Changes in the Surface EMG Signal and the Biomechanics of Motion During a Repetitive Lifting Task

Paolo Bonato, *Member, IEEE*, Patrick Boissy, Ugo Della Croce, and Serge H. Roy

***Abstract*—The analysis of surface electromyographic (EMG) data recorded from the muscles of the back during isometric constant-force contractions has been a useful tool for assessing muscle deficits in patients with lower back pain (LBP). Until recently, extending the technique to dynamic tasks, such as lifting, has not been possible due to the nonstationarity of the EMG signals. Recent developments in time-frequency analysis procedures to compute the instantaneous median frequency (IMDF) were utilized in this study to overcome these limitations. Healthy control subjects with no history of LBP ($n = 9$; mean age 26.3 ± 6.7) were instrumented for acquisition of surface EMG data from six electrodes on the thoraco-lumbar region and whole-body kinematic data from a stereo-photogrammetric system. Data were recorded during a standardized repetitive lifting task (load $= 15\%$ body mass; 12 lifts/min; 5-min duration). The task resulted in significant decreases in IMDF for six of the nine subjects, with a symmetrical pattern of fatigue among contralateral muscles and greater decrements in the lower lumbar region. For those subjects with a significant decrease in IMDF, a lower limb and/or upper limb biomechanical adaptation to fatigue was observed during the task. Increases in the peak box acceleration were documented. In two subjects, the acceleration doubled its value from the beginning to the end of the exercise, which lead to a significant increase in the torque at L4/L5. This observation suggests an association between muscle fatigue at the lumbar region and the way the subject manipulates the box during the exercise. Fatigue-related biomechanical adaptations are discussed as a possible supplement to functional capacity assessments among patients with LBP.**

***Index Terms*—Biomechanics, electromyography, fatigue, lifting, lower back pain, time-frequency.**

I. Introduction

LOWER back pain (LBP) specialists in the clinical and ergonomic sciences are in need of effective methods to objectively quantify paraspinal muscle strength and endurance deficits for optimizing the delivery of LBP rehabilitation. Surface electromyographic (EMG) techniques have played an important role in helping researchers understand normal functioning of concurrently active trunk muscles when specific static postures or movements against gravity take place. These procedures have been successfully applied to document the alteration to normal paraspinal muscle functioning associated with chronic or acute LBP [9], [20], [25], [30]. Laboratory and clinically based studies by our group [29], [30] and others [9], [20] have shown that by simultaneously monitoring changes in spectral estimates from multiple surface EMG electrode sites, it is possible to evaluate the relative contribution of individual paraspinal muscle groups during a sustained extension of the trunk. Because muscle strength and endurance deficits are common consequences of injury, pain, and disuse, paraspinal muscles compensate for these deficits, resulting in a relative alteration in their EMG spectral activity during induced localized muscle fatigue [30].

The alteration in EMG spectral activity has been well described for static tasks involving sustained trunk extension [30]. However, only a few studies have begun similar investigations for dynamic tasks, such as repetitive lifting [32], [34], [35]. A primary limitation to developing effective EMG analysis procedures for such dynamic tasks is that the traditional quantification procedures for spectral analysis assume that the signal is *wide-sense stationary;* i.e., the first and second statistical moments of the stochastic process do not change in time. This precondition can be satisfied by recording surface EMG signals during highly constrained isometric conditions in which there are negligible changes in muscle length, muscle force, or electrode position during the fatigue-inducing activity. The limitation to constant-force isometric contractions seriously compromises its clinical usefulness because many dynamic activities, such as lifting and load carrying, are commonly associated with LBP injury [22]. However, recent developments in the field of signal processing offer a mechanism to overcome the limitation of current EMG procedures to static conditions by the application of time-frequency transforms [2]. Our group [1], [2], [4], [17], [32] and others [15], [34], [35] have been recently investigating various time-frequency techniques to extend the assessment of muscle function to dynamic conditions. Cohen Class and Cohen–Posch Class transforms have been shown to be particularly well suited for computing fatigue-related changes in the frequency content of surface EMG signals recorded during dynamic repetitive tasks [1], [2], [4], [15], [17], [32]. We introduced a spectral parameter, the *instantaneous median frequency* (IMDF), in previous work [1], [4], [17], [32] based on these transformations. It was derived by replacing the power density spectrum with the EMG time-frequency representation in the

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equation defining the median frequency. We have shown that the IMDF may be used to assess localized fatigue during dynamic contractions [1], [4], [17], [32].

The EMG signal-processing procedure has evolved to a point where it can be tested under "real-world" conditions in the field of rehabilitation. One application, and the purpose of this paper, is to develop a method for augmenting current functional capacity evaluation (FCE) procedures for patients with LBP. FCEs are aimed at assessing an individual's ability to perform the roles, activities, and physical demands required in relation to a given work or activity context [16]. FCEs are increasingly used to determine program objectives and intervention strategies for the occupational rehabilitation of LBP subjects [16]. While this approach has merit at the disability level, it does not identify specific muscle strength and endurance deficits in LBP subjects associated with functional tasks.

Current FCE approaches to documenting impairment rely on qualitative assessments of patients during the performance of standardized simulated tasks [16]. Such features as the smoothness, effort, and range of movement are graded using qualitative descriptors or ordinal scales. In the study reported in this paper, we have selected a commonly used work assessment task, a standardized lift from midshank to waist height, and augmented the task with measurements of surface EMG signals from superficial paraspinal muscles. These measurements may provide quantitative objective information to characterize the fatigue-related changes that normally occur among these muscle groups as a result of the task. In addition, when extending the investigation of muscle function to dynamic tasks, the biomechanics of motion plays a crucial role. Compensatory mechanisms might occur as a product of the progression of fatigue during the task [5], [8], [10], [28], [36], [37]. Such changes in the biomechanics of motion may be interpreted as "strategies," i.e., ways to facilitate the task when it becomes more challenging to the subject because of the progression of fatigue [5], [27]. There are, therefore, two types of fatigue-related changes that may occur during a dynamic task: changes in the surface EMG characteristics and changes in the biomechanics of motion. This paper measures both and describes the association between the two in a sample population of healthy subjects with no history of LBP. The findings are discussed in terms of their relevance to augmenting current assessment procedures for functional capacity evaluation among patients with LBP.

II. Material and Methods

A. Experimental Protocol

Nine male subjects with no previous history of musculoskeletal impairment participated in this study. All subjects gave their informed consent before participating in this study, following procedures approved by the Institutional Review Board of Boston University. Subjects were recruited from a healthy young population (mean age of 26.3 ± 6.7) and were generally fit (mean body mass index of 21.7 ± 1.9). Each subject performed a 5-min lifting task under standardized conditions (Fig. 1). The task consisted of a cyclical lifting and lowering of a weighted box ($29 \times 25 \times 23$ cm) from shelves positioned at midshank and waist height, respectively, at a rate of 12 lifting cycles/min. The subjects were instructed to adopt a "free-lifting" method of performing the task in the sagittal plane. A brief training period was provided prior to the experiment to practice performing the task avoiding twisting of the trunk or erratic movements. The subjects were also trained to comply with the imposed work rate by following a metronome, which was used during the experiment. The weight of the box was adjusted individually to represent 15% of the subject's body mass. The average weight lifted was 10.5 ± 2.1 Kg and ranged from 8.6 to 12.3 Kg.

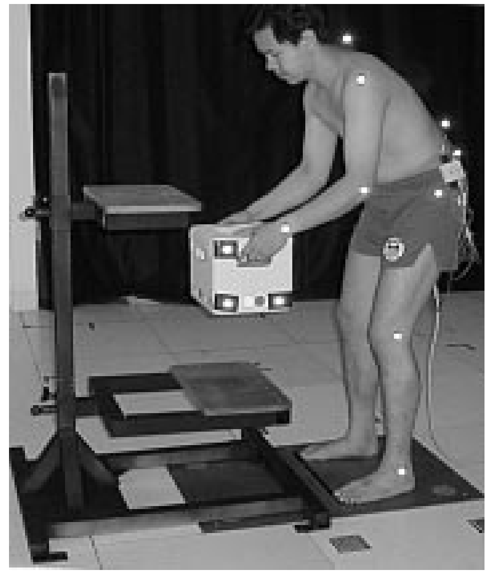

Fig. 1. Experimental setup. The subject is instrumented with passive reflective markers and surface EMG electrodes. Marker positions are tracked using a two-camera stereo-photogrammetric system (ELITE, B.T.S., Milan, Italy). The experimental task consists of a 5-min cyclical free-lifting task under standardized conditions. Work cycle is set at 12 lifts/min, and a metronome provides the cadence necessary to comply with the imposed work rate.

B. SEMG and Biomechanical Measures

Surface EMG (SEMG) data from three bilateral back muscles were recorded during the lifting task using single differential electrodes (common-mode rejection ratio $>$ 80 dB, input impedance > 10 E $+ 12\ \Omega/5$ pF; Model #DE-02, DelSys Inc., Boston MA) with a detection geometry consisting of parallel silver bars ($1 \times 1 \times 10$ mm) separated by 10 mm interelectrode distance. The electrodes were secured on the skin using a custom-made double-sided interface and positioned at the T10 vertebral level on the iliocostalis lumborum, at the L1 vertebral level on the longissimus thoracis, and at the L5 vertebral level on the multifidus muscles. Muscle sites were identified by palpation, and electrodes were positioned approximately 2 cm from the midline of the vertebral column. Signals were conditioned by a custom-made isolated amplifier (gain of 3000, bandwidth 20–450 Hz with 12 dB/oct rolloff and input noise $< 1.25\ \mu$V root mean square) and digitally sampled at a rate of 1024 Hz using a PC workstation with a 12-bit analog-to-digital board (MicroStar 2400). Surface EMG data were acquired for a 30-s time interval at the beginning (baseline) and the end (5 min) of the repetitive lifting task.

During the 30-s time intervals of EMG acquisition, the biomechanics of motion was monitored by using a two-camera stereo-photogrammetric system (ELITE, B.T.S., Milan, Italy). Prior to the lifting task, a volume of $1.5 \times 2.1 \times 1.2$ m was calibrated using a grid of 6 by 8 passive reflective markers that was positioned at five different distances from the cameras. Stereo-photogrammetric data were acquired at a sampling rate of 100 frames/s. Passive reflective markers were positioned on one side of the body (Fig. 1) (lateral malleolus, lateral epicondyle, greater trochanter, S2, L4, C7, dorsal side of ulnar styloid, elbow lateral epicondyle, and shoulder acromion process). Additional markers were put on one side of the box to track its trajectory.

C. Analysis

1) Surface EMG Data: When surface EMG signals are recorded during a dynamic contraction, such as lifting and lowering a box, changes in muscle force, muscle length, and position of the electrodes with respect to the active muscle fibers cause a continuous and rapid change in the EMG frequency content [32]. Concurrently, a more slowly evolving nonstationarity in the EMG signal is present as a progressive compression toward lower frequencies caused by the fatigue process [4], [17], [32].

Because of these nonstationarities in the EMG signal, the frequency content of the signal was derived using a time-frequency transformation [4], [17], [32]. A variety of time-frequency transformations have been utilized previously for similar applications [2], [15], [34], [35], and Cohen Class transformations have been favored [4], [15], [17], [32]. However, we recently demonstrated that further improvements in the time-frequency spectral estimates could be obtained using Cohen–Posch representations [1]. These distributions are preferable because they lead to a more stable estimate of the time-frequency parameter (the IMDF) utilized to assess muscle fatigue [1].

In this study, Cohen–Posch distributions [18] were derived from Choi–Williams time-frequency representations [6] by utilizing an iterative algorithm that adjusts the distribution in order to satisfy time and frequency marginals [1], [18]. Time and frequency marginals are obtained by integrating the time-frequency distribution along the frequency axis and the time axis, respectively. The time marginal of the Cohen-–Posch distribution is equal to the instantaneous energy of the analyzed signal. The frequency marginal is equal to the square of the magnitude of the Fourier transform of the input signal. Satisfying time and frequency marginals allows the time-frequency distribution to carry physical meaning. EMG time-frequency representations were averaged over finite time intervals to maximize the stability of the IMDF estimates. This technique has been fully characterized in previous work by our group [1], [4]. According to this approach, $DS(i,j)$ ($1 < i < N$ and $1 < j > M$), the discrete-time discrete-frequency Cohen–Posch time-frequency spectrum, may be replaced in the time intervals $((k-1)P+1, kP)$, $1 \leq k \leq N/P$, by

$$DS(k,j) = \frac{1}{P}\sum_{l=1}^{P} DS(l+(k-1)P, j). \tag{1}$$

Once the Cohen–Posch time-frequency representation of the EMG signal was obtained, the IMDF was computed by replacing the power density spectrum with the Cohen–Posch time-frequency representation in the formula defining the median frequency [1]. The integration of the time-frequency representation was upper bounded by using the upper frequency [4] of the EMG signal in order to maximize robustness of the IMDF estimation to changes in the EMG signal-to-noise ratio [4]. Therefore, the formula utilized for the estimation was

$$\sum_{j=1}^{\mathrm{IMDF}(k)} DS(k,j) = \sum_{j=\mathrm{IMDF}(k)}^{\mathrm{UF}(k)} DS(k,j) \tag{2}$$

where $\mathrm{IMDF}(k)$ indicates the instantaneous median frequency, $\mathrm{UF}(k)$ denotes the upper frequency, $DS(k,j)$ is the time-averaged estimate of the Cohen–Posch time-frequency representation of the surface EMG signal, and k and j are the time and the frequency indexes, respectively.

The use of these transformations allowed us to obtain the representation of the within-cycle nonstationarities (Fig. 2).

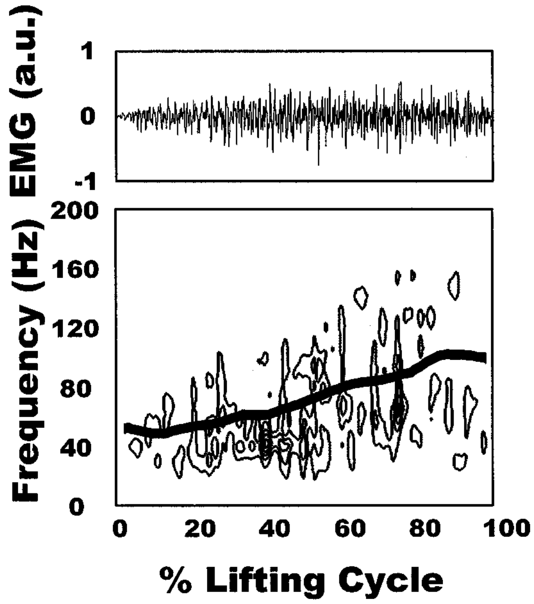

Fig. 2. Illustration of nonstationary behavior of EMG data collected during a lifting cycle, i.e., the time-interval during which the box was moved from the lower to the upper shelf. The upper plot shows the time course of the surface EMG signal recorded from the right multifidus muscle at L5. The lower plot (contour curves) shows the corresponding smoothed time-frequency distribution (Cohen–Posch) of the EMG burst, and the averaged ($n = 6$ cycles) IMDF is superimposed. Note the dramatic frequency modulation of the EMG data associated with the within-cycle nonstationarities.

Fig. 2 demonstrates the nonstationary behavior that marks the EMG data within a single lifting cycle. The upper plot shows the surface EMG signal recorded from the right multifidus muscle at L5. The lower plot represents the corresponding time-frequency representation together with the IMDF time course within the lifting cycle. It is worth noticing that the analysis was performed for the lifting portion of the task, which consisted of lifting and lowering the weighted box. Therefore, lifting cycle in Fig. 2 refers to the time interval corresponding to the movement of the box from the lower to the upper shelf. For the sake of clarity, the Cohen–Posch time-frequency representation is smoothed using a two-dimensional rectangular window of 30 ms and 10 Hz. This smoothed distribution clearly shows the dramatic change in the EMG frequency content within the lifting cycle. Smoothing over frequency was not necessary when computing the IMDF. Also, for the sake of clarity of the figure, the IMDF represented in Fig. 2 was obtained as the parameter time course within the lifting cycle averaged over six cycles. The dramatic frequency modulation illustrated in this figure may be related to changes in the force exerted by the muscle under study, modifications of the muscle fiber length, and the displacement between the surface EMG electrodes and the active muscle fibers. Specifically, a change in the thickness of the fat tissue between the electrodes and the muscle fibers occurs during the trunk movement, which is likely to be the main factor contributing to the frequency modulation of the EMG signal within each lifting cycle [3].

When considering a repetitive dynamic task, nonstationarities evolve similarly across lifting cycles because the biomechanics of the lifting task is quasi-periodic. In addition, an overall trend in the IMDF decay is apparent, which relates to slowly evolving nonstationarities caused by the fatigue phenomenon. We therefore selected time intervals of the EMG data corresponding to the same portion of the lifting cycle across successive cycles [32]. Because the same portion of the cycle was considered across successive cycles, the IMDF values obtained from each lifting cycle were equally affected by the within-cycle nonstationarities. In other words, this approach allows one to derive a time course of the IMDF parameter that is relatively independent of the EMG modulation related to the biomechanics of the task and thus closely related to the fatigue process [4].

Because this technique is based on the assumption that the biomechanics of the task for the chosen portion of the cycle is the same across cycles, it is desirable to utilize a portion marked by high repeatability of the biomechanics. Previous results obtained under controlled biomechanical conditions indicate that these portions are also marked by low variability of the IMDF [4]. This is because there are two possible sources of changes in the IMDF parameter during the task, i.e., muscle fatigue and the variability of the biomechanics of motion. It follows that the portion of the lifting cycle associated with minimum variability (across cycles) of the IMDF parameter is the one minimally affected by factors that relate to the biomechanics of motion, and thus the IMDF time course derived using such portion reflects muscle fatigue during the lifting task. Therefore, the following procedure was implemented.

1) Compute the IMDF for all the lifting cycles.
2) Resample the IMDF time series in order to have the same number of estimates within each cycle.
3) Divide the IMDF time series into time intervals (i.e., portions) lasting 1% of the lifting cycle and compute the standard deviation of the IMDF associated with each portion of the cycle.
4) Choose the portion of the cycle associated with minimum variability for deriving the time course of the parameter during the task.

This procedure was applied to compute IMDF values for the surface EMG signals recorded from the three contralateral muscle sites. Values at baseline (i.e., data from the first six lifting cycles) and at 5 min (i.e., data from the last six lifting cycles) of the task were derived for each muscle site. The IMDF values for each muscle were normalized with respect to the average baseline value for that muscle. To compare the EMG results with the biomechanical parameters derived in the sagittal plane, normalized IMDF values for contralateral muscles were pooled and averaged.

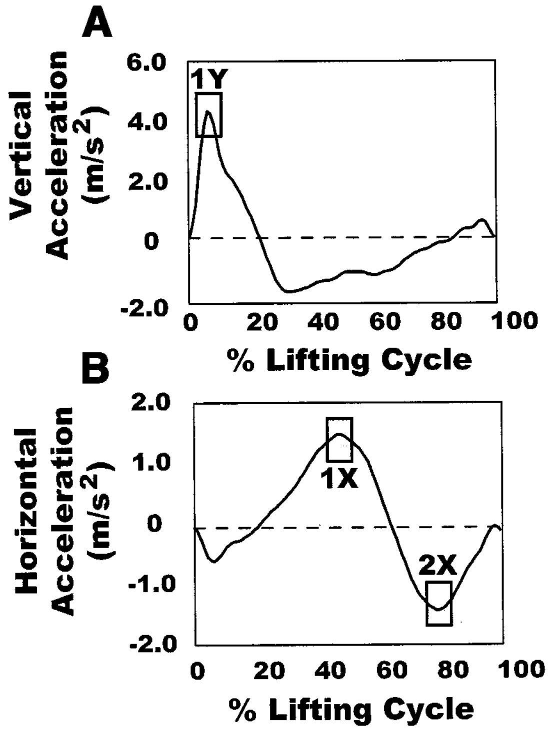

Fig. 3. Illustration of the functional portions of the lifting cycle as identified by the occurrence of the minima and maxima of the box acceleration. The plots represent mean acceleration curves from six lifting cycles for the (A) vertical and (B) horizontal components. Peak values corresponding to functional portions of the lifting motion when the subject 1) lifts the box from the lower shelf (1Y), 2) pushes the box toward the upper shelf (1X), and 3) prepares to deposit the box on the upper shelf (2X) are outlined by shaded areas.

2) Biomechanics: The trajectories of the reflective marker were filtered using an adaptive low-pass filter [7] to attenuate high-frequency noise components. The height of the box, computed from the markers positioned on the box, was then used to estimate box trajectory (i.e., position), velocity, and acceleration (vertical and horizontal components). The beginning and end of each lifting cycle were identified using the vertical component of the box trajectory and the vertical component of the box acceleration. Thresholds were set on the box height based on the position of the lower and upper shelves. A threshold was also set on the vertical component of the box acceleration based on the background noise observed from the acceleration data when the box was left on the lower and upper shelves. The beginning and end of the lifting cycle were detected as the time instants when the box position reached the height corresponding to the position of the lower and upper shelves, respectively, and the box acceleration was below the above-defined threshold.

From the vertical and horizontal components of the box acceleration, it is apparent that minima and maxima of the box acceleration in the sagittal plane mark each lifting cycle. The vertical and horizontal components are shown in Fig. 3. Three events are marked by boxes:

1) when the acceleration of the box in the vertical direction reaches its maximum (1Y);

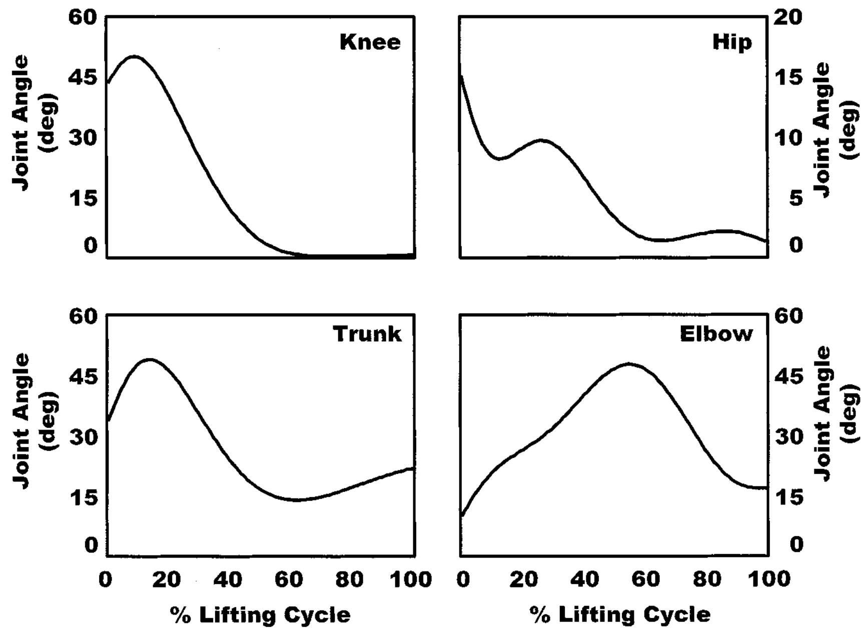

Fig. 4. Average angular displacements (positive values indicate flexion) from the beginning to the end of the lifting cycle at the knee, hip, trunk, and elbow for one subject. Values are normalized to a baseline condition taken before the start of the exercise. (i.e., the subject is holding the box on the upper shelf with knees and elbows fully extended with trunk in the upright position). Graphs represent changes in absolute joint angles computed from six lifting cycles at the beginning of the lifting exercise (first 30 s of the task).

2) when the acceleration of the box in the horizontal direction reaches its maximum (1X);
3) when the acceleration of the box in the horizontal direction reaches its second minimum (2X).

These three events correspond to functional portions of the lifting cycle when the subject 1) lifts the box from the lower shelf (1Y), 2) pushes the box toward the upper shelf (1X), and 3) prepares to deposit the box on the upper shelf (2X).

A model of the human body was defined based on the position of the passive reflective markers, according to the method proposed by Frigo [11]. For each recorded lifting cycle, we derived the angular displacements at the knee, hip, trunk, and elbow (Fig. 4). For each joint (hip, knee, and elbow), angles were computed from the trajectory of the relevant joint marker and distal and proximal joint markers. The trunk was modeled using the segment from S2 to C7, and the angle with respect to the vertical line in the sagittal plane was computed. From each estimate of angular displacement, total range of motion during the lifting cycle and the joint angles at 1Y, 1X, and 2X (i.e., peak occurrences in the acceleration components) were computed. These parameters were used to test the hypothesis that the kinematics of motion changed during the fatiguing exercise. In addition, in order to investigate whether the box is handled in a different way when fatigue is present during the dynamic task, we monitored maximum values in the horizontal and vertical components of the box acceleration. Finally, classical inverse dynamics of the two-dimensional model was computed from stereo-photogrammetric and anthropometrical data to estimate joint torque at L4/L5 [11].

D. Data Reduction and Statistical Analysis

The EMG and biomechanical data described in the analysis section were extracted from two 30-s epochs (one at the beginning of the exercise and one after 5 min of exercise). Each epoch consisted of six lifting cycles. The IMDF parameter values for each muscle were extracted from individual lifting cycles within each epoch. The IMDF values of each muscle for the baseline epoch ($n = 6$ values) and the 5-min epochs ($n = 6$ values) were normalized with respect to the average IMDF value computed from the six lifting cycles at baseline (first epoch) for that muscle. Normalized IMDF values for contralateral muscles were pooled (i.e., left and right L5 muscle sites) and averaged to allow comparison with the biomechanical parameters of sagittal plane movements. Total range of motion at the knee, hip, trunk, and elbow, as well as peak maximum values for the horizontal and vertical components of the box acceleration and joint torque at L4/L5, were extracted for each lifting cycle during the baseline epoch ($n = 6$ values) and at the 5-min epoch ($n = 6$ values). The EMG and biomechanical data were compared for each individual subject using Wilcoxon matched pairs (nonparametric) tests to identify significant changes during the lifting exercise. Statistical significance was set at $p < 0.05$. Summary tables were derived for the results obtained from the Wilcoxon tests on each subject. Associations were inferred based on the significant changes in the EMG and biomechanical parameters on a subject-by-subject basis.

III. Results

An example of the results of the EMG analysis is illustrated in Fig. 5. The average and standard deviation values derived from six cycles at the beginning (baseline) and the end (at 5 min) of the exercise are shown. The three plots on the left side represent the results obtained from the electrodes located at different spinal levels on the left side, while the three plots on the right side are related to the corresponding contralateral muscle sites.

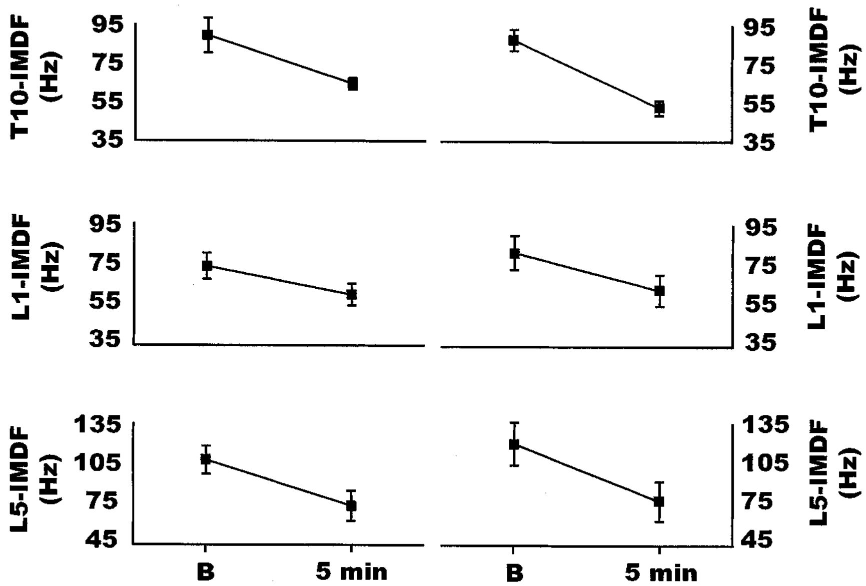

Fig. 5. Changes in IMDF between baseline (B) and minute five (5 min) of the exercise for one subject. Values represent average and standard deviation of IMDF estimates derived from six cycles for the multifidus muscle at L5, the longissimus thoracis muscle at L1, and the iliocostalis lumborum muscle at T10. Plots on the left side represent the results obtained from the electrodes located at different spinal levels on the left side, while plots on the right side are related to the corresponding contralateral muscle sites.

TABLE I
DIFFERENCES BETWEEN INSTANTANEOUS MEDIAN FREQUENCIES COMPUTED FROM SEMG RECORDED BILATERALLY AT THE BASELINE CONDITION AND AFTER 5 MIN OF CYCLICAL LIFTING FROM PARASPINAL MUSCLES AT L5, L1, AND T10. VALUES IN BOLD ARE SIGNIFICANT AT THE 0.05 LEVEL (WILCOXON TEST). "NS" INDICATES THAT CHANGES WERE NOT STATISTICALLY SIGNIFICANT

EMG	*S1*	*S2*	*S3*	*S4*	*S5*	*S6*	*S7*	*S8*	*S9*
L5	NS	***22.5***	***20.9***	***31.7***	***22.1***	***18.1***	NS	NS	NS
L1	***13.8***	NS	NS	NS	NS	***27.6***	NS	NS	NS
T10	***15.0***	NS	NS	18.3	NS	NS	NS	NS	NS

Fig. 5 indicates that contralateral muscle sites show a fairly symmetric (right versus left) frequency drop during the repetitive lifting task. Also, the magnitude of the IMDF decrease from baseline to minute five is relatively high, particularly at L5.

A summary of the results obtained for all the subjects recruited in this study is presented in Table I. Subject ($n = 9$) results are presented for normalized IMDF values of different spinal levels (averaged for controlateral sites). When significant differences were found between the baseline condition and the results at 5 min of the exercise (Wilcoxon test, $p < 0.05$), the percentage drop was computed and reported in Table I. Significant changes are shown in six subjects, five of whom demonstrated significant decreases in normalized IMDF at L5. In four of these subjects, the largest decrease in IMDF resulted from data recorded at L5. Significant changes in IMDF are also shown for data derived from L1 and T10, but the findings are less consistent and the magnitude of the change is generally smaller than the results corresponding to L5.

Significant changes were also present in the biomechanics of the lifting task. Figs. 6 and 7 present an example of the changes observed in the biomechanics of motion between baseline and minute five of the exercise. Data are presented for one subject. Fig. 6 shows changes in the range of motion at the knee, hip, trunk, and elbow. The magnitude of these changes is greatest at the knee and elbow joints, while changes of a smaller magnitude

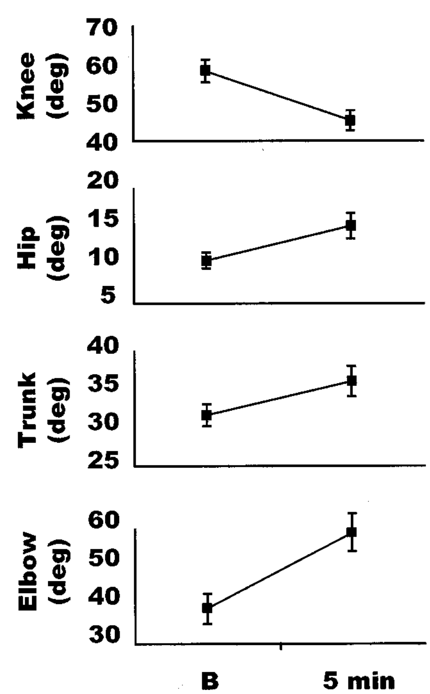

Fig. 6. Changes in the range of angular displacements observed at the knee, hip, trunk, and elbow between the baseline condition (B) and the end of exercise (5 min) for one subject. Graphs represent changes in the range of motion derived from six cycles. Values shown are mean and standard deviation.

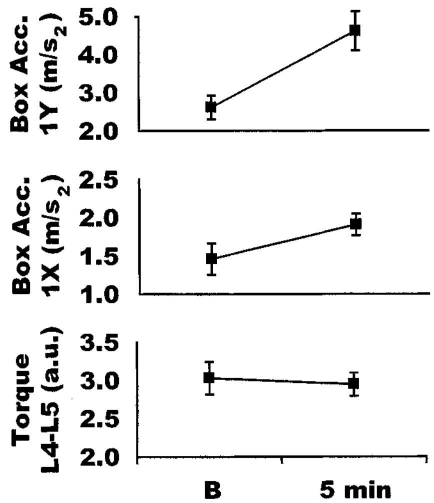

Fig. 7. Changes in peak box accelerations (upper and middle plots) and joint torque at L4–L5 (lower plot) between baseline (B) and the end of exercise (5 min) for one subject. Values represent mean and standard deviation of peak box acceleration for the vertical (upper plot) and horizontal (middle plot) components and of the torque at L4–L5 (lower plot). Values are derived from six lifting cycles. Torque values are normalized by the subject's weight and height.

are present at the hip and trunk. Fig. 7 shows that similar changes were observed in the maximum value of the acceleration in both the horizontal and vertical components. In this specific subject, however, these changes were not associated with a significant change in the maximum torque at L4/L5.

Table II summarizes the biomechanical results for all subjects. Changes of significant magnitude were shown at the knee and/or the elbow in the first six subjects (S1–S6). The last three subjects either did not show any significant change at these two joints or the magnitude of the increased range of movement was negligible (see the small increase—2.8°—in knee flexion/extension for subject S7). Changes were also observed at hip and trunk, but we focus here on knee and elbow angular displacements because these changes may be linked to fatigue-related strategies. Such strategies and motivations to emphasize changes at knee and elbow are discussed in the following section.

Among those subjects whose data indicated a significant change in the range of motion at either the knee or the elbow, different biomechanical adaptations to fatigue were identified. The first two subjects showed a significant increase in the knee flexion/extension range of motion. These changes may be seen as a *lower limb biomechanical adaptation* to fatigue. Changes in the knee range of motion in these two subjects reflect an increased knee flexion during the first portion of the lifting cycle, i.e., when the vertical component of the box acceleration reaches its maximum. Specifically in subject S1, an increased knee flexion of 8.6° was observed at the occurrence of the maximum vertical acceleration of the box, while in subject S2, this increase was as large as 15.5° (both increases were shown to be significant at $p < 0.05$ by a Wilcoxon test comparing baseline and minute five of the exercise).

In addition to an increased knee range of motion, the second subject (S2) showed a significant increase in the elbow range of motion. The same observation was made in subjects S3, S4, and S5. The increased elbow range of motion resulted from an increased elbow flexion following the first portion of the lifting cycle marked by a peak in the vertical component of the box acceleration, when the subject pulls the box from the lower shelf and brings it to the waist level. The maximum elbow flexion occurs in preparation to the second portion of the cycle when the horizontal component of the box acceleration reaches its maximum. Of interest is the unexpected finding that these subjects did not show a significant difference in elbow flexion at the occurrence of the first maximum in the horizontal component of the box acceleration. This is likely because the box is accelerated slowly in the horizontal direction and thus the maximum acceleration is reached once the elbow has been already partially extended. Changes in the elbow range of motion may be seen as an *upper limb biomechanical adaptation* to fatigue.

A significant reduction in the range of motion at the end of the lifting task is shown for the knee in subject S4 and for the elbow in subject S6. The first is due to a decreased knee extension in the last portion of the lifting cycle. For this subject, a significant increase of 10.1° in knee flexion was detected at the occurrence of the second minimum in the horizontal component of the box acceleration, when the subject prepared to position the box on the upper shelf. In subject S6 the decrease in elbow range of motion (7.6° decrease) reflects both a decreased maximum flexion (8.9° decrease) and an increased minimum flexion (1.3° increase), i.e., reduced elbow extension at the time the subject picked up the box from the lower shelf.

Similar changes are shown for the comparison of the maximum values of the two components of the box acceleration at baseline and at minute five of the repetitive lifting task. Table II shows a dramatic increase in the vertical component for subject S2 and in the horizontal component for subject S6. For these two subjects, the acceleration in at least one of the components doubles its value from the beginning to the end of the exercise. Also, for these two subjects, the peak torque at L4/L5 shows a significant increase. For subject S2, the increase in torque corresponds to approximately 13% of the value at baseline, while for subject S6, the increase is approximately 12%. The only other subject that showed a significant change in the maximum torque at L4/L5 is subject S5. However, for this subject, a decrease in the maximum torque at L4/L5 was demonstrated. Coincidentally, this subject is the one that showed the more dramatic *upper limb biomechanical adaptation* to fatigue, as the increase in elbow range of motion is the largest observed among all the subjects.

A comparison of Tables I and II indicates that upper and/or lower limb biomechanical adaptations to fatigue were observed for all the subjects (S1–S6) whose data demonstrated a significant IMDF drop at some of the muscle sites monitored during the repetitive lifting task. In contrast, those subjects (S7–S9) whose data showed negligible or no significant changes in the knee and elbow range of motion did not demonstrate any significant drop in the IMDF. This result indicates that there was an association between fatigue-related changes in the frequency content of the surface EMG data from back muscles and biomechanical adaptations to fatigue. Subjects who showed a significant IMDF drop at L5 (i.e., subjects S2–S6) also showed dramatic changes in the components of the box acceleration. This observation suggests an association between muscle fatigue at

TABLE II
DIFFERENCES BETWEEN BIOMECHANICAL VARIABLES RECORDED IN THE SAGITTAL PLANE OF MOVEMENT AT THE BASELINE CONDITION AND AFTER 5 MIN OF CYCLICAL LIFTING. VALUES IN BOLD ARE SIGNIFICANT AT THE 0.05 LEVEL (WILCOXON TEST). "NS" INDICATES THAT CHANGES WERE NOT STATISTICALLY SIGNIFICANT. VARIABLES INCLUDE OVERALL CHANGES IN THE RANGE OF ANGULAR DISPLACEMENTS IN DEGREES AT THE KNEE, HIP, TRUNK, AND ELBOW (RANGE), CHANGES IN THE BOX MAXIMUM VERTICAL (1Y) AND HORIZONTAL (1X) ACCELERATION, AND CHANGES IN THE ESTIMATED NORMALIZED TORQUE (NM/WEIGHT X HEIGHT OF THE SUBJECT) AT THE L4–L5 SPINAL LEVEL

	S1	*S2*	*S3*	*S4*	*S5*	*S6*	*S7*	*S8*	*S9*
RANGE									
KNEE	***7.0***	***11.7***	***6.3***	***-13.5***	NS	NS	***2.8***	NS	NS
HIP	***2.2***	***5.5***	***5.6***	***4.5***	***-2.0***	***8.0***	***2.5***	***4.2***	NS
TRUNK	NS	***3.3***	NS	***4.5***	NS	NS	***8.7***	***8.2***	NS
ELBOW	NS	***10.1***	***8.9***	***20.8***	***23.9***	***-7.6***	NS	NS	NS
BOX									
ACC_Y	NS	***122 %***	***62 %***	***77 %***	NS	***93 %***	NS	***21 %***	NS
ACC_X	NS	NS	NS	***31 %***	***45 %***	***150 %***	***16 %***	NS	NS
TORQUE									
L4-L5	NS	***0.31***	NS	NS	***-0.24***	***0.29***	NS	NS	NS

the lumbar region and the way the subject manipulates the box during the exercise. However, only when the acceleration of the box doubled from the beginning to the end of the task was there a significant increase in the torque at L4/L5.

IV. DISCUSSION

Fatigue-related changes in the EMG data were identified as an IMDF percentage drop from the beginning to the end of the lifting task. A significant IMDF decrease was observed in six subjects out of nine. These changes were more consistent for the multifidus muscle at L5 and less consistent for both the iliocostalis lumborum muscle at T10 and the longissimus thoracis muscle at L1. The six subjects for whom significant changes in the EMG characteristics were demonstrated also showed significant changes in the biomechanics of motion. This finding supports the hypothesis that associations between muscle fatigue and biomechanical adaptations to the task are present in healthy subjects.

Two macroscopic adaptations to the task were described as changes in joint range of motion. We indicated these changes as *upper limb* and *lower limb biomechanical adaptations* or *strategies*. We see changes in knee and elbow angular displacements, together with the constraints related to the position of the lower and upper shelves, as determinants of the trunk and pelvis position and orientation. In fact, given the box trajectory, the shoulder position may be assessed based on the elbow angular displacement, which also determines position and orientation of forearm and arm. Also, given the feet position, the hip position may be assessed based on the knee angular displacement, which determines the position and orientation of the shank and thigh. Once the position of the hip and shoulder are known, the position and orientation of the pelvis and trunk are determined based on the geometry of the body segments [19], [24]. The presence of lower and upper limb strategies appears to be consistent with previous work that has demonstrated the significant effect of upper limb [5] and lower limb [37] fatigue on the lifting techniques.

Changes in the box acceleration provide a method of identifying individual response strategies to the fatiguing exercise. Increases in the vertical and horizontal components indicate that the subject may have tried to take advantage of the box inertial components to facilitate the lifting task. When a dramatic change in the box acceleration was detected, a significant increase in the maximum torque at L4/L5 was also observed. A decrease in the maximum torque at L4/L5 was identified only for one subject. In this subject, the largest increase in elbow range of motion was observed at the end of the task, thus suggesting that an upper limb strategy may be effective in reducing the load on the trunk.

Previous studies by others have investigated changes in the lifting technique due to fatigue, but none of these studies is fully comparable with our investigation. Changes in the joint torques at both the upper and lower extremities were documented during a lifting task by Fogleman and Smith [10], thus suggesting considerable variation in the strategies used by different individuals. This variability in the subjects' response to the lifting task was also observed by Resnick [28]. Although these studies are relevant from the ergonomic standpoint and demonstrate fatigue-related biomechanical adaptations during lifting, two aspects distinguish them from our work: 1) fatigue was measured either by means of the Borg scale or by changes in the heart rate and 2) the lifting duration (i.e., 2 to 4 h) and repetition rate (one to three repetitions per minute) were significantly different from those specified in our study.

Sparto *et al.* [36] and Dolan and Adams [8] investigated the effect of fatigue on the biomechanics of lifting using experimental procedures that may be more closely compared with our study. Sparto *et al.* [36] documented a decrease in knee and hip range of motion and an increase in the spine peak flexion during a lifting test. The task was performed using a lifting device [36] set to simulate the inertial components of a box with weight equal to 25% of the subject's maximal lifting capability. This choice resulted in a load approximately three times greater than that utilized in our experiment. Subjects continued the exercise until they felt they could no longer continue, or until their heart rate exceeded 180 beats/min (approximately 2 min of exercise). Fatigue was documented as a reduction in lifting power. Differences between our study and theirs are primarily related to the load utilized in the test, which is a factor known to affect the lifting technique [12], [33] and the method of fatigue assessment, which they based on the mechanical power.

EMG-based fatigue assessment during lifting has been utilized by others. Dolan and Adams [8] adopted a protocol of lifting and lowering a 10-kg weight from floor to waist height

100 times. The repetition rate was self-selected and was between eight to nine lifts per minute. Their study demonstrated an increase in lumbar flexion, an increase in peak bending moment, and a decrease in peak spinal compression. Surface EMG data from the iliocostalis lumborum muscle at T10 and the longissimus thoracis muscle at L3 were recorded during static contractions before and immediately after the 100 lifts. A decrease in the EMG median frequency was documented at L3, but no significant correlation was found between muscle fatigue and biomechanical adaptations. Two main points contrast our study and Dolan and Adams' [8] investigation. First, the EMG parameters in our study were derived by analyzing data from the dynamic task, rather than interposed static tasks. Our approach is expected to be more sensitive to muscle fatigue than static sampling before and after the exercise [2], [4], [17], [32]. The lack of correlation between EMG and biomechanics in Dolan and Adams' [8] study may be attributed to the unavailability at that time of methods to analyze fatigue during the lifting task as opposed to sampling the EMG data before and after the task. A second difference in the two studies is that we recorded surface EMG data from the multifidus muscle at L5, which turned out to be the most sensitive muscle site to fatigue. This was previously shown in studies investigating static contractions, which indicated a fatigue pattern across paraspinal muscles consistent with our findings for dynamic conditions [30], [31].

The importance of this study to rehabilitation is based on evidence that muscle endurance and muscular activation patterns are severely compromised in patients with LBP [14], [23]. It is hypothesized that back muscle fatigue may lead to compensatory muscular and biomechanical strategies during repetitive lifting [36]. Although the mechanism of injury that supports the role of back muscle endurance as an etiologic factor of LBP has not been identified yet, there is growing scientific evidence supporting the likelihood that back muscle fatigue leads to a loss of trunk muscle coordination [26], [36] and to a progressive involvement (i.e., stress) of passive tissues of the spine [8], [21]. These effects may increase the risk of back injuries. Moreover, the lack of coordination of trunk muscles secondary to fatigue might interfere with the compensatory muscular patterns that are expected to occur secondary to LBP. Consequently, monitoring of back muscles during demanding tasks might help clinicians identify back muscle endurance and imbalance deficits, which can be used to validate rehabilitation modalities programmed to reverse these deficits. Current methods of assessing functional capacity evaluations through subjective scoring lack the objectivity and precision that quantitative EMG and biomechanical measures provide. In this context, our work may be seen as a first step toward augmenting current functional capacity evaluations in LBP subjects. The results of this study have demonstrated that fatigue-related EMG and biomechanical changes may be identified in healthy subjects during a repetitive lifting task. These trends in the EMG and biomechanical parameters may be altered in LBP subjects in a way that depends on their degree of muscle strength and endurance deficits. A primary problem remaining in using these methods in a clinical setting is the relatively high cost in time and equipment that would be required for such measurements. Further development is needed to identify which measures are most reliable and sensitive to changes following therapeutic interventions and to design equipment that can provide such measurements at reasonable costs.

V. Conclusion

The associations between muscle fatigue and changes in the biomechanics of motion during a repetitive lifting exercise in subjects with no history of LBP were described. The study is based on a novel approach of combining time-frequency analysis of the EMG signal with biomechanical measurements of the lifting technique. A consistent decrease in the IMDF was observed in almost all the subjects tested. For those subjects with a significant decrease in IMDF, a lower limb and/or an upper limb biomechanical adaptation to fatigue was observed during the exercise. Increase in the peak box acceleration components were documented in these subjects, which appeared to lead to an increase in the torque at L4/L5 when changes in the box acceleration were dramatic. Our work opens possibilities toward identifying muscle strength and endurance deficits in LBP subjects where a lack of "biomechanical adaptation to fatigue" or "excessive muscular response to the lifting task" could be identified.

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