ADAPTATION TO TRANSIENT POSTURAL PERTURBATIONS

FINAL REPORT

NASA GRANT NAG 9-291

October, 1992

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Submitted by:

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I. INTRODUCTION

This research was first proposed in May, 1986, to focus on some of the problems encountered in the analysis of postural responses gathered from crewmembers. The ultimate driving force behind this line of research was the desire to treat, predict, or explain "Space Adaptation Syndrome" (SAS) and hence circumvent any adverse effects of space motion sickness on crewmember performance. The aim of this project was to develop an easily implemented analysis of the transient responses to platform translation that can be elicited with a protocol designed to force sensorimotor reorganization, utilizing statistically reliable criterion measures.

This report will present: 1) a summary of the activity that took place in each of the three funded years of the project, 2) discussion of experimental results and their implications for future research, and 3) a list of presentations and publications resulting from this project. Most of the figures and the two tables referred to in the text are placed at the end of the body of the report, before the appendices.

II. SUMMARY OF YEARLY ACTIVITY ON THE GRANT

Reports were submitted to NASA at the end of the first and the second year of the grant period. The original award took place in April, 1988, so the first year period covered 4/1/88 - 3/31/89. The first and second year information presented here comes directly from the reports submitted:

First Year

GRANT ADMINISTRATION

The starting date for the first year was April 1, 1988. Other dates were noteworthy: the NASA postural measurement system was shipped to UMASS and arrived August 16, 1988, but the LSI-11 system had no terminal for user interaction. NASA subsequently shipped a VDT system with keyboard, which arrived September 16, 1988. The other event of note was the execution of a no-cost extension from April 1, 1989 to June 30, 1989. We received approval from our institutional Human Subjects Review Board to perform the experiments in August, 1986, and this approval was renewed for 1988-89.

ACTIVITIES AND ACCOMPLISHMENTS

Hardware

The activities during the first year of the grant concentrated on efforts to refurbish the LSI-11 based NASA-JSC postural measurement system. These efforts are chronicled first.

Upon arrival of the VDT and terminal in 9/88, the system was powered-up for the first time. Within fifteen minutes from initial power-up, the dual floppy disk (RX01) system spontaneously went into a rapid series of seeks, and proceeded to crash. The controller card was visually inspected and re-seated, but the floppy drive would not respond. The entire system was rendered useless because of the reliance on the disk drives for data acquisition and stimulus delivery software. After much digging, two discarded LSI-11 computers were discovered in moth-balls in the Electrical Engineering Department at UMASS. Through December, 1988 and January, 1989 the components from the three LSI-11 systems were mixed and matched until a final hybrid evolved. The hybrid system included double sided floppy drives (RX02), an LSI-11/23 CPU, and bypassed the SMU controller from the original system (which was suspect). This hybrid system was used to move the posture platform and collect data from the digital line scan cameras. Two problems still remained: 1) we had no means to inspect the data we acquired because the software previously written for the system relied on Tektronix graphics capabilities, and our VDT had no graphics capabilities, and 2) we had no system or applications software to modify any of the existing software for our purposes. These problems were solved by unearthing RT-11 v. 5.02 and FORTRAN 77 from various sources on campus, in addition to finding emulator software for Tektronix graphics terminals to run on IBM compatible PC's. By February, 1989, we could finally run the NASA/JSC postural measurement system and view the collected data. At that point in time, we felt that with some minor alterations to the software to customize our protocols, we were ready to start the first phase of data collection. Unfortunately, more serious snags developed.

In the process of running pilot sessions on lab personnel, the hardware/software system experienced intermittent failures which took the form of either losing control of the moving platform in the midst of a trial, or experiencing a hang-up in the software which required system reboot. The former occurrence occasionally resulted in high speed platform motions to the microswitched cut-off of the servomotor, an event of extreme danger to both the subject and the drive screw mechanism. As with the case of software hang-up, all previous data was lost and the session had to be restarted. Under many environmental conditions, using protocols of varied length (number of trials) and waveforms (sinusoidal, transient, and pseudorandom), between the two types of failures approximately 50% of the data acquisition was flawed.

During this time period (1/89-3/89) we were planning the modifications needed to the LSI-11 data acquisition programs for our needs. The major modification was the need to acquire the EMG data at 1000 Hz for four channels concurrently. The original software acquired the data at 25 Hz, much too slow for our purposes. The limitations of the double-buffering scheme for writing data to floppy disk combined with the limited RAM of the LSI-11 (128K) made the 1000 Hz unattainable. With this realization, we decided to use the LSI-11 system for stimulus delivery and digital line scan camera data acquisition only, and to acquire the EMG data and platform position data with an IBM compatible where the eventual analysis would be performed. We wanted to use the digital line scan cameras because they provide a real-time sagittal plane body silhouette position; unfortunately, they are mated to the LSI-11 bus and can not be interfaced directly to the PC without new interface cards. We were developing the software on both the LSI-11 and the PC to establish this stimulus delivery and data acquisition scheme, when the LSI-11 refused to boot up (6/89). Between the previous intermittent problems and the final difficulty booting up the system, I decided to abandon the LSI-11 based approach to our experiments at the end of the first year. Luckily, we had been developing software for the PC to analyze our data, so that progress was made on that front.

The other hardware related activity scheduled for the first year was the specification and purchase of an additional DC torque motor to add an ankle rotation degree of freedom to the existing translating platform. The calculations of required peak torque indicated that an Industrial Drives TT-2033-1010-A motor that produces 100 oz-in. of torque with a maximum speed of 6000 RPM will be sufficient for our purposes. The motor costs $517, and will be purchased at the start of the second year.

Software

This next section describes the software developments that took place throughout the first year, and those that took place at the end of the first year when the LSI-11 was abandoned. A data analysis package was developed to process the EMG signals. This package will accept up to 16 channels of analog EMG data (either raw or RMS), perform post-processing if necessary (integrate with a given time constant over a given window), and display the data (either one channel or multiple channels simultaneously). At this point the latency from the onset of platform motion can be manually digitized under cursor control or automatically selected and displayed by an algorithm adapted from previous work in our lab. The onset and duration of muscle activity are determined this way and stored to disk for each EMG channel. This software will operate on two different A/D boards: Metrabyte DASH-16 or Data Translation DT-2801.

Without the LSI-11 system to provide the velocity commands to the moving platform, the PC was needed for stimulus generation. The DASH-16 board has two channels of D/A, so software has been developed to output desired velocity commands to the platform. Output from the D/A is monopolar, so a bias voltage is used downstream to provide bidirectional control. Unfortunately, we discovered that the maximum sample rate to attain A/D data concurrently with D/A output under interrupt control was approximately 200 Hz, even running QuickBasic 4.0. After an extensive search of available products (at a reasonable cost) that would allow us to run concurrent A/D and D/A under interrupt control, none were found. Our solution to this dilemma is summarized in the section on our finalized data acquisition plans.

When the use of the digital line scan cameras was ruled out, we began to develop the software to perform batch analysis of body sway motion obtained with a video based system (MotionAnalysis System, Santa Rosa, CA). Our system is a 2-D video processor hosted by a SUN 3-160 Workstation, and uses two 60 Hz video cameras. Programs have been developed in the ExpertVision environment to provide sagittal plane coordinates of the fifth metatarsal head, the calcaneus, the lateral malleolus, the knee, the hip, the shoulder, the mastoid process, and the elbow. Preliminary experiments indicate a resolution of 0.1 cm.

One advantage of this increase in the amount of body position data (from two points with the digital line scan cameras to eight points with the video) is that the data will be compatible with a dynamic biomechanical model developed over the past year to predict joint reactive moments and L5/S1 loading. This will enhance our data analysis and interpretation.

FINALIZED DATA ACQUISITION PLAN

The decisions about the final data acquisition system have been made and will be implemented in the second year. We will use two 60 Hz video cameras and our video-processing system to acquire the body sway data. The laboratory IBM-XT will provide the velocity commands to the moving platform with the DASH-16 D/A channel output. The Zenith 386 system purchased to analyze the data will have a DT-2801 A/D board installed, and will then acquire the platform position and the EMG data at 1200 Hz. The IBM-XT will also output synchronizing signals to both the video processor and the Z-386 to allow data synchronization.

Subjects for the first phase of the experiments are being recruited from two large undergraduate Exercise Science courses, with the understanding that we are looking for a two year availability if possible.

REVISED TIME-LINE FOR SECOND YEAR ACTIVITIES

Given the delays in establishing a viable data acquisition and stimulus delivery system, the second year will be more intensive for actual experimentation. Phase I experiments with males will take place in November and December, 1989. Phase II experiments with males will be conducted in February and March, with the intervening time used to analyze Phase I results and to finalize the software for Phase II data acquisition. Meanwhile, the females for Phase I experiments will be selected in March, and Phase I experiments with females will begin at the end of March (after spring break). In April the software for running the Phase III experiments will be completed so that Phase III male data can be acquired in May, before the semester ends. June of 1990 will be spent analyzing acquired data and preparing the second year report. Manuscripts concerning results from Phase I and Phase II will be in preparation at this point.

BUDGET ACTIVITY IN THE FIRST YEAR

The budget for the first year went according to plan, with some slight stretching due to the delay in starting the second year. The equipment budget still contains about $2600, but this will be spent early in the second year to purchase the second axis DC torque motor (approximately $600) and the A/D board and screw terminal for the Zenith 386 (approximately $1500). Many supplies purchased during the first year will be carried over into the second year (particularly the EMG electrodes and the computer storage media), so that the heavy experimental activity of the second year should still remain within budget.

SUMMARY

First year activity concentrated on the development of a reliable stimulus delivery and data acquisition system. With the loan of the NASA/JSC postural measurement system (providing the recent model moving platform and the servoamplifier to add a second degree of freedom) and existing laboratory systems (the video-based MotionAnalysis System, the EMG measurement systems, and the IBM-XT with the DASH-16 A/D and D/A board), a two degree of freedom, dynamic postural measurement system has been assembled for essentially the cost of a Zenith 386, a DT-2801 A/D board, and a second DC torque motor. Actual hardware and software costs to assemble such a system from scratch would easily exceed $200,000; the system now is truly state of the art. Second year activity will take advantage of this system to run the experiments which we hope to use to design a dynamic posture testing protocol to systematically probe the sensorimotor adaptations to postural perturbations.

Year 2

GRANT ADMINISTRATION

The starting date for the second year was July 1, 1989. We received renewed approval from our institutional Human Subjects Review Board to perform the experiments.

ACTIVITIES AND ACCOMPLISHMENTS

Hardware

The data acquisition system described in the First Annual Report was fully implemented in the second year. The block diagram in Figure 1 presents the final configuration. PC-1 in this figure is the IBM-XT with the Metrabyte DASH-16 A/D and D/A board. D/A channel 1 drives the platform servo-amplifier, while D/A channel 2 sends a trigger signal to both the data acquisition board of PC-2 and to each of the VCR's (on audio channel 2, at 400 Hz). PC-2 is the Zenith 386 with the Data Translations DT-2801A A/D board. Five channels of data are acquired at 600 Hz for 4 seconds: platform potentiometer, and RMS-EMG's from ankle flexors and extensors and knee flexors and extensors. Data are saved in packed binary format to conserve disk space. Video records are transferred from videotape to digitized marker trajectories by the MotionAnalysis System (Santa Rosa, CA) VP-110 video processor working with a SUN 3-160 based workstation and Expert Vision software. These sagittal plane marker trajectories are then downloaded to a PC as ASCII files to allow further analysis and to be synchronized with the muscle activity patterns.

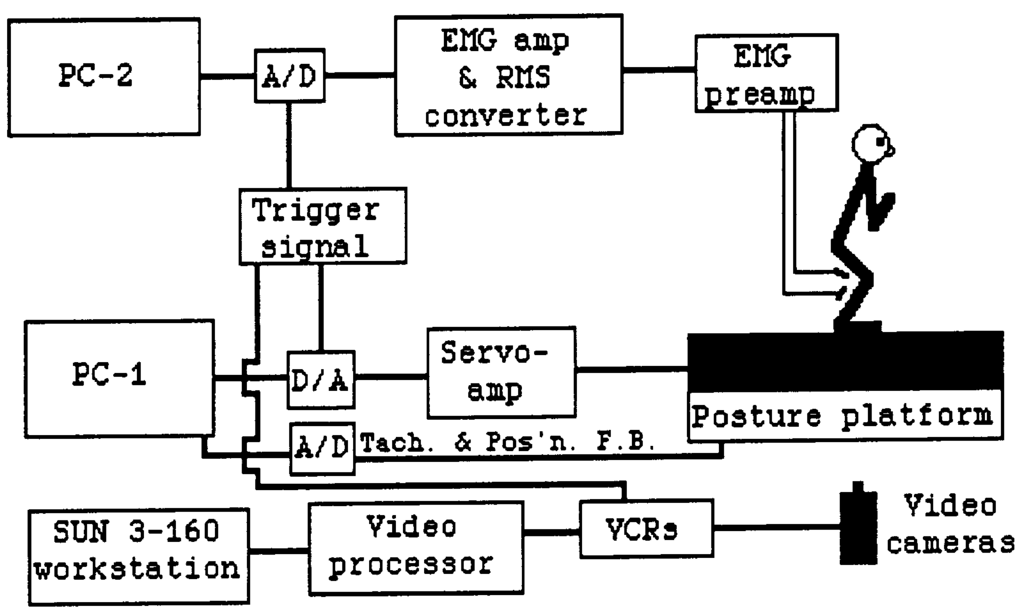


Figure 1: Block diagram of data acquisition system.

Software

A data analysis package was developed to process the EMG signals. This package will accept up to 16 channels of analog EMG data (either raw or RMS), perform post-processing if necessary (integrate with a given time constant over a given window), and display the data (either one channel or multiple channels simultaneously). At this point the latency from the onset of platform motion can be manually digitized under cursor control or automatically selected and displayed by an algorithm adapted from previous work in our lab. The onset and duration of muscle activity are determined this way and stored to disk for each EMG channel. This software will operate on data collected by two different A/D boards: Metrabyte DASH-16 or Data Translation DT-2801. Another analysis approach that has been implemented is the calculation of means and standard deviations of RMS-EMG's within particular time windows. The first experiment used three time windows: before platform motion, during platform motion, and after platform motion.

Experiment 1

The first experiment was completed in the second year. This experiment represented a combination of Phase I testing for both males and females. Sixteen subjects (8 male, 8 female) participated in the experiment which consisted of three days of testing. Subjects were translated backwards at a peak velocity of 25 cm/sec for either 0.3 s or for 1.5 s. Ten consecutive trials were acquired for each duration. Subjects also stood either on the level platform or on a slant board tilted backwards by 10 degrees. Therefore each session consisted of 40 dynamic trials. The first day was for acclimation purposes; the order of tilt and duration presentation for each of the two remaining sessions was balanced for order across the subjects. All trials were performed with eyes closed, arms folded gently across the chest, and knees straight but not locked.

Analysis of the EMG data is almost complete, but the video data is still being processed. Once ASCII files of all the marker trajectories are complete, this sway data will be analyzed for settling time (at the shoulder) and will also be used as input to a biomechanical model which predicts joint reactive forces and moments, particularly at L5/S1.

REVISED TIME-LINE FOR THIRD YEAR ACTIVITIES

The second experiment will require rearward rotation of the standing human, so the first activity will be to add this capability to the platform. The servo-motor will be purchased and mechanically coupled to the platform (the motor has been selected, as described in the First Annual Report). The second experiment will be completed by December, 1990, at which time the results of the first experiment will be in manuscript form. The final 6 months of the third year will be spent developing the criteria for an adaptive protocol based on Experiments 1 and 2, with Experiment 3 demonstrating the feasibility of the approach. At the end of the year specific recommendations for implementation of the protocol pre- and post-flight will be made.

BUDGET ACTIVITY IN THE SECOND YEAR

The budget for the second year went according to plan, with no major deviations from the proposal to report.

SUMMARY

Second year activity on the grant concentrated on the actual implementation of the reconfigured data acquisition system, culminating in the completion of the first major experiment aimed at developing an adaptive protocol for astronaut screening. Third year activities will complete the stimulus delivery system by adding rotation to the platform, and then complete the experiments needed to recommend a new protocol which will assess the standing human's capability to reorganize their sensorimotor behavior.

Year 3

GRANT ADMINISTRATION

The activity summarized here under Year 3 actually covers the final year of grant funding and a one year no-cost extension on the project. One item of note was that not all of the money was spent from the budget; money was left in to support the construction of the rotating addition to the translating platform because of a lack of access to facilities that could machine metal to the required tolerances within our department. To have the work done in another department, an excessive delay was unavoidable because of staffing cutbacks at UMASS and departmental priorities. Hence our efforts were concentrated on the analysis and interpretation of the data we had acquired with the translating platform.

Figure 2: CV's for a 0.3 s translation with no tilt

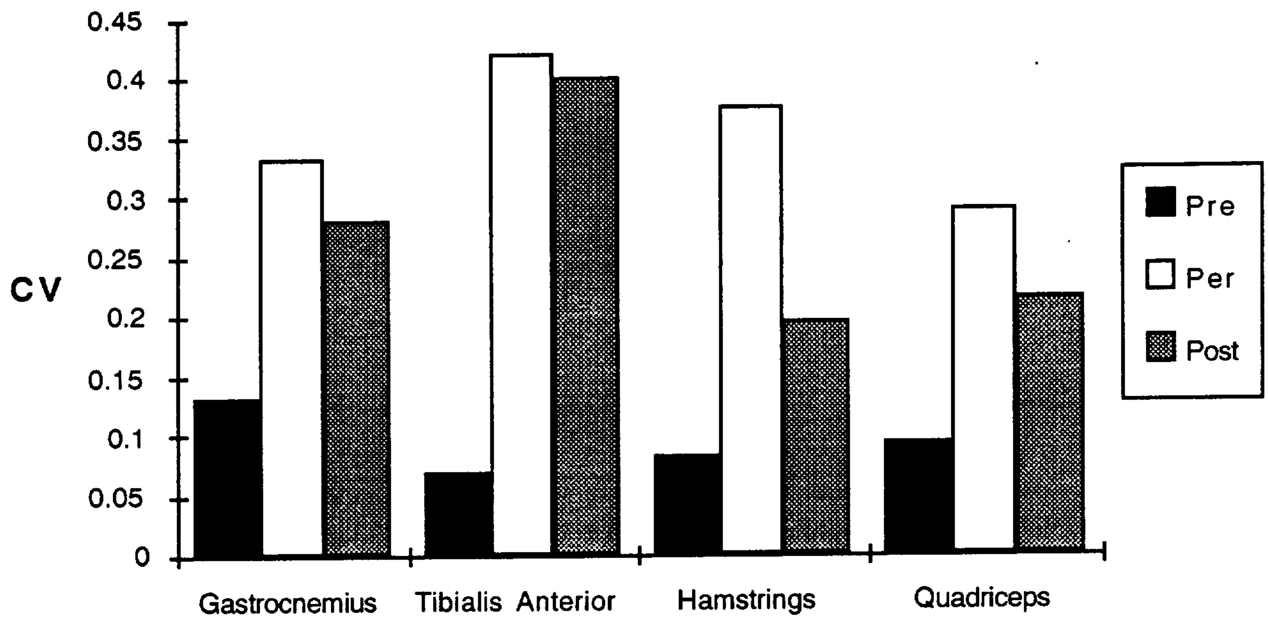


Figure 3: CV's for a 0.3 s translation with rearward tilt

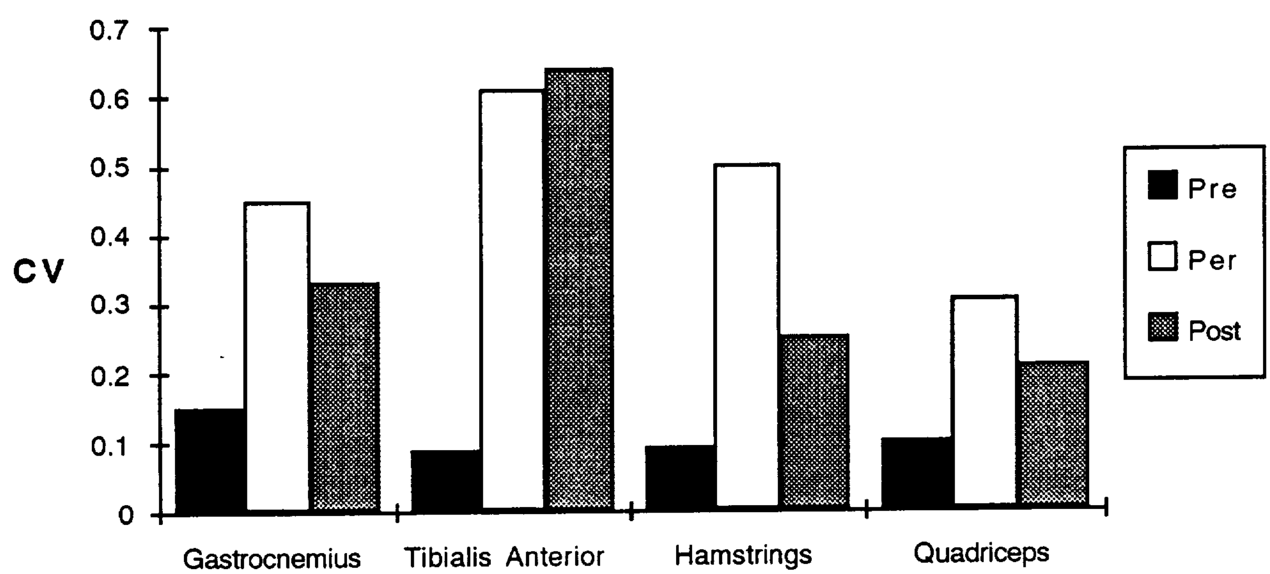


Figure 4: CV's for a 1.5 s translation with no tilt

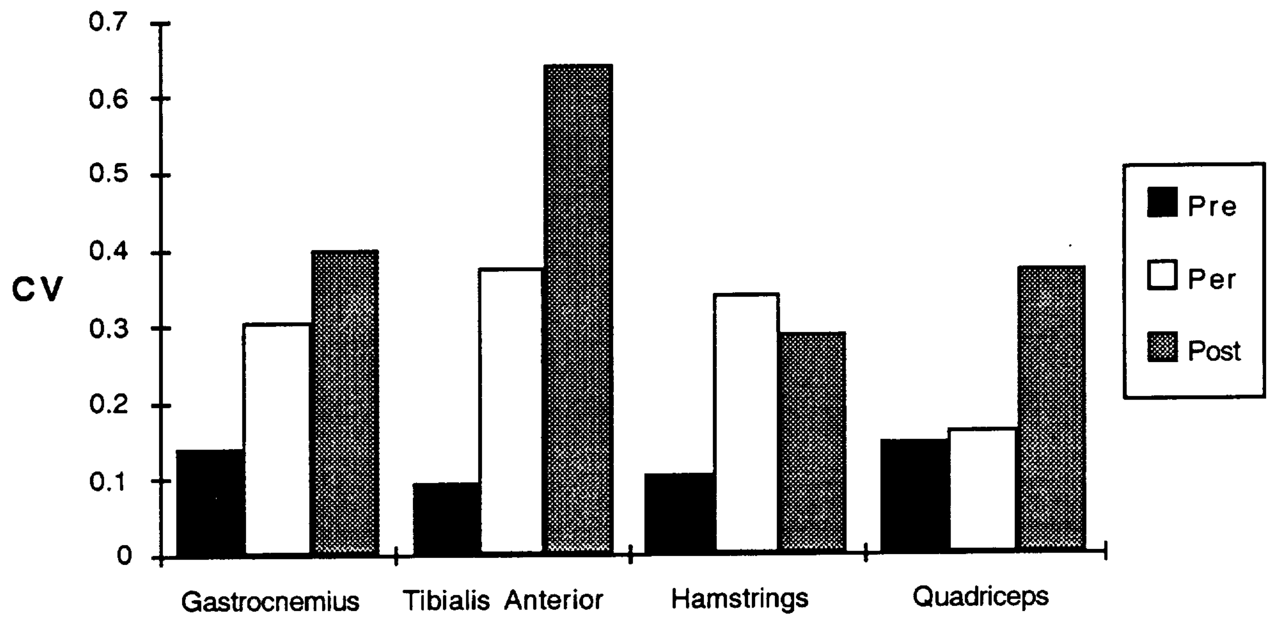


Figure 5: CV's for a 1.5 s translation with rearward tilt

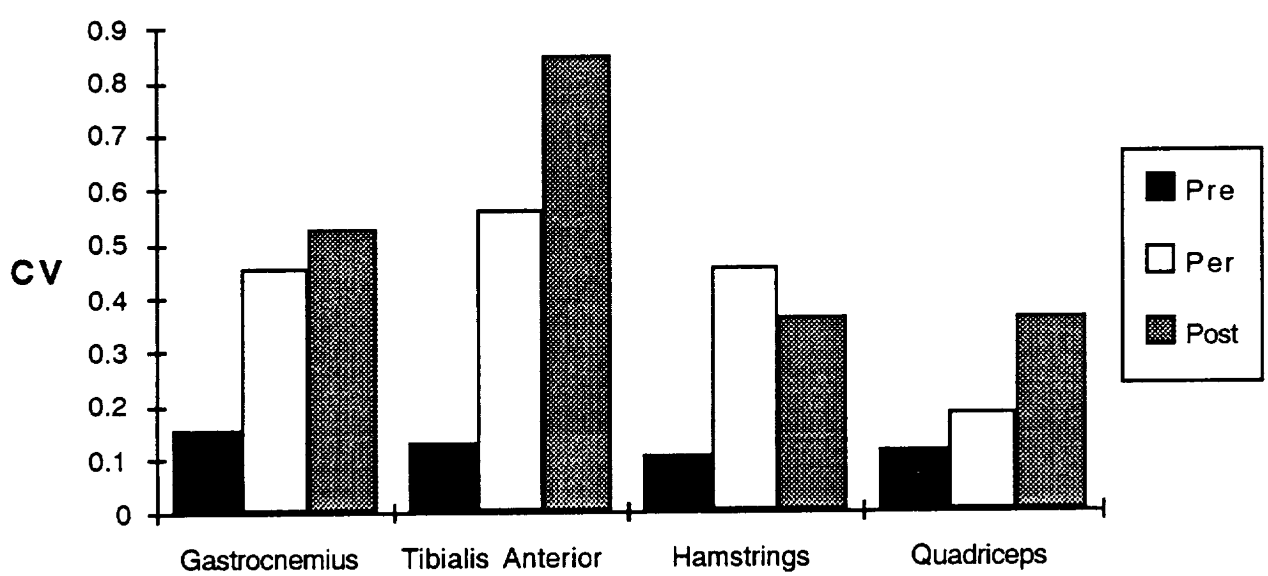


Figure 6: Average RMS-EMG's for a 0.3 s translation with no tilt

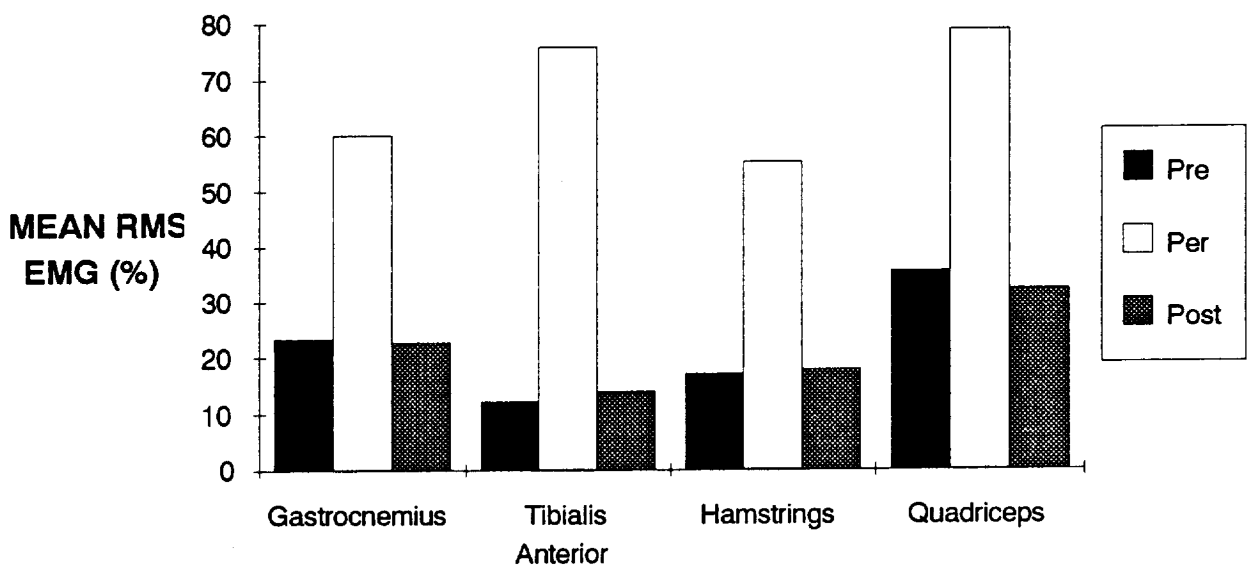


Figure 7: Average RMS-EMG's for a 0.3 s translation with rearward tilt

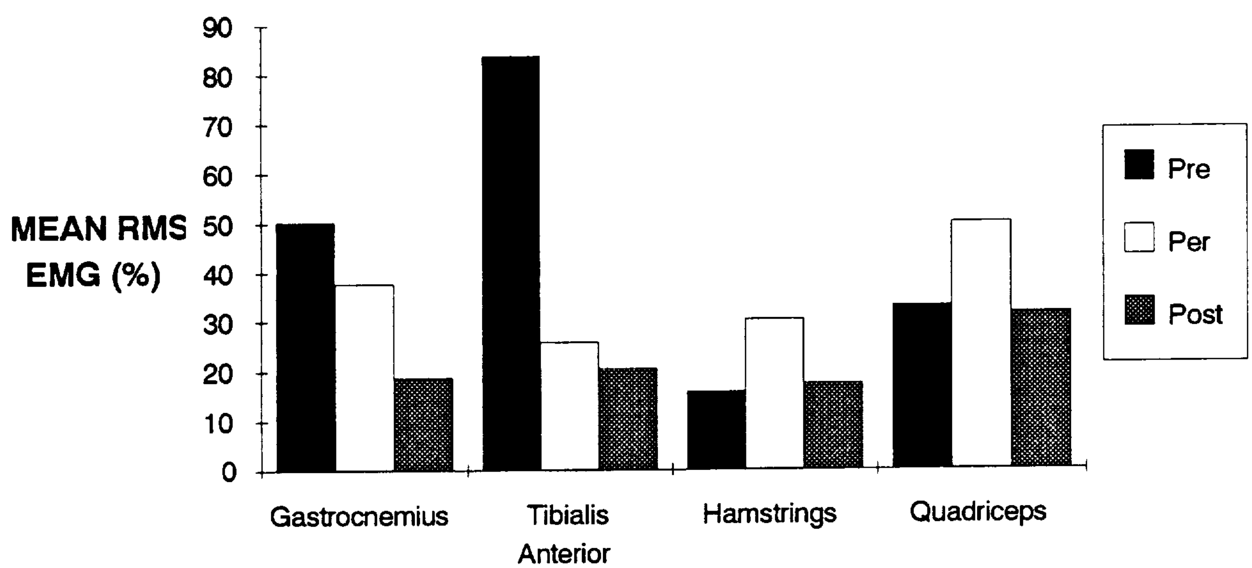


Figure 8: Average RMS-EMG's for a 1.5 s translation with no tilt

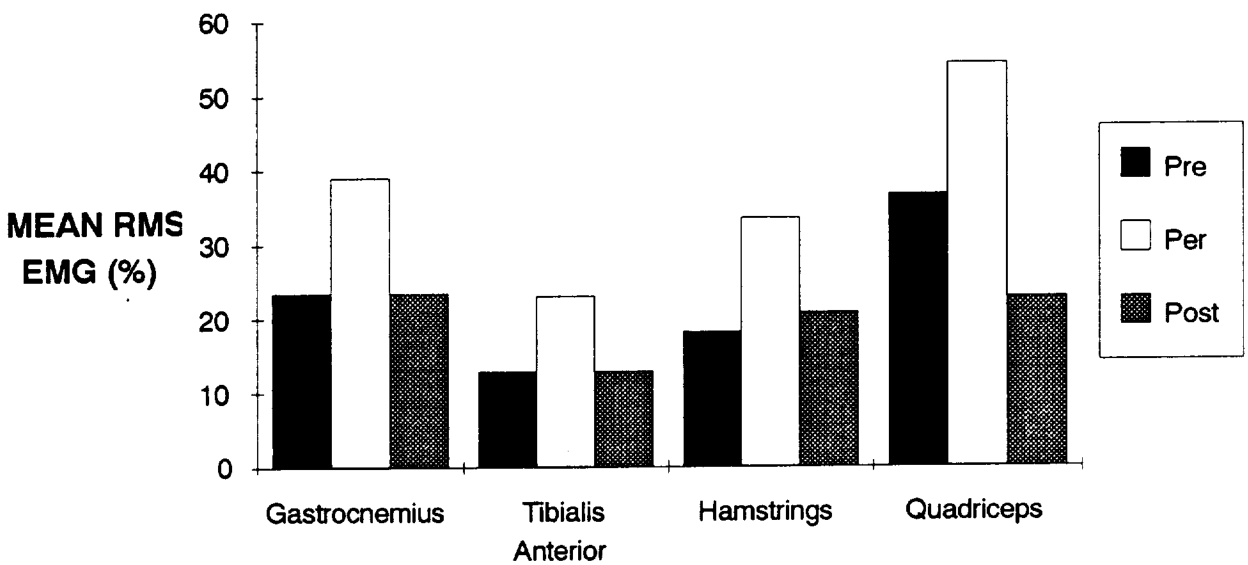


Figure 9: Average RMS-EMG's for a 1.5 s translation with rearward tilt

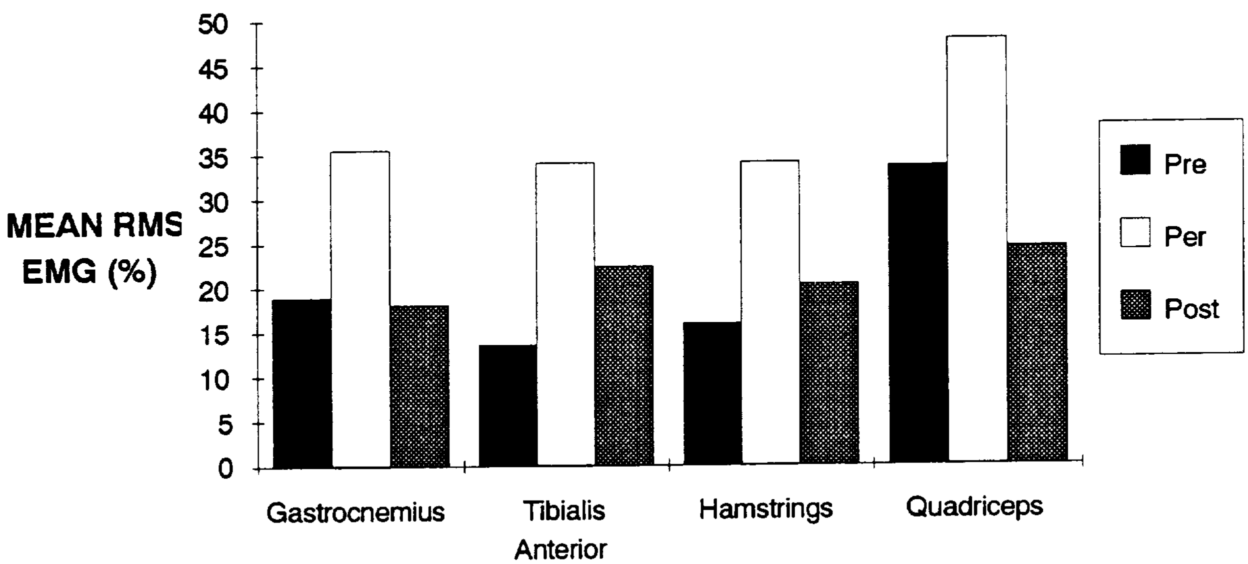


Figure 10: Pre-translation CV's

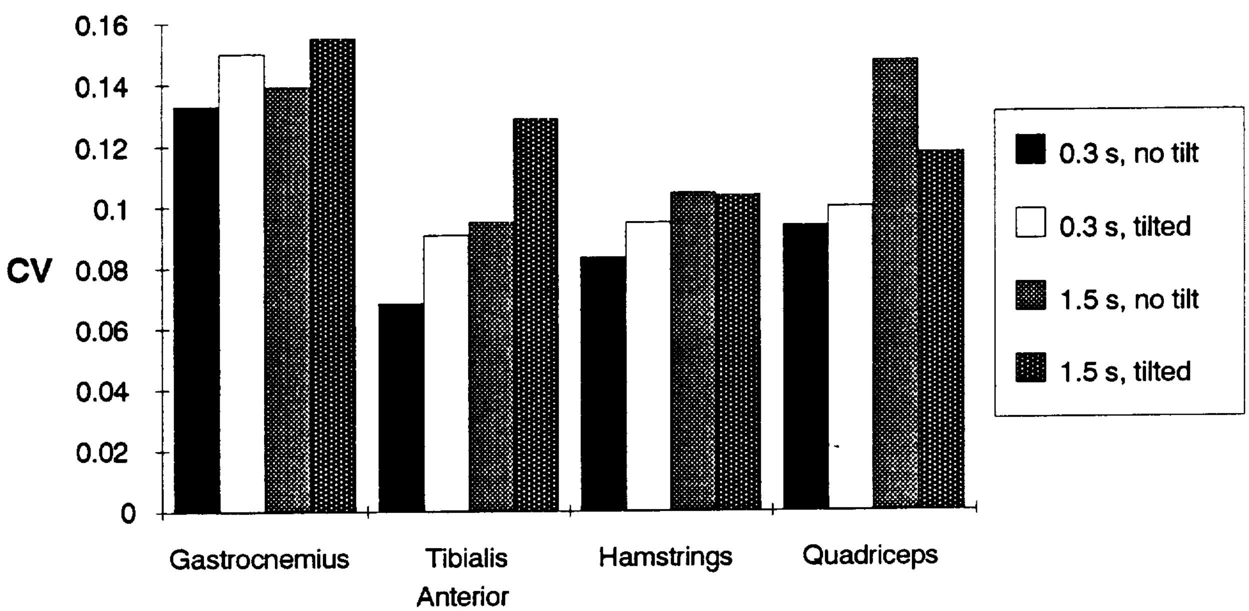


Figure 11: Per-translation CV's

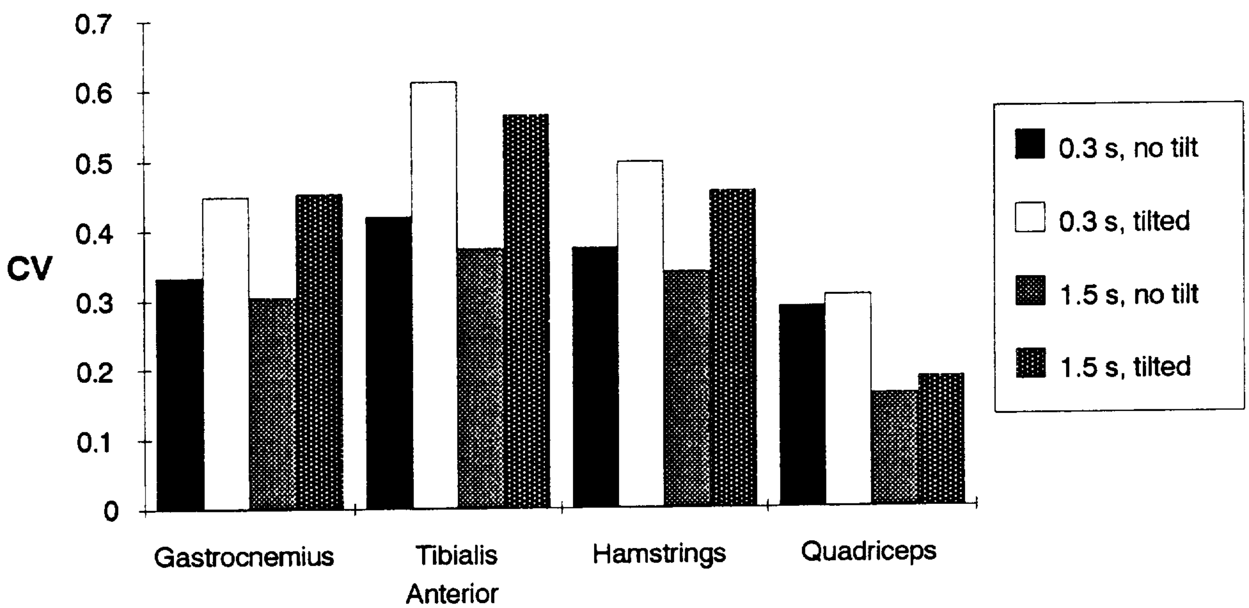


Figure 12: Post-translation CV's

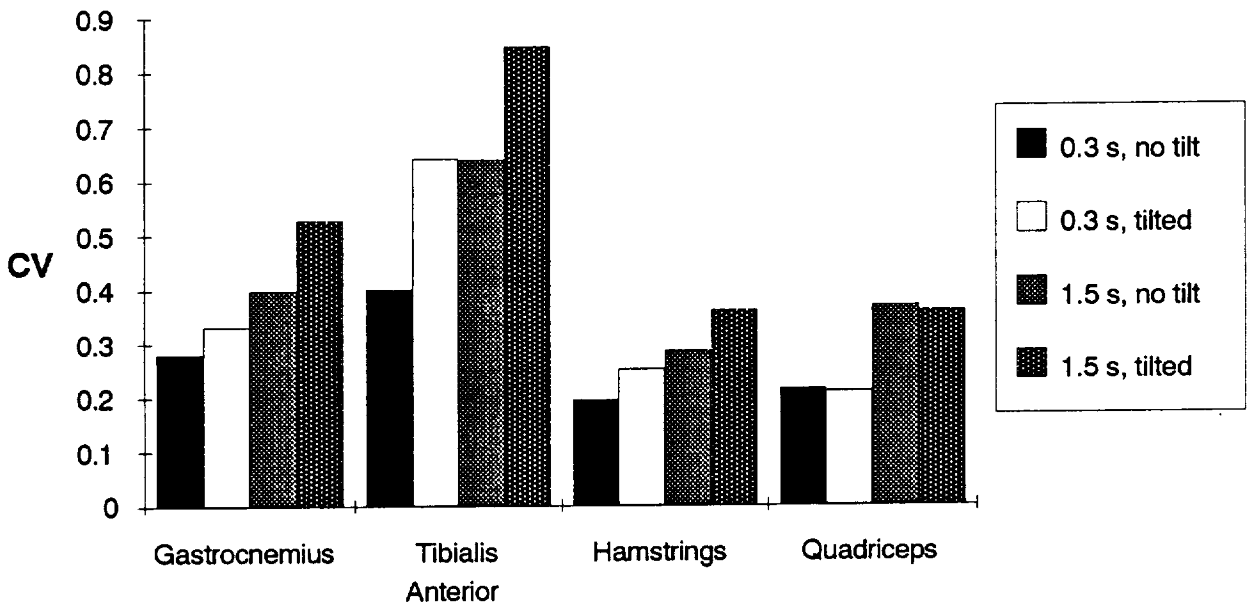


Figure 13: Pre-translation Average RMS-EMG's

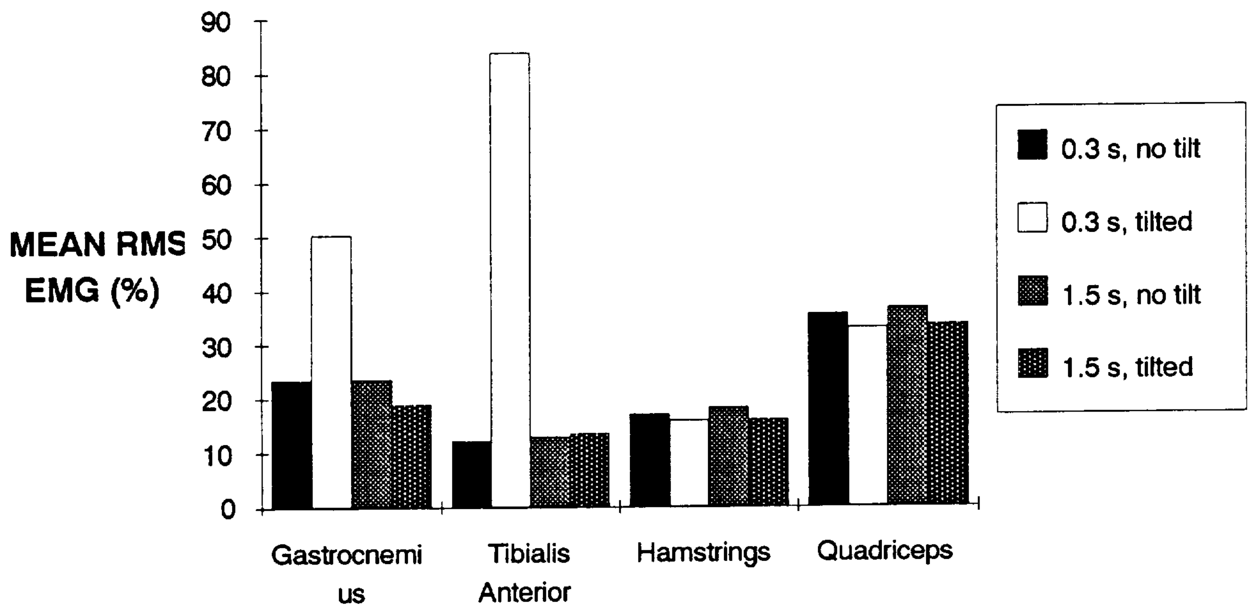


Figure 14: Per-translation Average RMS-EMG's

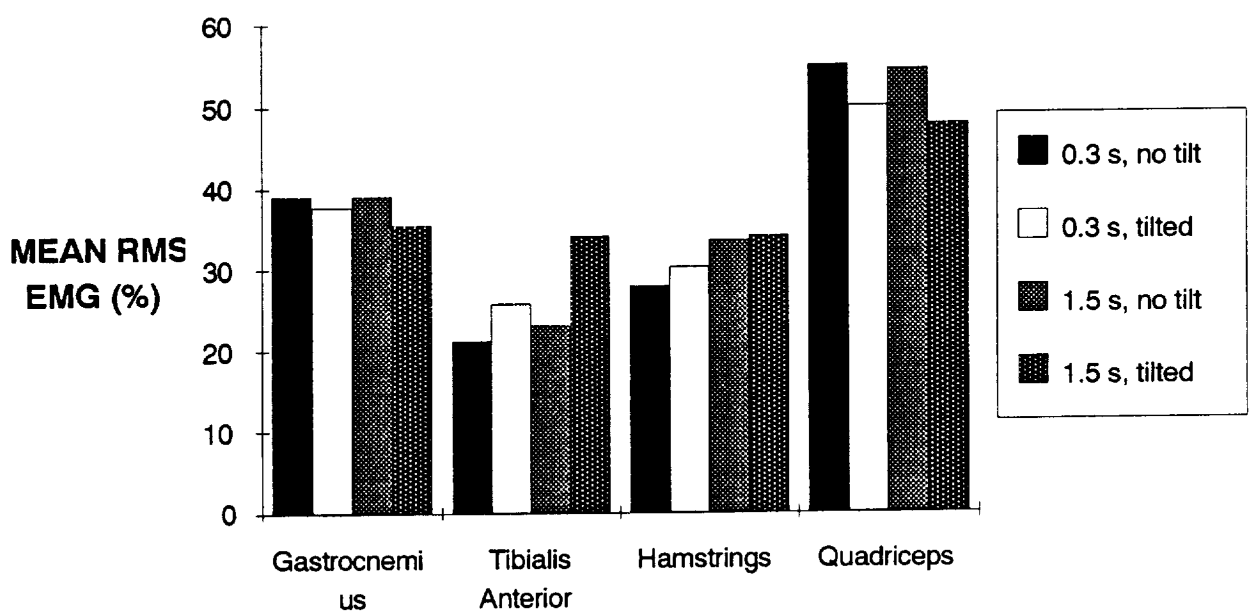
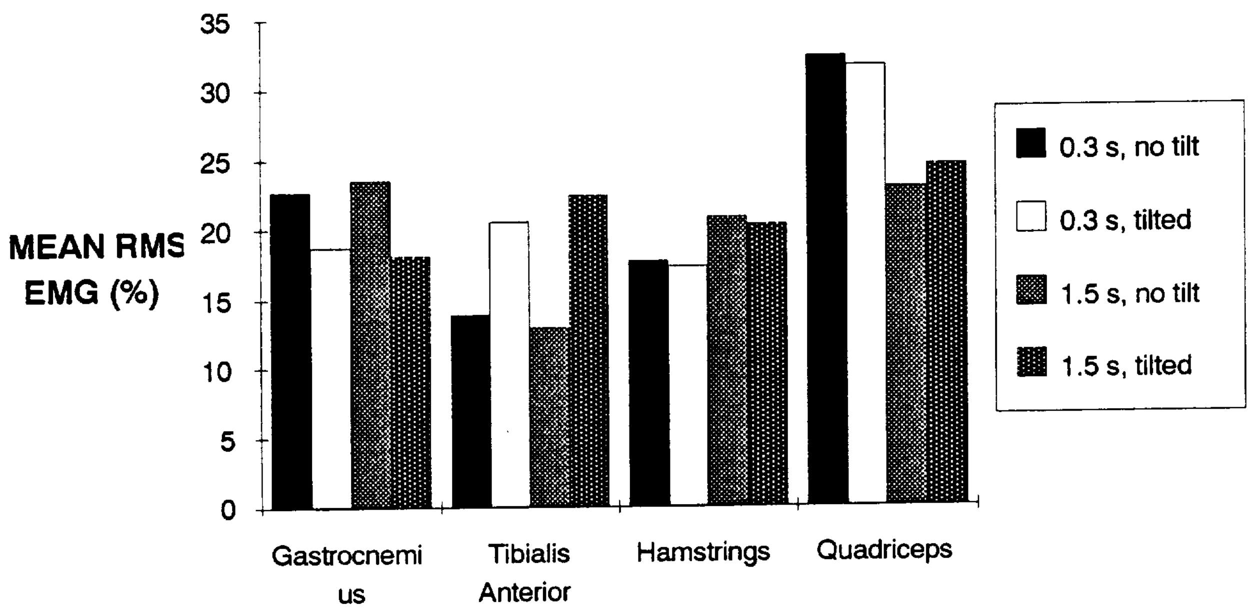


Figure 15: Post-translation Average RMS-EMG's



Window	Variable	Greater on day:
Pre-translation	MEAN G	3
	CV TA	2
	MEAN TA	3
	MEAN H	3
Per-translation	MEAN G	3
	CV TA	2
	MEAN TA	3
	CV H	2
	MEAN H	3
Post-translation	CV G	2
	MEAN G	3
	CV TA	2
	MEAN TA	3
	CV H	2
	CV Q	3

Table 1: ANOVA analysis results with post hoc tests for significant day effects. CV=coefficient of variation, MEAN=average RMS-EMG value scaled to the maximum for that muscle on that day, G=gastrocnemius, TA=tibialis anterior, H=hamstrings, Q=quadriceps.

Window	Variable	Effects		
		Gender	*Duration*	*Tilt*
Pre-translation	CV G	M>F		Y>N
	MEAN G			N>Y
	CV TA	F>M	L>S	Y>N
	MEAN TA	F>M	L>S	
	CV H	F>M	L>S	
	MEAN H			N>Y
	CV Q	F>M	L>S	
	MEAN Q			N>Y
Per-translation	CV G	M>F		Y>N
	MEAN G	F>M	S>L	N>Y
	CV TA	F>M	S>L	Y>N
	MEAN TA	F>M	L>S	Y>N
	CV H	M>F	S>L	Y>N
	MEAN H	M>F	L>S	Y>N
	CV Q	M>F	S>L	Y>N
	MEAN Q	F>M	S>L	N>Y
Post-translation	CV G	M>F	L>S	Y>N
	MEAN G	M>F		N>Y
	CV TA		L>S	Y>N
	MEAN TA	F>M		Y>N
	CV H		L>S	Y>N
	MEAN H	F>M	L>S	
	CV Q		L>S	
	MEAN Q	M>F	S>L	

Table 2: ANOVA analysis results with post hoc tests to indicate significant gender, tilt, and stimulus duration effects. CV=coefficient of variation, MEAN=average RMS-EMG value scaled to the maximum for that muscle on that day, G=gastrocnemius, TA=tibialis anterior, H=hamstrings, Q=quadriceps, M=male, F=female, L=long duration stimulus (1.5 s), S=short duration stimulus (0.3 s), Y=tilted platform (10 degrees of dorsiflexion, and N=no tilt.

APPENDIX A:

Manuscript from the presentation at the XIth International Symposiom of the Society for Postural and Gait Research, Portland, OR, May 24-27, 1992.

Posture and Gait: Control Mechanisms, 1992

XIth International Symposium of the Society for Postural and Gait Research
Portland, May 24-27, 1992

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75 ms for the neck muscles. However, the acceleration was small i.e. (0.35 m/s^2), producing a small and slow downward movement (i.e. 10 mm amplitude after 175 ms). There is obviously no support for vestibular mechanisms being involved in the backward-sway response following a translation, while their contribution cannot be excluded following a rotation.

Our results suggest that the final shape of the response pattern can be formed differently depending on the nature of the multi-sensorial input from pelvis, spine and vestibular afferents. This is in agreement with more recent hypotheses concerning the organization of robust task-dependent postural synergies which can be modified and tuned at lower levels in the motor system (Mc Pherson et al 1988, Hirschfeld and Forssberg 1991).

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RELIABILITY OF EMG PARAMETERS IN RESPONSE TO TRANSIENT POSTURAL PERTURBATIONS

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Introduction

Moving platform posturography has served as a research and a clinical tool for many years. A wide variety of measures have been taken from the subject standing on the perturbable surface, including kinematics such as body sway (Andres and Anderson, 1980; Valk-Fai, 1973), kinetics such as ground reaction forces (Kapteyn et al., 1983), foot pressures (Nakagawa et al., 1990), and muscle activity patterns in the form of EMGs (Nashner, 1976; Diener et al., 1983). Few studies have assessed the retest reliability of criterion measures; Thyssen et al. (1982) reported coefficients of variation (CVs) of resultant average sway displacement taken from force platform data, and Goldie et al. (1989) presented reliability statistics on variability measures taken from force platforms. However, recent investigations have tended more toward the use of EMGs because they represent the motor output of the complex sensorimotor postural control system. The objective of this study was to assess the statistical reliability of EMG criterion measures for transient postural responses so that investigations of the adaptation of muscle activity patterns can be interpreted.

Methods

Eight male and 8 female university undergraduates volunteered for participation in this study. After signing an informed consent form in accordance with University policy, subjects underwent three days of testing. The first day was for acclimation to the equipment and protocol; the data from days 2 and 3 were subsequently analyzed. Subjects were translated rearward (see Andres and Anderson, 1980, for description of stimulus delivery system) at 25 cm/s peak velocity for two different durations (0.3 s and 1.5 s). Two different static tilt conditions were also employed (no tilt and 10 degrees of dorsiflexion) while the platform translated. Ten consecutive trials for each duration and tilt combination were administered in a balanced order across days and subjects, resulting in 40 trials per daily session.

Body sway kinematics were obtained with a MotionAnalysis System (Santa Rosa, CA) using two 60 Hz cameras; these data are not presented here. The focus here is on muscle activity patterns. Surface EMG was recorded from ankle flexors (tibialis anterior, TA), ankle extensors (gastrocnemius/soleus complex, G), hip flexors (quadriceps femoris, Q), and hip extensors (medial hamstring, H). Disposable silver-silver chloride surface electrodes (Hewlett Packard 14445C) were used; the raw EMGs were preamplified, integrated with a 55 msec time constant and amplified (amplification by Analog Devices model AD525 while the RMS integrator was Analog Devices AD536A). Characteristics included low noise (0.3 μV p-p 0.1 Hz to 10 Hz), low nonlinearity (0.003%), high common mode rejection ratio (120 dB), high input impedance (10^9 ohms) and a high gain bandwidth product (25 Mhz). The RMS-EMGs were sampled at a rate of 600 Hz for 4 s, with the first second occurring prior to translation. The resulting RMS-EMGs were scaled to the maximum value from that muscle for that day for each subject. These scaled responses were segmented into three windows: pre-translation, per-translation, and post-translation. Mean RMS-EMG (MEAN) and the coefficient of variation (CV) were calculated for each of these 3 windows. MEANs represent the relative level of muscle activity expressed as % of maximum activity of that muscle on that day, while the CVs represent the variability of the responses about the mean.

Statistical analysis followed a repeated measures ANOVA with subsequent calculation of intraclass coefficients for reliability.

Results

Figures 1 a-c depict the MEAN scaled RMS-EMGs in the respective windows for each duration/tilt combination across days and subjects. Figures 2 a-c present the CV results in similar fashion.

The results of the intraclass reliability analysis are summarized next. The intraclass R ranged from 0.05-0.97 for the MEANs, and from 0.03-0.97 for CVs. There were a total of 96 intraclass R values for the 8 parameters (MEANs and CVs for each of the 4 muscles) in the three windows for the four conditions, so only a brief summary is possible here. Prior to translation, only 2 MEANs (G and Q for 0.3s with no tilt) and 4 CVs (TA and Q for 1.5s with no tilt, and G and Q for 1.5s with tilt) had an intraclass R greater than 0.8. During platform motion only 2 CVs (G for both durations in the platform tilted condition), and post-translation only 2 MEANs (H for 1.5s with tilt, and Q for 0.3s without tilt) had an intraclass R greater than 0.8. The most reliable criterion measures were the MEAN (for 0.3s and no tilt) and CV (for 0.3s with tilt) from Q during the pre-translation window, the CV of G (1.5s, tilt) during the translation, and the MEAN of H with a 1.5s duration and tilted platform post-translation.

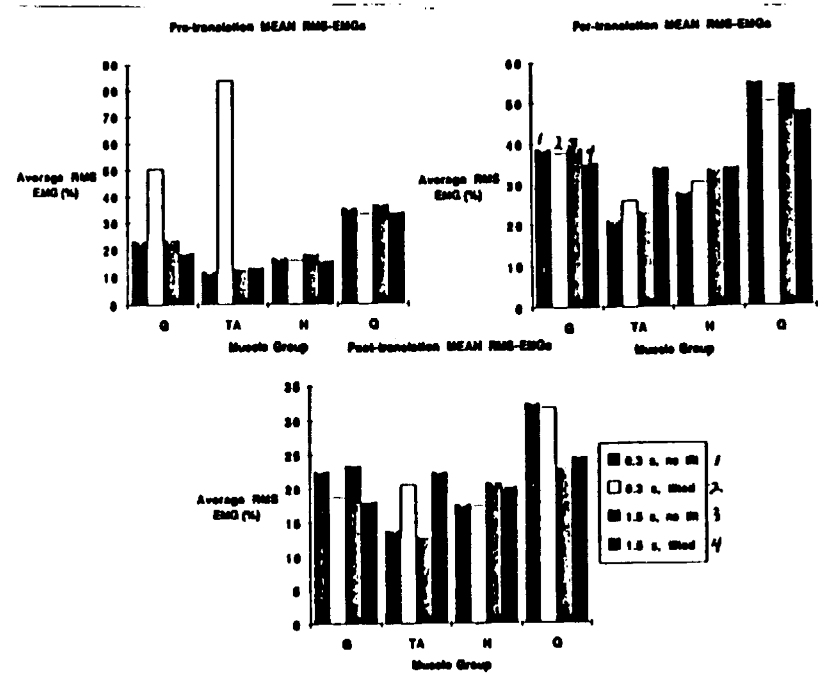

Figure 1: MEAN RMS-EMGs averaged across subjects and days for the following time windows: a) pre-translation, b) per-translation, and c) post-translation. Note that the durations of the per- and post-translation windows depend on stimulus duration.

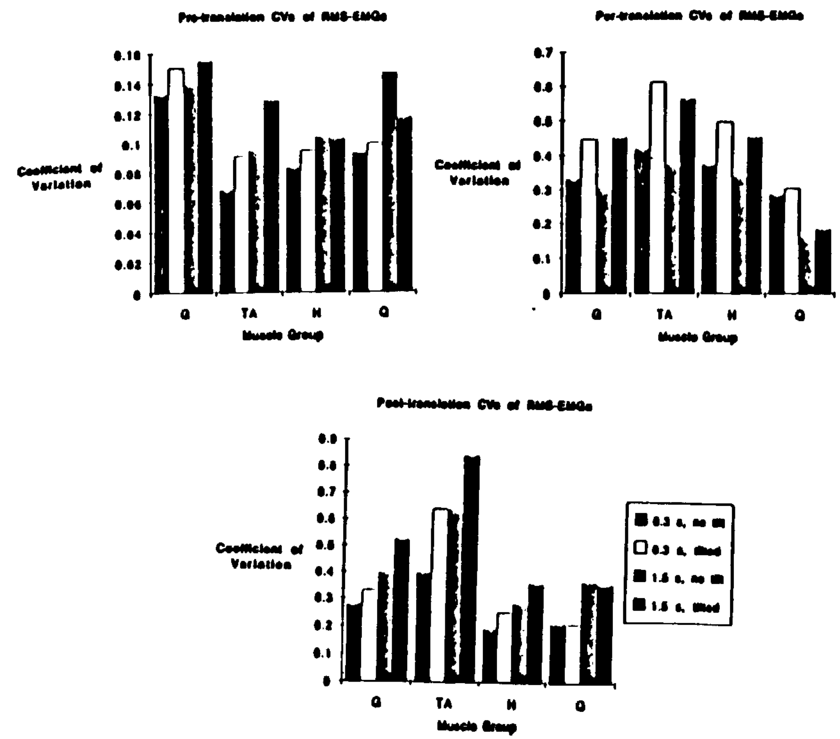

Figure 2: Coefficients of variation within the respective windows, across subjects and days: a) pre-translation, b) per-translation, and c) post-translation. Note that CVs before the stimulus are low, but after the platform stops the high values indicate that activity persists and is modulated.

Discussion

Muscle activities in response to motion of the base of support have been widely presented as criterion measures of postural response. However, latency selection has often involved hand digitization and peak EMG voltages are of questionable utility in making comparisons between sessions or between subjects. The EMG criterion measures selected here were not always reliable or stable. However, different stimulus durations elicited different responses as noted previously (Diener et al., 1984). Responses during platform motion were different than those measured after platform motion ceased; per-translation responses in our study include the stretch reflex (short latency), the medium latency, and long latency compensatory responses, while post-translation responses represented additional compensatory responses either 0.3s or 1.5s after stimulus onset. The selection of these three response windows was based on nystagmography, which separately deals with per- and post-rotatory eye movements, while the pre-translation window represents muscle activity due to pre-innervation. Subjects knew that the platform would translate rearward, but the onset of the motion occurred randomly within a 13 s window, defining the postural set due to anticipation. The preliminary finding from this investigation is that if EMG parameters are to be employed as criterion measures in clinical settings or to probe when the neuromuscular system has completely adapted to a particular platform motion, their reliability must be established.

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THE INFLUENCE OF FOOT SOLES COOLING ON STANDING POSTURAL CONTROL ANALIZED BY TRACKING OF THE CENTER OF FOOT PRESSURE

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Introduction

Sensory information from the sole must be quite important factor for postural control in standing. Particularly, information such as pressure sensation from the mechanoreceptor of the sole is of consequence for humans[1]. There are several reports on the sensory information from the soles aiming to know functional role for postural control[2-4], and various research methods have been developed recently. But these methods are still might need to be improved.

In the past, we have done many studies on this matter using synthetic cryo-air transpire to make the soles less sensitive[5]. And these studies suggested that pressure sensation from each part of the sole plays a different functional role in posture control. But these studies also demonstraited difficulties in evaluating and cooling methods.

In this research, we improved evaluating and cooling methods. Then, we investigated the functional role of pressure sensation in soles on standing postural control based on the relationship between the position of the center of foot pressure (CFP) and control of standing posture.

At first, we investigated method of CFP tracking. Using this method, we clarified relationship between velocity of moving target and aspect of tracking. And through this process, we found the cycle of moving target which was appropriate for evaluating postural control depending on CFP position. Secondly, using this cycle of moving target, we investigated postural control depending on CFP position at cooling soles to make it less sensitive by a cooling apparatus that produced for this study.

Material and Methods

Exp. I

We investigated CFP tracking method which is a new evaluating method. The subjects were ten healthy male. They were 19 to 22 years old with an average age of 20.8 years.

Distribution of pressure giving at sole agree with CFP position and the individual difference was small. We prescribed range of CFP moving by relative length of foot length. And subjects were required to track a continuously moving target to change distribution of pressure giving at the sole continuously. The standing postural control was analyzed by mean error between target- and CFP-movement. And this was considered index as to investigate functional role of pressure sensation.

The subjects were required to track a continuously moving target displayed on a monitor while standing on a force plate. The monitor was setted up at 1.5m ahead of subject and the presented target spot was moved reciprocally and slowly on vertical line and CFP signal spot. The velocity of the target movement was decided as 0.05, 0.10, and 0.15Hz with a triangular wave

APPENDIX B:

Table B-1: The calculation of intraclass correlation coefficients.

	A	B	C	D	E	F	G	H	I	J	K
1	Variable	Wind	DUR	TILT	MSDays	MSTrials	MSSubjects	ERDays	TruVar	A	B
2	C2V	1	1	1	0.00338	0.005595	0.024258	-0.0002215	0.0010439	-0.00011075	0.00027975
3	C3V	1	1	1	0.00357781	0.00651545	0.00611648	-0.00029376	0.00012693	-0.00014688	0.00032577
4	C4V	1	1	1	0.00288	0.00078472	0.00911267	0.00020953	0.00031163	0.00010476	3.9236E-05
5	C5V	1	1	1	0.000125	0.0075125	0.018555	-0.00073875	0.0009215	-0.00036938	0.00037563
6	AV2G	1	1	1	62.6757	9.4616	452.7302	5.32141	19.502725	2.660705	0.47308
7	AV3G	1	1	1	1271.6138	4.8478	570.821	126.6766	-35.03964	63.3383	0.24239
8	AV4G	1	1	1	225.59	4.5882	140.184	22.10018	-4.2703	11.05009	0.22941
9	AV5G	1	1	1	0.52894	19.472	1137.44	-1.894306	56.845553	-0.947153	0.9736
10	C2V	1	1	2	0.002645	0.012133	0.043052	-0.0009488	0.00202035	-0.0004744	0.00060665
11	C3V	1	1	2	0.002205	0.0098376	0.0068483	-0.00076326	0.00023217	-0.00038163	0.00049188
12	C4V	1	1	2	0.0155403	0.0046377	0.0160703	0.00109026	0.0000265	0.00054513	0.00023189
13	C5V	1	1	2	0.03872	0.0012432	0.0166393	0.00374768	-0.00110404	0.00187384	0.00006216
14	AV2G	1	1	2	133.038	16.222	443.095	11.6816	15.50285	5.8408	0.8111
15	AV3G	1	1	2	1181.61	22.138	445.68	115.9472	-36.7965	57.9736	1.1069
16	AV4G	1	1	2	384.61	11.05	151.94	37.356	-11.6335	18.678	0.5525
17	AV5G	1	1	2	583.31	21.53	774.992	56.178	9.5841	28.089	1.0765
18	C2V	1	2	1	0.030225	0.0041474	0.036945	0.00260776	0.000336	0.00130388	0.00020737
19	C3V	1	2	1	0.00045125	0.00307556	0.025943	-0.00026243	0.00127459	-0.00013122	0.00015378
20	C4V	1	2	1	0.068445	0.0042696	0.018172	0.00641754	-0.00251365	0.00320877	0.00021348
21	C5V	1	2	1	0.0154013	0.0117742	0.1545633	0.00036271	0.0069581	0.00018136	0.00058871
22	AV2G	1	2	1	703.446	29.02	541.16	67.4426	-8.1143	33.7213	1.451
23	AV3G	1	2	1	2109.82	10.893	685.07	209.8927	-71.2375	104.94635	0.54465
24	AV4G	1	2	1	652.4	21.12	175.91	63.128	-23.8245	31.564	1.056
25	AV5G	1	2	1	389.9	33.99	895	35.591	25.255	17.7955	1.6995
26	C2V	1	2	2	0.014045	0.013243	0.075107	0.0000802	0.0030531	0.0000401	0.00066215
27	C3V	1	2	2	0.196515	0.007123	0.05007	0.0189392	-0.00732225	0.0094696	0.00035615
28	C4V	1	2	2	0.025383	0.0113613	0.026483	0.00140217	0.000055	0.00070109	0.00056807
29	C5V	1	2	2	0.00032	0.002262	0.1442447	-0.0001942	0.00719624	-0.0000971	0.0001131
30	AV2G	1	2	2	776.79	27.17	500.18	74.962	-13.8305	37.481	1.3585
31	AV3G	1	2	2	278.22	32.445	232.03	24.5775	-2.3095	12.28875	1.62225

	A	B	C	D	E	F	G	H	I	J	K
32	AV4G	1	2	2	519.89	14.2042	239.64	50.56858	-14.0125	25.28429	0.71021
33	AV5G	1	2	2	353.2	26.45	629.32	32.675	13.806	16.3375	1.3225
34	C2V	2	1	1	0.04489	0.0044455	0.0406952	0.00404445	-0.00020974	0.00202223	0.00022228
35	C3V	2	1	1	0.465888	0.12106	0.1004952	0.0344828	-0.01826964	0.0172414	0.006053
36	C4V	2	1	1	0.026645	0.012985	0.02769	0.001366	0.00005225	0.000683	0.00064925
37	C5V	2	1	1	0.010695	0.007064	0.023845	0.0003631	0.0006575	0.00018155	0.0003532
38	AV2G	2	1	1	788.2	14.7	501.7	77.35	-14.325	38.675	0.735
39	AV3G	2	1	1	2094.7	69.9	554.9	202.48	-76.99	101.24	3.495
40	AV4G	2	1	1	720.8	73.9	387.1	64.69	-16.685	32.345	3.695
41	AV5G	2	1	1	416.1	16.6	580.2	39.95	8.205	19.975	0.83
42	C2V	2	1	2	0.0005	0.0107113	0.135217	-0.00102113	0.00673585	-0.00051057	0.00053557
43	C3V	2	1	2	0.81205	0.11848	0.15808	0.069357	-0.0326985	0.0346785	0.005924
44	C4V	2	1	2	0.00578	0.017442	0.100533	-0.0011662	0.00473765	-0.0005831	0.0008721
45	C5V	2	1	2	0.013913	0.00529	0.03837	0.0008623	0.00122285	0.00043115	0.0002645
46	AV2G	2	1	2	481.4	60.3	525.1	42.11	2.185	21.055	3.015
47	AV3G	2	1	2	1411.6	167.9	474.7	124.37	-46.845	62.185	8.395
48	AV4G	2	1	2	764.2	107.7	424.7	65.65	-16.975	32.825	5.385
49	AV5G	2	1	2	329.1	25.8	706.6	30.33	18.875	15.165	1.29
50	C2V	2	2	1	0.03916	0.00111	0.08191	0.003805	0.0021375	0.0019025	0.0000555
51	C3V	2	2	1	0.4759	0.0605	0.06444	0.04154	-0.020573	0.02077	0.003025
52	C4V	2	2	1	0.3768	0.01004	0.0609	0.036676	-0.015795	0.018338	0.000502
53	C5V	2	2	1	0.002	0.001303	0.00548	0.0000697	0.000174	0.00003485	0.00006515
54	AV2G	2	2	1	724.8	36.2	601.3	68.86	-6.175	34.43	1.81
55	AV3G	2	2	1	1813.7	130.1	687.9	168.36	-56.29	84.18	6.505
56	AV4G	2	2	1	268.5	126.4	648.2	14.21	18.985	7.105	6.32
57	AV5G	2	2	1	190	15.6	486.9	17.44	14.845	8.72	0.78
58	C2V	2	2	2	0.00001125	0.00424	0.19463	-0.00042288	0.00973094	-0.00021144	0.000212
59	C3V	2	2	2	0.017701	0.01179	0.04869	0.0005911	0.00154945	0.00029555	0.0005895
60	C4V	2	2	2	0.126803	0.00651	0.068173	0.0120293	-0.0029315	0.00601465	0.0003255
61	C5V	2	2	2	0.00761	0.002267	0.03344	0.0005343	0.0012915	0.00026715	0.00011335
62	AV2G	2	2	2	128.8	75.5	491.6	5.33	18.14	2.665	3.775

	A				E	F	G	H	I	J	K
6	AV3G		2	2	2	0.6938	269	591.5	-26.83062	29.54031	-13.41531	13.
6	AV4G		2	2	2	5.9	121.7	814.9	-11.58	40.45	-5.79	6.0
6	AV5G		2	2	2	628.4	33.7	595.3	59.47	-1.655	29.735	1.6
6	C2V		3	1	1	0.02398	0.00415	0.04838	0.001983	0.00122	0.0009915	0.000207
6	C3V		3	1	1	0.0619	0.7294	0.2702	-0.06675	0.010415	-0.033375	0.0364
6	C4V		3	1	1	0.002645	0.0137	0.0829	-0.0011055	0.00401275	-0.00055275	0.00068
6	C5V		3	1	1	0.000113	0.02842	0.12216	-0.0028307	0.00610235	-0.00141535	0.00142
7	AV2G		3	1	1	222.2	13.2	623.1	20.9	20.045	10.45	0.
7	AV3G		3	1	1	1341.4	107.9	522.4	123.35	-40.95	61.675	5.3
7	AV4G		3	1	1	266.1	14.6	222.5	25.15	-2.18	12.575	0.
7	AV5G		3	1	1	282.9	37.7	1850	24.52	78.355	12.26	1.8
7	C2V		3	1	2	0.00002	0.0381	0.07096	-0.003808	0.003547	-0.001904	0.00190
7	C3V		3	1	2	0.5017	0.126	0.2086	0.03757	-0.014655	0.018785	0.006
7	C4V		3	1	2	0.0717	0.08513	0.0848	-0.001343	0.000655	-0.0006715	0.004256
7	C5V		3	1	2	0.2322	0.0149	0.13	0.02173	-0.00511	0.010865	0.00074
7	AV2G		3	1	2	497.3	13.3	582.3	48.4	4.25	24.2	0.6
7	AV3G		3	1	2	1011.3	412.5	658	59.88	-17.665	29.94	20.62
8	AV4G		3	1	2	142.2	18.1	391.2	12.41	12.45	6.205	0.9
8	AV5G		3	1	2	637.5	98.3	759.4	53.92	6.095	26.96	4.9
8	C2V		3	2	1	0.2055	0.0107	0.1086	0.01948	-0.004845	0.00974	0.00053
8	C3V		3	2	1	0.9625	0.092	0.2646	0.08705	-0.034895	0.043525	0.004
8	C4V		3	2	1	0.4433	0.023	0.0476	0.04203	-0.019785	0.021015	0.0011
8	C5V		3	2	1	0.01755	0.0419	0.1927	-0.002435	0.0087575	-0.0012175	0.00209
8	AV2G		3	2	1	1035.4	66.7	497.7	96.87	-26.885	48.435	3.3
8	AV3G		3	2	1	1153.6	44.1	509.1	110.95	-32.225	55.475	2.2
8	AV4G		3	2	1	53.8	191.2	464.1	-13.74	20.515	-6.87	9.
8	AV5G		3	2	1	826.22	175.4	1249.7	65.082	21.174	32.541	8.
9	C2V		3	2	2	0.3976	0.0533	0.2876	0.03443	-0.0055	0.017215	0.00266
9	C3V		3	2	2	0.03121	0.1372	0.42895	-0.010599	0.019887	-0.0052995	0.0068
9	C4V		3	2	2	0.6994	0.0319	0.0965	0.06675	-0.030145	0.033375	0.00159
9	C5V		3	2	2	0.4898	0.00356	0.1138	0.048624	-0.0188	0.024312	0.00017

	A				E	F	G	H	I	J	K
9	AV2G		3	2	2	243.4	21.1	718.6	22.23	23.76	11.115	1.0
9	AV3G		3	2	2	83.9	390.6	676.3	-30.67	29.62	-15.335	19.
9	AV4G		3	2	2	110.5	37.8	596	7.27	24.275	3.635	1.
9	AV5G		3	2	2	504.9	91.6	786.6	41.33	14.085	20.665	4.

	L	M	N
	R	%A	%B
	0.86066452	-0.09131008	0.23064556
	0.41505408	-0.48028278	1.06522869
	0.68395651	0.22993042	0.08611307
	0.99326327	-0.39814066	0.40487739
	0.86156059	0.11754042	0.02089898
	-1.22769274	2.21920006	0.00849268
	-0.60924214	1.5765123	0.03272984
	0.99953497	-0.01665412	0.01711914
1	0.93856267	-0.22038465	0.28182198
1	0.67802228	-1.11452477	1.43650249
1	0.03298009	0.67843164	0.28858827
1	-1.32702097	2.25230629	0.07471468
1	0.69975287	0.26363647	0.03661066
1	-1.65125202	2.60157961	0.04967241
1	-1.53132816	2.45860208	0.07272608
1	0.24733417	0.7248849	0.02778093
1	0.181892	0.70584924	0.11225876
1	0.9826061	-0.10115677	0.11855067
2	-2.76650891	3.53155404	0.23495488
2	0.90035604	0.02346676	0.0761772
2	-0.29988543	1.24625989	0.05362555
2	-2.07971448	3.06381392	0.01590056
2	-2.70871468	3.58865329	0.1200614
2	0.56435754	0.3976648	0.03797765
2	0.81300012	0.0106781	0.17632178
2	-2.92480527	3.78254444	0.14226083
2	0.04153608	0.52946041	0.42900351
2	0.99778155	-0.01346323	0.01568169
3	-0.55302091	1.49870047	0.05432044
3	-0.19906909	1.05923803	0.13983106

	L	M	N
3	-1.16946253	2.11018945	0.05927308
3	0.4387593	0.51921121	0.04202949
3	-0.1030785	0.99383957	0.10923893
3	-3.63592291	3.43128826	1.20463465
3	0.03773926	0.49331889	0.46894186
3	0.5514783	0.15227511	0.29624659
3	-0.5710584	1.54175802	0.02930038
3	-2.7749144	3.64894576	0.12596864
4	-0.86205115	1.67114441	0.19090674
4	0.28283351	0.68855567	0.02861082
4	0.99630224	-0.07551787	0.07921563
4	-4.13695597	4.38746204	0.74949393
4	0.94250644	-0.11600171	0.17349527
4	0.63739901	0.22473286	0.13786813
4	0.08322224	0.80194249	0.11483527
4	-1.97366758	2.61997051	0.35369707
4	-0.7993878	1.54579703	0.25359077
4	0.53424851	0.42923861	0.03651288
5	0.5219143	0.46453424	0.01355146
5	-6.38516449	6.44630664	0.93885785
5	-5.18719212	6.02233169	0.16486043
5	0.6350365	0.12718978	0.23777372
5	-0.20538833	1.14518543	0.06020289
5	-1.63657508	2.44744876	0.18912633
5	0.585776	0.21922246	0.19500154
5	0.60977613	0.35818443	0.03203943
5	0.9999422	-0.02172712	0.02178493
5	0.63645512	0.1214007	0.24214418
6	-0.8600179	1.76452555	0.09549235
6	0.77242823	0.15977871	0.06779306
6	0.73799837	0.10842148	0.15358015

	L	M	N
6	0.99882705	-0.45360304	0.45477599
6	0.99275985	-0.14210333	0.14934348
6	-0.05560222	0.9989921	0.05661011
6	0.50434064	0.40988012	0.08577925
6	0.77091044	-2.4703923	2.69948187
6	0.96809409	-0.13335344	0.16525935
6	0.99907498	-0.23172069	0.23264571
7	0.64339592	0.33541968	0.0211844
7	-1.56776417	2.36121746	0.20654671
7	-0.19595506	1.13033708	0.06561798
7	0.84708108	0.13254054	0.02037838
7	0.99971815	-0.53664036	0.53692221
7	-1.4050815	1.80105465	0.60402685
7	0.15448113	-0.15837264	1.00389151
7	-0.78615385	1.67153846	0.11461538
7	0.14597287	0.83118667	0.02284046
7	-0.53693009	0.9100304	0.6268991
8	0.63650307	0.31722904	0.04626789
8	0.16052146	0.71003424	0.1294443
8	-0.89226519	1.79373849	0.0985261
8	-2.63756614	3.2898715	0.34769463
8	-8.31302521	8.82983193	0.48319328
8	0.90892579	-0.12636222	0.21743643
8	-1.0803697	1.94635322	0.13401648
8	-1.26595954	2.17933608	0.08662345
8	0.88407671	-0.29605688	0.41198018
8	0.33886533	0.52078099	0.14035368
9	-0.38247566	1.19714882	0.18532684
9	0.92724094	-0.24709174	0.3198508
9	-6.24766839	6.91709845	0.33056995
9	-3.30404218	4.27275923	0.03128295

	L	M	N
9	0.66128583	0.30935152	0.02936265
9	0.87594263	-0.45349697	0.57755434
9	0.81459732	0.12197987	0.06342282
9	0.35812357	0.52542588	0.11645055

APPENDIX C:

Manuscript from the Journal of Biomechanics that presents the dynamic biomechanical model that has been modified for this project.

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VALIDATION OF A BIODYNAMIC MODEL OF PUSHING AND PULLING

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Abstract—Pushing and pulling during manual material handling can increase the compressive forces on the lumbar disc region while creating high shear forces at the shoe–floor interface. A sagittal plane dynamic model derived from previous biomechanical models was developed to predict L5/S1 compressive force and required coefficients of friction during dynamic cart pushing and pulling. Before these predictions could be interpreted, however, it was necessary to validate model predictions against independently measured values of comparable quantities. This experiment used subjects of disparate stature and body mass, while task factors such as cart resistance and walking speed were varied. Predicted ground reaction forces were compared with those measured by a force platform, with correlations up to 0.67. Predicted erector spinae and rectus abdominus muscle forces were compared with muscle forces derived from RMS-EMGs of the respective muscle groups, using a static force build-up regression relationship to transform the dynamic RMS-EMGs to trunk muscle forces. Although correlations were low, this was attributed in part to the use of surface EMG on subjects of widely varied body mass. The biodynamic model holds promise as a tool for analysis of actual industrial pushing and pulling tasks, when carefully applied.

INTRODUCTION

Carts of various sizes, weights and configurations are frequently pushed or pulled manually in many industrial situations. Among these are the tyre manufacturing industry, the fiberglass manufacturing industry, commercial laundries, and the airline industries. A large proportion of these tasks involve the worker pulling on doors and hoses, pushing on carts, and generally attempting a task which imparts a high shearing force to the feet, with resulting slip and fall injuries (Safety Sciences, 1977). It is estimated that over 20% of the worker's compensation costs each year are a result of fall or slip related injuries in the U.S. (Szymusiac and Ryan, 1982). A study concerning a large manufacturing operation in England reported that 36–45% of back pain was caused by a slip or fall (Manning, 1983). These statistics paint a dangerous picture for workers involved with pushing or pulling tasks, which increase the risks of slipping and falling or overexerting the back.

Several investigators have addressed the performance aspects of push/pull tasks. Dempster (1958) studied static pull forces during standing. Kroemer measured maximal isometric pushing forces in 65 different positions and assessed the effects of varied foot friction during pushing (Kroemer, 1969, 1971). Ayoub and McDaniel (1974) studied the loading of the lumbar spine during static pushes and pulls against a wall, in various body postures. One Swedish study (Winkel, 1983) concerned the manual handling of food and beverage carts on wide-body airplanes and measured only the hand forces exerted on stationary carts. Several recommendations were made about cart configuration and loading as a result of this study. However, none of these studies considered the dynamic case where the worker moves during the push or pull task.

Pushing and pulling hand forces have been measured while the subject walked on a treadmill (Snook *et al.*, 1970) with different handle heights and adjustable treadmill resistance. Strindberg and Peterson (1972) used psychophysical methods to study force perception while pushing trolleys. These studies began to approach more realistic dynamic simulations of actual industrial situations. A German group studied the load on the spine during the transport of dustbins (Jager *et al.*, 1984). These authors utilized a simple static model of L5/S1 torques, and they also measured the EMG activity of back, leg, and hand muscles. However, no validation of their model was offered, and the EMG information for the back muscles was not compared with the L5/S1 torque predictions.

The doctoral research of Lee (Lee, 1982; Lee *et al.*, 1989) formulated a dynamic biomechanical model of cart pushing and pulling. The inputs to the model included subject anthropometry, body postures during dynamic tasks, and hand forces exerted on the cart handle. The specific model predictions were horizontal and vertical foot forces and gross torso muscle and vertebral column loadings when pushing or pulling. Laboratory validation of the model took place with a cart simulator. Dynamic foot forces, hand forces required to move the cart, body motions at various speeds, and back muscle actions were measured while six subjects pushed or pulled the cart simulator. Foot force predictions were compared to measured (by force platform) foot forces, while predicted torso muscle forces were converted to 'equivalent' integrated electromyograms (IEMGs) and then compared

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with the measured IEMGs. These validation experiments were performed at three handle heights (66, 109, 152 cm) and three cart resistances (light, medium, heavy). Two speeds (approximately 1.8 and 3.6 km h^{-1}) were used at only the middle handle height. The four male and two female subjects ranged in stature from 161 to 175 cm, and the body mass range was 50–80 kg.

Lee *et al.* (1989) reported the results of static calibrations performed for the dynamic experiment, but the dynamic results have not been published. The purpose of our investigation was to validate a model (based on Lee's model) in the laboratory with a broader anthropometric population.

METHODS

Model development

The biomechanical model used for this study was based on that used by Lee (1982). The mathematical relationships and assumptions which allow the prediction of the desired outputs (foot reactive forces and L5/S1 compressive forces) will be described. There were several intermediate steps before the predictions were made; these will also be examined.

Two different sets of link lengths were calculated. The distance between the joint markers (LEDs), in the sagittal plane, was measured by the position detection system. The Y and Z coordinates were calibrated first, corrected for distortion, and smoothed with a digital filter with optimal cut-off frequencies [determined for each data record as described by Jackson (1979)] before calculating relative link lengths (link lengths in the sagittal plane, possibly foreshortened if the link was out of plane). Absolute link lengths were based on ratios to total stature (Drillis and Contini, 1966).

The position of L5/S1 was calculated because there was no marker on that location. These calculations were based on Chaffin and Andersson (1984, Chap. 6). The link center of gravity (CG) positions were calculated with respect to relative link length. Ratios based on Dempster (1955) were used.

Link masses were calculated as a ratio to total body mass. A major assumption made at this point was that the left arm position was the same as the right arm, as viewed from the sagittal plane. When markers were placed on the inside of the left arm, they were never detected by the position detection system because they were blocked by the right arm. Therefore, the masses of the forearm and hand link and the upper limb link were doubled to represent both arms acting in the same sagittal plane position. The ratios were again from Dempster (1955).

Joint angles were calculated from joint marker coordinates with respect to the horizontal. The joint angles correspond to the angle from the horizontal of respective links (i.e. joint angle No. 1 was the angle from the horizontal of link No. 1, the forearm and hand; see Fig. 1).

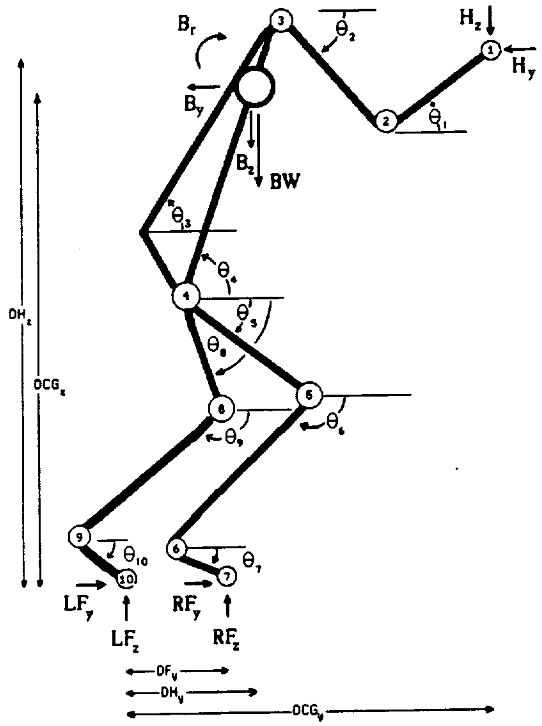


Fig. 1. Free body diagram of double support stance with joint angles used by the model and locations of joint center markers. Variables and resulting equilibrium equations are given in the text.

The whole body CG position was calculated using the segmental moments method (Miller and Nelson, 1976). The location of the upper body CG was calculated with the same approach using only links above L5/S1 (hands, forearms, upper arms, and trunk from shoulder to L5/S1).

Mass moments of inertia were calculated for each link as the product of the link mass and K^2, where K was the product of relative link length and a constant representing the radius of gyration (Chandler *et al.*, 1975; Plagenhoef, 1966).

Link CG, whole body CG, upper body CG, and joint angle data were differentiated twice so that the respective accelerations could be used for calculation of inertial forces. The derivatives were calculated with a finite-impulse–response digital recursive filter (Oppenheim and Schafer, 1975; Lanshammar, 1982). Optimal filter coefficients were derived for these data and subsequently applied uniformly.

The linear and rotational inertial forces resist body linear and angular accelerations. These inertial forces were calculated for each link as the negative of the product of the link mass and the linear acceleration in Y and Z directions. Whole body inertial forces acting at the whole body CG were calculated as the negative of the product of the whole body mass and the Y and Z accelerations of the whole body CG. Since there was no overall body angular acceleration, the whole body rotational inertial force was calculated as the sum of the individual link rotational inertial forces.

Calculation of ground reaction forces at the feet

Two different situations exist during normal walking gait: a single support phase, when only one foot contacts the support surface while the other foot swings through to its next placement; and a double support phase, when both feet are in contact with the support surface (see Fig. 1). Foot force calculations are detailed below for each case.

Single support. Assume:

(1) two arms act as one (sum Y and Z forces from the separate handles);

(2) quasi-static equilibrium,

$$\Sigma F_y = 0 = RF_y - H_y - B_y,$$
$$RF_y = H_y + B_y,$$
$$\Sigma F_z = 0 = RF_z - H_z - BW - B_z,$$
$$RF_z = H_z + BW + B_z,$$

where RF_y is the Y reactive force of the right foot; RF_z is the Z reactive force of the right foot; H_y is the total Y hand force; H_z is the total Z hand force; BW is the total body weight; B_y is the Y body inertial force; and B_z is the Z body inertial force.

Moments at the foot during single support were not used by the model and were not needed because there were only two unknowns with two equations.

Double support. Assume:

(1) quasi-static equilibrium;

(2) two arms act as one;

(3) moment arms from heel marker for pushing, from toe marker of rear foot for pulling,

$$\Sigma F_y = 0 = LF_y + RF_y - H_y - B_y,$$
$$\Sigma F_z = 0 = LF_z + RF_z - H_z - B_z - BW.$$

Left foot back.

$$\begin{aligned}\Sigma M_x = 0 = & -RF_z * (DF_y) + (B_z + BW) * DCG_y \\ & - B_y * (DCG_z) - H_y * (DH_z) \\ & + H_z * (DH_y) + B_r \\ = & \, DCG_y * (B_z + BW) + DH_y * (H_z) + B_r \\ & - DF_y * (RF_z) - DCG_z * (B_y) \\ & - DH_z * (H_y),\end{aligned}$$

$$\begin{aligned}RF_z = (& DCG_y * (B_z + BW) + DH_y * (H_z) + B_r \\ & - DCG_z * (B_y) - DH_z * (H_y))/DF_y,\end{aligned}$$

$$LF_z = H_z + B_z + BW - RF_z,$$

where LF_y is the Y reactive force of the left foot; LF_z is the Z reactive force of the left foot; DCG_y is the Y distance from the rear heel to the whole body center of gravity; DF_y is the Y distance from the rear heel to the Z foot force of other foot; DH_y is the Y distance from the rear heel to the handle; DCG_z is the Z distance from the floor to the whole body center of gravity; DH_z is the Z distance from the floor to the handle; and B_r is the rotational body inertial force.

Similar equations result when the right foot is back for the other portion of double support during right foot stance, with appropriate exchange of right and left foot reactive forces. At this stage, the system is indeterminate because we have three equations (M_x, F_y, and F_z) and four unknowns (LF_z, RF_z, LF_y, and RF_y). The solution requires another equation to become determinate; the model assumed that the friction utilization under both feet was the same, and hence the ratios of horizontal (y) to vertical (z) foot forces at each foot were equal (Lee, 1982).

Calculation of trunk muscle forces

The calculation of L5/S1 compressive forces and the muscle forces contributing to these forces begins with the calculation of abdominal pressure, because this pressure counteracts some of the contraction force of the erector spinae muscles (see Fig. 2). An empirical prediction of abdominal pressure was performed using previously reported equations (Lee *et al.*, 1989; Chaffin and Andersson, 1984) derived from work done by Morris *et al.* (1961).

The moment arm at which F_{ABD} acts has been assumed by Chaffin (1975) to vary as the sine of hip angle, with an erect position having a moment arm of 7 cm, increasing to about 15 cm when stooped over at $\phi_H = 90°$ from vertical (where ϕ_H = the angle from the hip-to-shoulder link to vertical). The argument that F_{ABD} acts parallel to the compressive force on L5/S1 was presented by Chaffin and Andersson (1984). The line of action of rectus abdominus has also been parallel to the compressive force on L5/S1 in other studies (Schultz and Andersson, 1981; Chaffin and Andersson, 1984). This model assumes that all muscle forces act normal to the shear force to create compression only. Reactive shearing forces are then produced by lumbar facet joints, as described in Chaffin and Andersson (1984).

The following equations were used by the model to calculate back and muscle forces (see Fig. 2):

$$\begin{aligned}F_C = & \, ESMF + RAMF - F_{ABD} \\ & + \sin(\alpha) * (BW_u + H_z + UB_z) \\ & + \cos(\alpha) * (UB_y + H_y),\end{aligned}$$

$$F_S = \cos(\alpha) * (BW_u + H_z + UB_z) + \sin(\alpha) * (UB_y + H_y),$$

where F_C is the L5/S1 compressive force; F_S is the L5/S1 shear force due to external forces only (assuming that all muscles act to create compression only); $ESMF$ is the erector spinae muscle force (when resultant moment at L5/S1 was negative), calculated as the resultant moment at L5/S1 divided by the moment arm [0.06 m for males and females, based on Kumar (1988)]; $RAMF$ is the rectus abdominus muscle force (when resultant L5/S1 moment was positive), calculated as the resultant moment divided by the moment arm [0.10 m for males and females, from Kumar (1988)]; F_{ABD} is the abdominal force due to intra-abdominal pressure; α is the angle from horizontal to L5/S1-shoulder link; BW_u is the body weight above

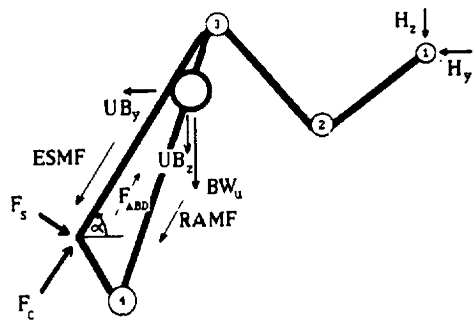


Fig. 2. Free body diagram of the body above the hips, used to calculate L5/S1 shear and compressive forces and erector spinae and rectus abdominus muscle forces. All trunk forces are acting parallel or normal to the compressive force. (Note: it is assumed these muscles act to provide only compression forces on the spinal discs.)

L5/S1; UB_z is the upper body Z inertial force; and UB_y is the upper body Y inertial force.

Static calibration experiments

Linear regressions were formed between the predicted torso muscle forces (erector spinae muscle forces [*ESMF*s] and rectus abdominus muscle forces [*RAMF*s]) and the measured RMS-EMGs from both the right and the left groups of these muscles during a gradual increase of push or pull forces. The subject either pushed or pulled in a free static posture. Starting with no exerted hand force, the subject built up to a near maximum push or pull force in a 5 s period. The regression coefficients were used subsequently to transform dynamic RMS-EMG values to muscle force values to provide a basis for comparison to predicted dynamic muscle force values.

Experimental design and statistical analysis

A fractional factorial design was used to examine the accuracy and precision of the model predictions as a function of the independent variables, described below. In particular, 20 subjects were selected according to a design which emphasized anthropometric extremes (< 20th percentile, > 80th percentile for both stature and body weight) and included subjects of 50th percentile stature and weight, with two males and two females in each of the five cells (see Table 1). Healthy young subjects (18–31 yr) volunteered for this experiment. After signing an informed consent form, consistent with University policy, they were weighed and their heights were measured.

Model validation involved three types of analyses: (i) correlations of measured vs predicted variables; (ii) ANOVA analysis of measured and predicted variables; and (iii) ANOVA analysis of residuals formed by taking the difference between measured and predicted values. Scheffe multiple comparison tests were performed *post hoc*, when appropriate.

Linear statistical models were formed for the ANOVA analysis with the following factors (and corresponding levels): subject ($n=20$), sex (male, female), height (short, average, tall), weight (light, average, heavy), cart resistance (low, high), walking speed (60, 100 steps min^{-1}), and direction of resistance (push, pull). These factors will be abbreviated SN, SX, HT, WT, CR, WS, and DR, respectively. Only the main effects are presented in this report; first order interactions were not significant.

Table 1. Subject stature and weight descriptions for experimental design to emphasize anthropometric extremes

		Weight (percentile)		
		< 20	50	> 80
Males				
Height (percentile)	< 20	0.170 m 626.0 N	— —	1.75 m 885.5 N
	50	— —	1.78 m 712.0 N	— —
	> 80	1.85 m 689.0 N	— —	1.84 m 878.0 N
Females				
Height (percentile)	< 20	1.54 m 438.0 N	—	1.58 m 845.0 N
	50	— —	1.62 m 572.5 N	— —
	> 80	1.72 m 569.5 N	— —	1.76 m 689.0 N

Stature and weight percentiles were derived from National Health Survey (1965). There were two subjects in each cell, so only the mean values are given.

Analytical methods

The differences between independently measured or derived variables and the same variables predicted by the biodynamic model provided the most critical assessment of model performance. Beyond visual comparisons, one technique to assess differences between predicted and measured quantities was to compare the mean values of the predicted and measured dependent variables for each of the two time-windows (single and total support period for the right leg).

Another method of comparing measured and predicted quantities involved the formation of residuals. The GRF residuals were calculated on a point by point basis throughout the support phase by summing the model predicted force exerted on the foot by the ground and the measured force exerted on the ground by the foot. This was equivalent to subtracting measured values of force exerted on the foot by the ground from the model predicted forces. These residuals were actually intra-subject comparisons. Interpretation of these results requires some explanation: the average represents the sum of the differences between predicted and measured values divided by the number of differences taken. These residual parameters were submitted subsequently to ANOVA analysis to determine which subject and task factors contributed significantly to the errors in the predictions.

Data acquisition hardware

The cart simulator had handles 0.5 m apart and oriented so that they were horizontal; hence the subject gripped each handle with the hands prone. The simulator travelled on teflon bushings over aluminum rails, while the cart resistance came from a strap which passed over a variable number of dowels which were affixed to the bottom of the simulator. By changing the tension on this strap, cart resistance varied from 88 to 128 N (the horizontal force necessary to keep the cart moving at approximately 0.5 m s^{-1}). The electronics for the portable handles and the EMGs were carried by the cart (see Fig. 3).

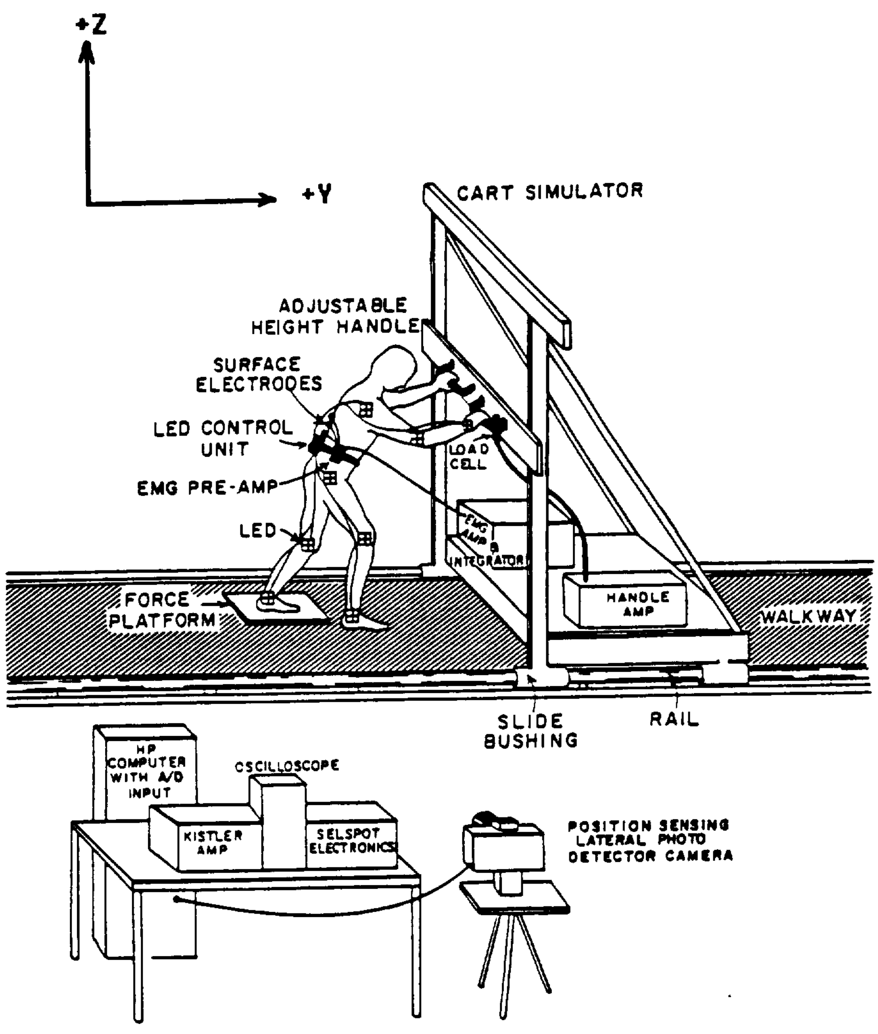


Fig. 3. Diagram of laboratory equipment configuration and the coordinate system used for kinematics and forces exerted on the body.

Approximated sagittal plane joint center coordinates were sensed by a single camera SELSPOT system (Selcom Selective Electronic Inc., Valdese, North Carolina, U.S.A.) sampling at 50 Hz. The 10 LEDs were placed on the subject as shown in Fig. 1. To prevent distortion the following conditions were placed on LED position detection: (1) the data window was in the central portion of the camera viewing field; (2) lens distortion was compensated for by calibration; and (3) reflection problems were minimized by keeping the reflective laboratory floor out of the viewing window. Hand forces were measured with portable handles which used columns instrumented with strain gauges to detect the horizontal and vertical components of force. The force platform was a Kistler force platform (Kistler Instrumente AG, Winterthur, Switzerland, model 9231A) with six Kistler charge amplifiers (5001) and a Kistler central control unit (5671A). The surface of the force platform was covered with painted plywood, as was the rest of the runway. The static coefficient of friction (COF) of the surface when rubber shoes were worn was approximately 0.7, while the dynamic COF was about 0.6 (see Andres *et al.*, 1984).

The myoelectrical activity of the torso muscles was detected by bipolar surface electrodes (Hewlett Packard, Andover, Massachusetts, U.S.A., model 14445C disposable electrodes). The electrodes were placed 3 cm lateral to the midline on both sides of the spine at the L2 and L3 level, about 5 cm apart so that the activity of one side vs the other could be observed. The rectus abdominus activity was recorded by placing the electrodes in a similar manner on the abdomen (centered 3 cm lateral to the linea alba, above the navel). The electrode signals were sent to preamplifiers in a small box attached to the subject's belt (input impedance $= 10^9\ \Omega$, common mode rejection ratio > 120 db), then to a custom amplifier which converted the raw EMGs to derived RMS values (with a time constant of 55 ms) for the four channels of information (left and right erector spinae, and left and right rectus abdominus).

RESULTS

Figure 4 displays an example of measured vs model predicted results. Quantitative results follow from correlation, ANOVA, and residual analyses.

Correlation analysis of measured vs model predicted variables

Both foot force and trunk force variables were compared by forming a simple linear regression between measured and predicted values. Comparisons were made for the average values of the variables over both single and total right leg support phases of the gait cycle (see Table 2). Notice that the vertical foot force during single support had the slope parameter closest to unity. However, for total right leg support the predicted average vertical foot force did not correlate nearly as well with its measured counterpart. This implies that predictions during double support (that portion of the total right leg support period excluding single support) were not as valid as those during single support.

Another analysis of the relationships depicts the grand mean of the average value of the variable during single support by subject body weight category for each direction of exertion. The greatest discrepancy emerged for the horizontal GRFs during pushing, with consistent model overprediction. Model overprediction was also apparent for the vertical GRFs during pushing.

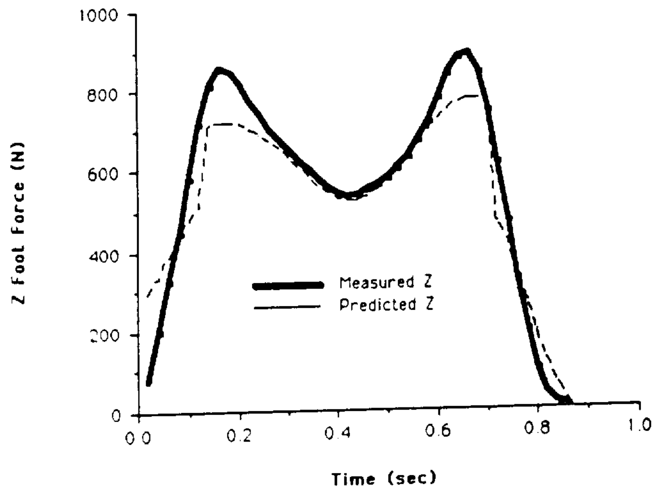
Another parameter selected for analysis was the maximum value of the variable within the time-window, because the direct comparison between measured and predicted values possible with the GRFs may have been more sensitive to changes in maximum values than to average values (see Table 2). Notice again that the slope of the relationship was closest to unity (0.7) for the vertical GRF during single support. Less of the variability in the data was explained by these regressions (using maximum values) compared with the average values shown earlier in Table 2.

Comparisons of the trunk muscle forces derived from RMS-EMGs with model predicted trunk muscle forces indicated that even less of the variability in the data was explained by the simple linear regression model. Most of these regressions were not significant at the 0.05 level. The following section will present

Table 2. Average and maximum GRF correlation results with linear regression coefficients

Variable	*R*-square	Intercept	Slope
	Average GRFs		
Single support			
RF_y	0.63	−10.9	0.59
RF_z	0.66	27	0.93
COF	0.1	0.15	0.1
Total support			
RF_y	0.67	−12.9	0.55
RF_z	0.12	357	0.26
COF	NS	—	—
	Maximum GRFs		
Single support			
RF_y	0.45	14.3	0.3
RF_z	0.45	164.4	0.78
COF	0.06	0.27	0.05
Total support			
RF_y	0.32	55.4	0.18
RF_z	0.21	435	0.42
COF	NS	—	—

RF_y = horizontal GRF, RF_z = vertical GRF, COF = ratio of RF_y to RF_z. Notice that the correlations during single support exceeded those from the total support phase, demonstrating the improved performance of the model during single support. Also note that the slope parameters were closest to unity during single support with the vertical GRFs.



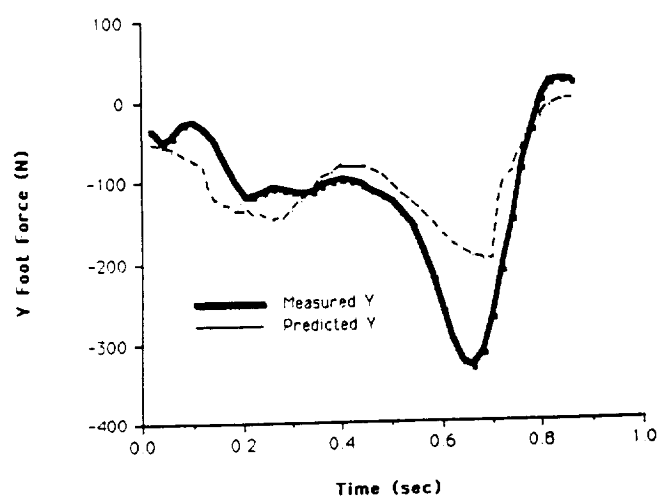


Fig. 4. Measured and predicted ground reaction forces for an example pulling trial. Notice the underprediction of the vertical GRF and the peak horizontal GRF for this particular trial with this subject. This discrepancy was more marked during double support (at either end of the right leg support GRF curves).

more detailed comparisons of derived and predicted trunk muscle forces by performing ANOVA analysis with subject and task factors included in the statistical models.

ANOVA analysis of model predicted variables

The biodynamic model predicted variables and the measured variables were analyzed with the inclusion of subject (SN, SX, HT, WT) and task (CR, WS, DR) factors. Predicted values exceeded measured values for both horizontal and vertical GRFs, whether for single or total right leg support (as seen by comparing grand means in Table 3). Horizontal GRFs were dependent always on direction of exertion, while all vertical GRFs were affected by individual subject factors. Only the mean measured and predicted vertical GRFs during the total right leg support phase had the same factors in the ANOVA analysis.

Predicted trunk muscle forces were subjected to the same analysis (see Table 3). The model underpredicted the *ESMFs*, but the predicted *RAMFs* fell between the derived values for the right and left side. Since a wide range of subject anthropometries were purposefully used in these experiments, similar analyses were performed separately for each of the three weight categories (see Table 4). One noticeable difference be-

Table 3. Average foot and trunk muscle force ANOVA summary

Variable	*R*-square	Mean	Subject factors	Task factors
		Foot forces		
Single support				
Meas. RF_y	0.84	−12.18	—	DR
Pred. RF_y	0.69	1.49	SN	DR
Meas. RF_z	0.66	582.15	SN, SX, HT	DR
Pred. RF_z	0.86	609.38	SN	—
Meas. *COF*	0.14	1.15	—	—
Pred. *COF*	0.14	0.29	—	—
Total support				
Meas. RF_y	0.88	−12.82	HT	WS, DR
Pred. RF_y	0.69	1.93	SN	DR
Meas. RF_z	0.8	490.71	SN, SX, HT	—
Pred. RF_z	0.7	518.93	SN, SX, HT	—
Meas. *COF*	0.18	0.76	—	—
Pred. *COF*	0.29	0.22	—	CR, WS
		Trunk forces		
Single support				
ESMF	0.39	561.62	SN	CR
FML	0.73	765.2	SN	DR
FMR	0.71	858.06	SN	DR
RAMF	0.32	71.01	SN	—
FRL	0.36	35.48	SN	—
FRR	0.51	88.59	SN	DR
Total support				
ESMF	0.4	511.23	SN	CR
FML	0.74	742.29	SN	DR
FMR	0.69	860.35	SN	DR
RAMF	0.28	85.79	SN	DR
FRL	0.36	32.94	SN	—
FRR	0.51	88.06	SN	DR

'Pred.' before the variable denotes the predicted value, and 'Meas.' denotes the measured value, RF_y = horizontal GRF, RF_z = vertical GRF, *COF* = ratio of RF_y to RF_z, *ESMF* = erector spinae muscle force predicted by the model, *FML* = erector spinae muscle force derived from the RMS–EMG of the left erector spinae muscle, *FMR* = erector spinae muscle force derived from the RMS-EMG of the right erector spinae muscle, *RAMF* = rectus abdominus muscle force predicted by the model, *FRL* = rectus abdominus muscle force derived from left rectus abdominus muscle, and *FRR* = rectus abdominus muscle force derived from right rectus abdominus muscle. Subject and task factors are explained in the text. Horizontal GRFs all depended on the direction of exertion, with differences in the subject factors involved. Vertical GRFs were affected by more subject factors. The trunk forces all depended on the individual subject factor; the only task factors of importance were the direction of exertion and the cart resistance. Notice that the variability in the measured variables was more readily explained by the selected subject and task factors than the variability in the predicted variables.

tween derived and predicted means was that direction of exertion was not a significant factor in the model predicted values. Model predicted mean *ESMF*s were less than their derived counterparts, most notably for the subjects in the average weight category. The model predicted that mean *RAMF*s exceeded the derived values for light and average weight categories, but fell between derived values for the heavy subjects.

Analysis of the differences between measured and model predicted variables

The residuals between measured and predicted horizontal GRFs were affected by subject, walking speed, and direction of resistance (SN, WS, and DR, as shown in Table 5). The positive residual means for both horizontal and vertical GRFs reaffirm the biomechanical model's bias towards overprediction. The vertical GRF residuals were affected by gender and height (SX and HT). The horizontal to vertical GRF ratio residuals were not explained by the linear statistical model. Average residual values were partitioned by subject body weight category for both single and total right leg support for the horizontal and the vertical GRFs. The horizontal GRF residuals were closest to zero for the subjects in the average weight category (WT = 2) when pulling. During pushing the residuals stayed positive across body weight. The vertical GRF residuals were minimized with the light subjects (WT = 1) for both pushing and pulling.

Table 4. Average trunk muscle force ANOVA summary partitioned by subject body weight category

Weight	Variable	R-square	Mean	Subject factors	Task factors
			Trunk extensors		
Single support					
Light	*ESM F*	0.37	504.5	SN	—
Light	*F MR*	0.86	760.4	SN	DR
Light	*F ML*	0.95	620.4	SN	DR
Average	*ESM F*	0.4	534.2	SN	—
Average	*F MR*	0.79	1141.2	SN	DR
Average	*F ML*	0.65	1038.9	SN	DR
Heavy	*ESM F*	0.38	622	SN	—
Heavy	*F MR*	0.69	810.9	SN	DR
Heavy	*F ML*	0.77	761.8	SN	WS, DR
Total support					
Light	*ESM F*	0.39	476.6	SN	OR
Light	*F MR*	0.87	748.2	SN	DR
Light	*F ML*	0.95	620	SN	DR
Average	*ESM F*	0.4	481.6	SN	—
Average	*F MR*	0.79	1151	SN	DR
Average	*F ML*	0.67	1011.2	SN	DR
Heavy	*ESM F*	0.39	561.3	SN	—
Heavy	*F MR*	0.67	821.8	SN	DR
Heavy	*F ML*	0.78	722.2	SN	WS, DR
			Trunk flexors		
Single support					
Light	*RAM F*	0.14	35.8	—	—
Light	*F RR*	0.52	3.8	SN	DR
Light	*F RL*	0.4	5.3	SN	DR
Average	*RAM F*	0.24	37	—	—
Average	*F RR*	0.53	19.7	SN	DR
Average	*F RL*	0.52	7	SN	DR
Heavy	*RAM F*	0.32	116	SN	—
Heavy	*F RR*	0.55	191.1	SN	DR
Heavy	*F RL*	0.37	73.8	SN	—
Total support					
Light	*RAM F*	0.13	39.1	—	—
Light	*F RR*	0.52	4	SN	DR
Light	*F RL*	0.42	5.9	SN	DR
Average	*RAM F*	0.31	37.7	SN	—
Average	*F RR*	0.51	17.8	SN	DR
Average	*F RL*	0.5	6.5	SN	DR
Heavy	*RAM F*	0.28	146.8	SN	—
Heavy	*F RR*	0.55	190.6	SN	DR
Heavy	*F RL*	0.37	67.6	SN	—

Note that the predicted trunk muscle forces (*ESM F* and *RAM F*) were not dependent on the direction of exertion, and that the model underpredicted the ESMF for the subjects of average weight.

Residuals were formed between model predicted trunk muscle forces and derived left and right trunk muscle forces for the total right leg support phase only (Table 5). All of the trunk force residuals were affected significantly by the individual subject factor (SN). Residuals between predicted extensor muscle force and derived muscle forces for each side were large and negative. However, the residuals for the trunk flexors were smaller. Right side *ESMF* residuals were closest to zero for the heavy subjects. However, on the left side the *ESMF* residuals were closest to zero for the subjects in the average weight category. Right and left side residual averages for *RAMF* resembled each other more closely, with minimum residuals for subjects in the average weight category performing pulls.

DISCUSSION

The novelty of this particular model is the capability to calculate dynamic parameters that relate to either back overexertion risk or foot slip risk. Previous investigations of pushing and pulling tasks used static analysis, and predictions of internal back forces or ground reaction forces were not attempted. Limitations of the present model will be discussed first in light of the major assumptions required. The effects of

Table 5. Average GRF and trunk muscle force residual ANOVA summaries

Variable	R-square	Mean	Subject factors	Task factors
		GRF residuals		
Single support				
RF_y	0.38	12.74	SN	WS, DR
RF_z	0.48	29.19	SX, HT	DR
COF	0.09	−0.34	—	—
Total support				
RF_y	0.39	13.15	SN	WS, DR
RF_z	0.6	31.62	SN, SX, HT	—
COF	0.09	−0.41	—	—
		Trunk force residuals		
Total support				
ESM F(L)	0.29	−2785.5	SN	—
ESM F(R)	0.87	−857.8	SN	DR
RAM F(L)	0.76	−13.1	SN	DR
RAM F(R)	0.75	−20.2	SN	DR

The positive means for the GRF residuals indicate the model's tendency to overpredict, although these average residuals are small. Note that residuals were larger for the trunk forces.

these assumptions on model performance will then be described. Finally, the validation approach used here will be compared with other validations of biomechanical models.

Model performance can not exceed the quality of the input to the model; hence the input to the model must be considered as a source of discrepancy between measured and predicted values. Sagittal plane joint center position measurement with the SELSPOT real-time position measurement system had two inherent problems: firstly, the location of joint centers was only grossly estimable with markers placed on the skin or clothing at an approximated joint center, as determined by palpation. This problem has been documented extensively before (see Zahedi *et al.*, 1987 for relevant arguments) and will not be probed here. Secondly, the use of a lateral photodetector caused distortion problems due to the lens, the detector, and signal reflection (Gustafsson and Lanshammar, 1977). The steps described in the data acquisition section compensated for these distortions.

Kinematic considerations played a large role in the determination of single vs total right leg support. Model GRF prediction differed depending on the type of support used; without the use of foot switches or a second force plate, the determination of left foot contact with the ground relied on a foot LED floor clearance which had a 1.0 cm tolerance and a Y coordinate velocity of $< 10\%$ of maximum Y foot velocity. Confounding this determination was the subject's tendency to drag the feet along the floor, perhaps to maximize proprioceptive cues, particularly when going backwards during a pull. Uncertainty in selecting double vs single support could cause discontinuities in GRF predictions in the transition region.

Several anthropometric variables were based on previous work done on limited samples of cadavers [link lengths and masses from Dempster (1955) and Clauser *et al.* (1969); radii of gyration from Chandler *et al.* (1975); diaphragm area from Morris *et al.* (1961) and Fisher (1967)]. The individual subjects performing the current experiments may not be described accurately by these proportionality constants and average areas, particularly because of the wide range of anthropometries selected. Detailed individual anthropometric measures would have been required to circumvent these innacuracies, but these were beyond the scope of this study.

The GRF calculations during double support came from moment equilibrium equations. The selection of the reference point for moments taken around the foot–ground interface was an estimate because the center of foot force application in the sagittal plane could not be determined without recording moments from the force platform. Therefore the proper location of the foot contact reference point for horizontal moment arms could have been anywhere between the heel and the toe marker (the model assumed the reference points were the heel marker for pushing and the toe marker for pulling). It was possible that inaccuracies in the horizontal moment arm estimates contributed to model prediction errors during double support.

The other assumption about double support was that the friction utilization was the same under both feet. Since only one force platform was available, this could not be directly verified. When overcoming cart inertia (or exerting push or pull forces maximally against a stationary object) most subjects concentrated their foot forces on one foot, using the other foot for balance (Chaffin *et al.* (1983); also based on observation of the volitional postures and foot forces used by subjects in the static calibration experiments). This concentrated the vertical forces on a smaller surface area, and thus increased the contact pressure at that particular shoe–floor interface. How these static exertions relate to the dynamic case remains unclear.

Since the linear regression coefficients derived during the static calibration experiments were used to transform the RMS-EMGs (obtained during the dynamic experiments) into derived muscle forces for comparison with predicted muscle forces, they played a role in evaluating model performance, and so are discussed here. Muscle activity, as described by EMGs, is highly variable. Processed EMGs (i.e. integrated, average, RMS) have been shown to describe more accurately levels of muscle force output than raw EMGs, particularly in isometric situations (see Redfern (1989) for a complete review of these arguments). It is also well known that surface EMGs are susceptible to signal decrements due to adipose tissue interposed between the sensor and the muscle tissue (Basmajian and DeLuca, 1985). The subjects in these experiments were not of uniform somatotype, and so the quality of surface EMG obtained from the heavy subjects was suspect. Another factor may have been the use of isometric exertions for the calibrations. Most previous work correlating RMS-EMGs with muscle force utilized discrete isometric force levels only (Basmajian and DeLuca, 1985), as opposed to our ramp of increasing force.

Vertical and horizontal GRFs were compared for both single and total right leg support because of the sensitivity of model predictions to the transition from one support phase to the other. The most rigorous validation of the model GRF predictions came from the single support phase, since none of the assumptions necessary to make the double support solutions determinate were required.

Measured average horizontal GRF values throughout the total right leg support phase were less than the model predictions. However, inspection of the residual analysis results revealed that larger intra-subject discrepancies existed when the double support phase was included, as opposed to those during single support.

Performing the same comparisons for the vertical GRFs, single support will be discussed first. The amount of variability explained by the regression analyses was similar for measured (66%) and predicted (86%) vertical GRFs. Considering the residuals formed by taking the difference between predicted and measured vertical GRFs during single support, the mean average residual value was 29.2 N, indicating that the model overpredicted more than it underpredicted.

During total right leg support, the mean average values were again greater for predicted than for measured vertical GRFs. The intra-subject comparison disclosed that the mean average force residual for the total right leg support phase was essentially the same as for the single support phase; this implied that the patterns of model over- or underprediction during double support were not consistent from subject to subject, and hence were counteracted when combined across subjects. The statistical consensus was that model predictions of GRFs during double support were not as valid as those during single support. This must be attributed to both the limitations of the moment arms used and the assumption of the equal friction utilization by both feet during double support.

Since the derived *ESMFs* came from transformations of measured RMS-EMG values with regression coefficients obtained from the static experiments, any limitations in the regressions will be reflected in the derived *ESMFs*. The ANOVA analysis of derived *ESMFs* indicated that anywhere from 68 to 74% of the variability in the data was explained by the independent main factors.

Considering the analysis of residuals formed by subtracting the derived *ESMFs* from the predicted *ESMFs*, the activity of the right side erector spinae group was predicted better than the activity of the left side muscle group. On an intra-subject basis, the predictions underestimated average *ESMFs* over the total right leg support phase. The previous discussion of surface EMGs taken from varied somatotypes is also germane, but, given all of these qualifiers, predicted *ESMF* performed qualitatively the same as right side derived *ESMF*. It should be stressed that the support phase used throughout this study occurred on the right foot, and therefore it was not surprising that right side derived *ESMFs* seemed more realistic than left side values; the left side muscles were involved to some degree in the swing phase of the left leg, which occurred during right leg support.

The analysis by subject body weight category provided better insight into the indirect validation of predicted muscle forces compared with muscle forces derived from dynamic RMS-EMGs. Perfect correspondence of derived and predicted values would yield residuals equal to zero. In most cases the *ESMF* residuals were closest to zero for the medium weight subjects. Adding the consideration that the erector spinae are trunk extensors, they were expected to be most active during pulling. Indeed, in most cases, the residual parameters were closer to zero for the pulling tasks in this investigation.

Derived and predicted *RAMFs* were examined in a similar manner. Based on the argument that the left side of the muscle group may have some involvement as a synergist during left leg swing, only the right side rectus abdominus muscle group will be discussed here. Minimal anterior muscle activity was seen during the pulling tasks. As was the case with the predicted erector spinae forces, more subject factors (SX, HT, WT) were present in the ANOVA results than for the derived forces in the anterior muscles. It should be stressed that the linear regressions derived from the static calibration experiments did not, in general, explain as much of the variability in the RMS-EMGs of the rectus abdominus as for the erector spinae. This was attributed partly to the variation in adipose tissue distribution in the anterior part of the trunk as opposed to the posterior low back region.

The *RAMF* residual parameters were closest to zero for either the light or the average weight subjects. The rectus abdominus was most active during pushing; i.e. the residual averages during pushing were low. However, residual maxima (all positive) during pushing increased as weight increased, reflecting an increasing model overprediction with heavy subjects.

Lee (1982) also reported much better model performance in predicting GRFs than EMGs. He found that predicted vertical GRFs correlated with measured vertical GRFs ($r^2 = 0.65$) and predicted horizontal GRFs correlated with measured horizontal GRFs ($r^2 = 0.56$). Mean errors were small (60 and 19 N, respectively). This performance was similar to that found in the current investigation. Lee found that there was a subject effect in the differences between predicted and measured GRFs. This was again found in the current investigation in the residual analysis.

The validation approach in this investigation was similar to Lee (1982), with the exception that he used the regression relationships between isometric RMS-EMGs and exerted trunk forces to transform predicted dynamic trunk muscle forces to predicted dynamic RMS-EMGs, which were then compared with the measured dynamic RMS-EMGs. Since there have been no other dynamic biomechanical model analyses of pushing and pulling, related validation techniques can only be found in different tasks. Static predictions of mean spine compression acting at L3 have been correlated with mean intradiscal pressure measurements (Schultz and Andersson, 1981), with a correlation coefficient of 0.91. This same group later developed a model to predict trunk muscle forces during isometric weight-holding and force resistance tasks (Schultz *et al.*, 1982), which they validated by comparing their calculated trunk muscle tensions with mean myoelectric signal levels. Correlation coefficients ranged from 0.34 to 0.92, depending on the muscle group and on the function used to predict muscle force. More germane to the current investigation were predictions of dynamic trunk loading; Jager and Luttman (1989) developed a dynamic 19-segment model to assess lumbar stress during load lifting. Their validation consisted of comparing their model calculations with intradiscal pressure measurements taken from the literature; only static holds were compared.

Direct comparisons of predicted and measured ground reaction forces have been performed by others (Pandy and Berme, 1988, 1989). These studies simulated GRFs by assuming joint moment trajectories and performing open-loop (single support) or closed-loop (double support) analyses during normal or pathological walking. However, cart pushing or pulling tasks have not been similarly examined.

Previous biomechanical model validation approaches have not systematically varied subject anthropometry and gender. The evidence from this investigation suggests that this particular dynamic biomechanical model was valid when a wide range of anthropometries was studied. Model predictions, including both GRFs and trunk muscle forces, were better during single support phases than during double support phases.

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