Comparisons to Posture Results

Johansson, *et al.* (1988) used a "swiftness" parameter which closely resembles ω_n. Using the "swiftness" parameter results for a standing subject with a rigid body mass moment of inertia, the responses are slower (range $2.90 - 6.89\frac{rad}{s}$ compared to $8.8155\frac{rad}{s}$ found in this experiment). This is expected as ω_n should decrease with increasing mass moment of inertias. The damping ratios (estimated from their results and assuming that "swiftness" was a good approximation to ω_n) are higher in their study (0.3-0.5 compared with 0.0968-0.1102 from Table 2) and closer to the range expected by Kearney and Hunter (1990).

Fatigue Effects

The modified Sorenson test was a sufficient measure of fatigue as fatigue was defined for this paper. It predicted fatigue effects and the results indicated that this was true.

Loss of stability or control of the torso was not seen as a function of fatigue, but delays were seen in the response. Delays increase the tendency of a system towards instability. This could indicate that the subject, if performing under similar conditions, may be at a higher risk of injury. Most of our actions are preprogrammed motor responses that are altered slightly to fit the current conditions (e.g. see van Galen and Wing, 1984 or Schmidt, 1982). Performing under different parameters during a fatigued response (changing B and K with delays added to the feedback) could adversely affect the ability to correctly control that motion. Most actions are practiced many times under unfatigued conditions so that not only is the motion compromised by its inherent parameters, it is being performed without the benefit of practice under those conditions. Under these conditions the likelihood of adverse effects increases.

There may be an underlying fatigue resistant response to a sudden load. The time to peak and the magnitude of the response at the second peak not being significant indicates that a process is operating which is independent of fatigue (except as a function of force, where it was expected to be different due to the input differences). The best hypothesis is that these variables are a function of the passive tissues. Passive tissue does not rely upon metabolism over such a short period and its response should not have been affected by the type of fatigue induced in this experiment. Another hypothesis is that, perhaps, the nervous system automatically compensates for the fatigue and requests a larger response from the muscles. An interesting experiment would be, after fatiguing the subject, to repeatedly present them with the pulse trial and look for differences or variations over time. These results should answer most of the question of whether or not that response is a function of the passive tissues.

Finally the effect of fatigue is not truly quantifiable. Fatigue was a blocking variable so that differences attributed to fatigue may be due to the ordering of the fatigue trials last. More data is necessary before forming hard conclusions.

Model Order and Type

No significant payoff for using higher order models for estimation was seen in these results. In fact, the estimations began to become numerically ill-conditioned when the order was increased to five. That also may just be an artifact of this experimental methodology. These results indicate that a double-pendulum model may not be appropriate under these conditions. It would be interesting to measure the differential accelerations between the head and the torso or to more rigidly restrain the head and neck via a more restrictive cervical collar to determine if more complex linear models should be used.

The most difficult analysis was modeling the change about 500 milliseconds into the response (see Fig. 9). The linear estimations could not account for the change. Adding quasi-linear elements to the model did not increase its predictive ability nor could they account accurately for the change. The hypothesis is that the change is regulated by higher-order feedback because all of the muscle reflexes (or lower-order feedback) should have been completed by that time. The quasilinear modeling results do indicate that the stiffness decreases and the damping increases. This makes sense assuming that the subject's goal is returning to the upright position in the minimum time. The relatively long input time ($250 milliseconds$) allows the torso to become very stiff. The torso must then damp out the input energy quickly. Interesting experiments could be performed for better determining which sections of the response were by higher or lower feedback centers. Perhaps a more thorough nonlinear modeling effort resulting in better simulations and a better understanding of what is happening approximately $250 milliseconds$ after the input is removed could be made.

Future Research

A detailed sensitivity analyses should be performed on such variables as the preload, pulse width and duration, subject anthropometry and other frequencies of input. These results are valid under just a few experimental conditions and it is not known how these would compare under different conditions. Much more information about the human torso's response to a sudden load would also be gathered.

EMG responses were not included in this paper as they are the subject of another paper in production at the time of this publication. But information about the timing response, rate and resulting force level for the muscles other than these measured are necessary when considering non-sagittaly-symmetric motions in the future.

The nonlinear aspects and modeling of these results should be considered in much greater detail. As mentioned above, further experiments should be designed and performed with the intention to isolate where these changes are occurring. Those should yield some clues as to why and how these differences are occurring.

More research is needed to back up the claim made earlier that the likelihood of an injury increases as a function of fatigue. Studies where performance variables are tracked over time and then correlated to some fatigue metric have not been found.

ACKNOWLEDGEMENTS

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Sagittal plane dynamics of the spine *in vivo* Part II: Muscle contraction delays and time constant changes.

by

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ABSTRACT

Electromyographic (EMG) and muscle force activity were examined for timing and levels of co-contration utilizing a sudden loading/short-duration, high-intensity fatigue paradigm. The EMG patterns indicate the torso follows a suboptimal minimum time control pattern. They also show a high degree of co-contraction between anterior and posterior muscles lasting throughout the subject's response. The output force was adjusted for fatigue (Park and Meek, 1992 in press) and again showed that the sudden loading paradigm requires a high degree of co-contration. Six subjects were restrained such that sagittal plane motion was limited to that above the L5/S1 joint. Two square pulse magnitudes (15% and 20% of body weight) were applied to the center of mass of the head, arm, torso (HAT) over a 250 millisecond period during and after which the resultant motions were recorded (for a total of 4000 milliseconds). Five trials of each magnitude, randomly determined and randomly mixed together, were given preceeding a set of torso muscle fatiguing exercises. After the fatiguing exercises another set of of randomly determined and randomly mixed together pulses were presented for a total of 20 trials per subject.

Biomechanical consequences of sudden loading and fatigue on the spine *in vivo*.

by

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ABSTRACT

A sudden loading/short-duration, high-intensity fatigue paradigm was utilized for exmaining dynamic and electromyographic (EMG) parameters of the human torso. The study of the dynamic parameters shows that the compression levels at the L5/S1 joint are significantly higher than those predicted from static or quasi-static models under similar conditions. The results of the EMG study show a high level of cocontraction between the anterior and posterior muscles. This cocontraction adds to the already high compression levels seen just from the dyamic loading. Taken together a "floor" or minimum level of spinal compression was computed. This "floor" was larger under the higher force and the fatigue conditions. The resulting "floor" indicates that sudden loading of the human spine should be avoided or reduced in the workplace because of a higher risk of possible low back injury.