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A single metric for quantifying biomechanical stress in repetitive motions and exertions

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The relative effects of repetition, force and posture were studied in order to investigate how continuous biomechanical measurements can be combined into a single metric corresponding to subjective discomfort. A full factorial experiment was conducted involving repetitive wrist flexion from a neutral posture to a given angle against a controlled force. Seven subjects performed the task using two paces (20 and 4 motions/min), two force levels (15 and 45 N) and two angles (15 and 45°) for 1 h each. Discomfort was reported on a 10 cm visual analogue scale anchored between 'no discomfort' and 'very high discomfort'. Repeated measures analysis of variance showed that all main effects were statistically significant ($p<0{\cdot}05$) and no significant interactions were observed. A linear regression model was fitted to the data and used for generating frequency weighted digital filters that shape continuous recordings of repetitive motions and exertions into an output proportional to relative discomfort. The resulting high-pass digital filter had a 22 dB/decade attenuation slope. A simulated industrial task used for validating the model involved repetitively transferring pegs across a horizontal bar and inserting them into holes against a controlled resistance. Angular wrist data were recorded using an electrogoniometer and filtered. Six subjects performed the task of the three conditions consisting of (1) 15° wrist flexion, 15 N resistance and 6 motions/min, (2) 15° wrist flexion, 45 N resistance and 12 motions/min, and (3) 45° wrist flexion, 45 N resistance and 15 motions/min. Subjective discomfort was reported after performing the task for 1 h. Pearson correlations between subjective discomfort ratings and the integrated filtered biomechanical data for individual subjects ranged from 0·90 to 1·00. The pooled correlation across subjects was 0·67. This approach may be useful for physical stress exposure assessment and for design of tasks involving repetitive motions and exertions.

1. Introduction

While repetitive motions, forceful exertions and extreme postures have been recognized as risk factors for cumulative trauma disorders (CTD) (Hymovich and Lindholm 1966, Rothfleisch and Sherman 1978, Armstrong and Chaffin 1979, Armstrong *et al.* 1982, Silverstein *et al.* 1986, Tanaka *et al.* 1988, Aarås *et al.* 1988, Moore *et al.* 1991, Moore and Garg 1994), quantitative dose-response relationships between these risk factors and CTDs are not yet available. This may be due to

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limitations associated with quantifying dose using multi-factor biomechanical measurements of force, posture and repetition.

In order to provide guidelines for industrial task design and evaluation, investigators have turned to the psychophysical approach by using short-term responses to physical stress corresponding to controlled laboratory experiments. Previous research has suggested that the risk of injuries increase when task requirements exceed psychophysical acceptance levels (Snook *et al.* 1978, Snook 1978, Snook 1985, Liles *et al.* 1984, Herrin *et al.* 1986). The psychophysical approach has been supported by physiological measures (Dahalan and Fernandez 1993, Kim and Fernandez 1993, Öster *et al.* 1994). Psychophysical methods have also been useful for evaluating tool designs (Schoenmarklin and Marras 1989b, Ulin *et al.* 1990, Örtengren *et al.* 1991) and to examine specific upper extremity tasks for establishing work design guidelines (Krawczyk and Armstrong 1991, Krawczyk *et al.* 1992, Krawczyk *et al.* 1993,Ulin *et al.* 1992, Snook *et al.* 1995).

Few studies have explored the combined effects of repetition, force and posture on the resulting psychophysical measures of subjective discomfort, or have attempted to model the combined relationships among them. Several investigations have studied main effects but have not considered interactions. Krawczyk and Armstrong (1991) studied the effect of posture, force and repetition on perceived exertion using a latin square design, thus interaction effects were not studied. Krawczyk *et al.* (1992) also studied the effects of repetition and movement distance (posture) on perceived exertion using a visual analogue scale, and on preferred weight (force) using the method of adjustment. Ulin *et al.* (1993b) studied the effects of work location (posture) and repetition on perceived exertion for screwdriving tasks. Ulin *et al.* (1993 c) also examined the effects of tool mass (force) and posture on perceived exertion for screwdriving tasks. Snook *et al.* (1995) investigated maximum acceptable forces for various paces, grips and types of motions, but since wrist flexion angle was fixed between 45° above and below neutral for all flexion and extension conditions, the wrist angle effect was not considered. Kim and Fernandez (1993) also used the method of adjustment to determine the maximum acceptable frequency for drilling tasks at different forces and wrist flexion angles (posture). Dahalan and Fernandez (1993) used the same approach to determine the maximum acceptable frequency (repetition) for gripping tasks at different forces and durations. None of the above mentioned studies combined the effects of all three factors (force, posture and repetition).

New analytical methods are becoming available for quantifying direct biomechanical measurements in repetitive manual tasks. Spectral analysis was demonstrated as a suitable method for quantifying repetitiveness and postural stress (Radwin and Lin 1993). Radwin *et al.* (1994) also demonstrated the concept of using frequency weighted filters for processing continuous biomechanical measurements from repetitive motions in order to combine postural deviations and repetitive motions. The relative effects of wrist flexion angle and repetition on subjective discomfort determined the shape of these filters. Frequency weighted filters may be useful in electronic exposure assessment instruments for assessing repetitive motion stress (Radwin and Yen 1993). Although frequency weighted filters show promise for quantifying repetitive and postural stress, filters for the combination of repetition, posture and force have not yet been investigated.

The objective of this study was to quantify and to model the relative effects of repetitive motion and force in order to investigate how biomechanical measurements of posture, force and repetition can be combined into a metric corresponding to subjective discomfort. This study expanded the frequency weighted filter approach introduced by Radwin *et al.* (1994) by adding a force factor. The relative effects of force, repetition and posture on subjective discomfort were used to demonstrate how force and frequency weighted angle filters can be established. It was hypothesized that there would be interactions among these factors. An experiment was also conducted to validate the resulting model using a simulated industrial task.

2. Methods and results

2.1. *Experiment 1: Model development*

2.1.1. *Apparatus:* A fixture similar to the one used by Snook *et al.* (1995) was modified and incorporated into this experiment for controlling both force and wrist flexion range (figure 1). A Model B-5 electromagnetic brake and a TC-5 torque controller (Magnetic Power Systems Inc., St Louis, MO) were both used for controlling resistance to motion. The input current to the brake was adjusted to control torque level. Torque resistance from the brake was calibrated by hanging a small container from the outer edge of a 15 cm diameter circular plate centred on the crank spindle. Lead beads were added by increments into the container until the weight provided enough torque to overcome the resistance from the brake for a given current. Since the weight of the container was always perpendicular to the moment arm (the radius of the circular plate), the product of the weight of the filled container and the radius of the plate determined the torque level. This calibration procedure was performed for eight different levels of current in a random order, and was replicated three times. Linear regression ($r^2 = 1{\cdot}00$) was used for determining the required input current corresponding to the desired resistance.

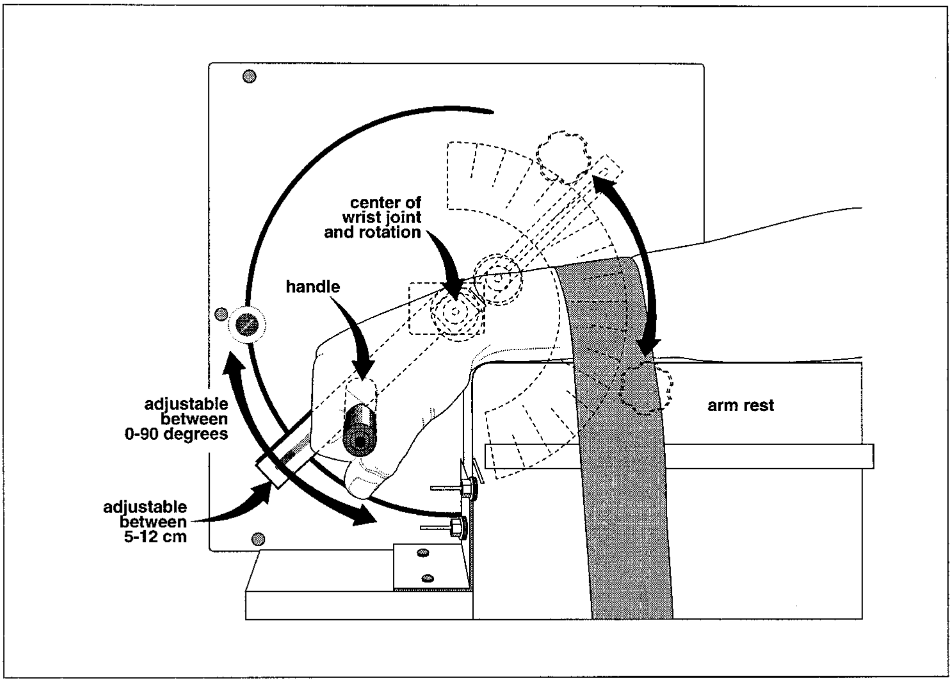

Figure 1. Fixture for controlling force and flexion angle.

A Plexiglas handle (2 cm diameter and 13 cm long) was attached perpendicularly to an aluminium bar fixed to the brake spindle (figure 1). The handle was adjusted so that the wrist joint centre of rotation was aligned with the rotation centre of the crank when grasping the handle. The distance between the centre of grip and wrist joint centre of rotation for each subject was measured to determine the torque level that assured every subject flexed against the same resistance. A moveable weight counterbalanced the load of the handle and bar. A clutch was installed at the brake spindle so that subjects flexed against resistance from the brake to a pre-determined angle, but experienced nominal resistance (0·015 Nm) during extension. An arm rest with a Velcro strap was used for providing stability and for controlling forearm posture to assure that the task was performed with the proper wrist motion. An adjustable mechanical stop was set for the resting posture at the horizontal position and another was adjusted to set the wrist flexion range. The fixture was located on top of a table that was adjusted so that subjects sat in an upright position with the elbow aligned with the top of the arm rest at seated elbow height and the forearm and upper arm formed a right angle.

The task was paced by two auditory cues and a visual display. The wrist flexed on one tone and extended on another tone. The period between the first tone and the second tone was set to 1 s for all conditions. Consequently the actual rest period varied, depending on the pace. The rest time remaining was indicated on a visual display. Subjects were instructed to relax in between exertions.

2.1.2. *Experimental design and procedure:* The experiment was a $2 \times 2 \times 2$ repeated measures full factorial design with pace, force and wrist flexion angle as the independent variables. Subjective discomfort was the dependent variable. The experimental task involved repetitively flexing the pronated wrist from a neutral posture to a given angle against a controlled resistance while holding a handle in a power grip. Wrist flexion was between a neutral posture and 15° or 45° flexion. Force was controlled at 15 and 45 N by setting the brake at appropriate torque levels. The task was performed for 20 and 4 motions/min. All experimental conditions were presented in a random order and only one condition was presented to a subject in a 24-h period. Every condition was performed continuously for 1 h. A two-min warm-up period was provided at the beginning of each session.

Discomfort was measured immediately at the end of each session using a 10 cm visual analogue scale anchored as 'No discomfort' at 0 cm, and as 'Very high discomfort' at 10 cm. A thin 0·5 cm vertical mark was placed on the line to indicate the mid-point of the scale. Subjects drew a vertical line across the horizontal scale to indicate their discomfort level. Subjects were advised that symptoms of discomfort included aching, fatigue, soreness, warmth, cramping, pulling, numbness, tenderness, pressing or pain. They were required to be symptom-free at the beginning of every session.

Multiple regression and mixed model analysis of variance, with subject as a random effects variable, were used to analyse the discomfort data. The resulting regression model was then used to specify the attenuation slopes for frequency weighted filters (Radwin *et al.* 1994).

2.1.3. *Subjects:* Seven subjects (six males and one female) ranging between 21 and 25 years of age were recruited from the university campus. Informed consent was provided by all subjects, who were free to withdraw at any time during the course of the study. Experimental procedures were reviewed and approved by the university Human Subjects Committee. All subjects were right-handed and had no restriction of hand/arm motion or history of hand/arm injuries. None of the subjects had industrial work experience. Subjects were paid on an hourly basis for participating. One of the seven subjects dropped out of the experiment owing to outside activities prior to the last two sessions. Data from these two sessions were treated as missing values and were estimated using the maximum likelihood method (BMDP Statistical Software Inc., Los Angeles, CA). The two estimated values were used in place of the two missing data in all analyses.

2.1.4. *Results:* Mean and standard deviation discomfort ratings for all eight conditions of pace, exertion and angle are shown in figure 2. Results from repeated measures mixed model analysis of variance revealed that pace ($F(1,6) = 19{\cdot}45$, $p < 0{\cdot}05$), exertion ($F(1,6) = 45{\cdot}94$, $p < 0{\cdot}05$), and angle ($F(1,6) = 13{\cdot}47$, $p < 0{\cdot}05$) were all significant main effects. As the pace changed from 4 to 20 motions/min, mean discomfort ratings increased from 2·83 (SD = 1·78) to 4·52 (SD = 2·13). The increase in force from 15 to 45 N increased mean discomfort ratings from 2·45 (SD = 1·44) to 4·90 (SD = 2·00). As the wrist flexion angle increased from 15° to 45°, mean discomfort ratings increased from 3·10 (SD = 1·86) to 4·25 (SD = 2·25). No significant interaction effects between pace and exertion ($F(1,6) = 0{\cdot}31$, $p > 0{\cdot}05$), exertion and angle ($F(1,6) = 4{\cdot}17$, $p > 0{\cdot}05$), pace and angle ($F(1,6) = 0{\cdot}14$, $p > 0{\cdot}05$), or among the three factors ($F(1,6) = 0{\cdot}07$, $p > 0{\cdot}05$) were observed.

In order to determine the attenuation slope, expressed as decibels per decade, for a frequency weighted filter (Radwin *et al.* 1994), a linear regression model was fitted with the mean logarithm transformed discomfort as the dependent variable, and the logarithm of frequency, exertion and wrist flexion angle as the independent variables. The resulting regression model was:

$$D = 10^{(-0{\cdot}183 + 0{\cdot}508 \log E + 0{\cdot}245 \log A + 0{\cdot}270 \log F)} - 1$$

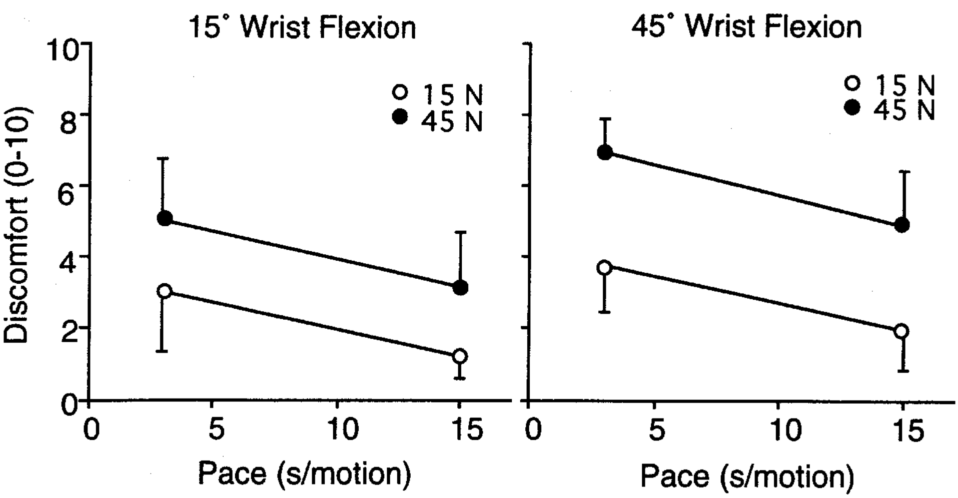

Figure 2. Mean and standard deviation of discomfort ratings for all combinations of conditions (7 subjects).

($R^2 = 0{\cdot}975$, $F(3,4) = 51{\cdot}095$, $p < 0{\cdot}01$), where D was discomfort (scale between 0 to 10), E was the exertion level (N), A was the wrist flexion angle (degrees), and F was frequency (Hz), which was equivalent to pace divided by 60. Equal discomfort strata were determined by solving the regression equation for a given discomfort.

Strata with exertion contours plotted against frequency and angle are shown in figure 3 for an arbitrary discomfort level of 4. Equal discomfort strata indicate corresponding exposures of exertion level, wrist angle and frequency that will result in equivalent levels of discomfort. For example, a task that requires 28 N exertion and 21° wrist flexion at the frequency of 0·2 Hz and a task that requires 30 N exertion and 39° wrist flexion at the frequency of 0·1 Hz would both result in a discomfort level of 4.

2.1.5. *Force and frequency weighted angle filter:* When biomechanical measurements include force and posture, both factors contain the properties of magnitude and frequency. A frequency weighted angle filter (Radwin *et al.* 1994) weighs flexion angle (degrees) by the corresponding frequency in proportion to the equal discomfort function. Combining posture and force therefore requires a 3-dimensional filter network that has the dimensions of force, posture and frequency. The current study assumes that force is a constant level. Consequently the wrist angular data are filtered through a frequency weighted angle filter and adjusted by the corresponding force factor. This process enables continuous flexion angle data to be filtered and integrated, so that it can be quantified as a single value.

The attenuation slope (dB/decade) for a frequency weighted filter can be obtained by algebraically solving the regression equation at a given discomfort and exertion level. For example, the angular attenuation over one decade can be determined by:

$$\begin{aligned} A_{\mathrm{dB}} &= 20 \times \log\left[\frac{X_{F/10}}{X_F}\right] \\ &= 20 \times (\log X_{F/10} - \log X_F) \end{aligned}$$

where A_{dB} is the angular attenuation, X_F is the wrist flexion angle (degrees) at

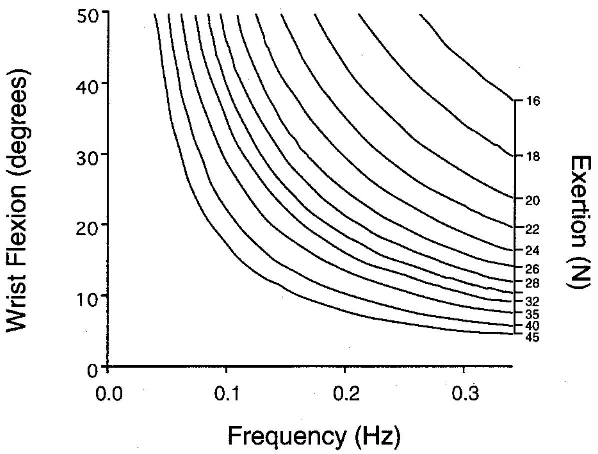

Figure 3. Equal discomfort strata for discomfort level 4. Maximum acceptable discomfort strata for 90% of female population.

frequency F (Hz) and $X_{F/10}$ is the wrist flexion angle at a frequency one decade less than F for a given discomfort and exertion level. If log ($X_{0\cdot06\ \mathrm{Hz}}$) = 2·24 for discomfort level 4 with a 20 N exertion, then for a frequency one decade greater, log ($X_{0\cdot6\ \mathrm{Hz}}$) = 1·14, yielding 22 dB attenuation. Based on the discomfort model, the attenuation slope for the corresponding frequency weighted angle filter is therefore 22 dB/decade.

The frequency weighted filter was modelled in MATLAB™ (The MathWorks Inc., Natick, MA) as a finite impulse response (FIR) high-pass filter having a 22 dB/decade attenuation slope in the linear phase region, as suggested by the discomfort model (figure 4). Finite impluse response filters are nonrecursive filters that have outputs solely dependent on the present and past inputs. The primary advantages of FIR filters are that they can be designed to have linear phase and that they are inherently stable.

The difference equation for the FIR filter is:

$$X_E(nT) = \sum_{k=0}^{N} b_k X[(n-k)T],$$

where $X_E(nT)$ is the output associated with the current sample time nT and $X(nT—kT)$ is the input value, k sample points in the past. The output value $X_E(nT)$ of the frequency weighted filter is the weighted sum of the input for the current sample time, $X(nT)$, and the input values of the preceding N samples points. Coefficients b_k for the resulting difference equation are listed in table 1.

The filter cut-off frequency was set at 1 Hz because it is believed that most industrial tasks would have frequencies below this range. The upper bound for wrist flexion was fixed at 75° in order to cover both the female and male range of wrist motion (NASA 1978, Marley and Fernandez 1995). This angle was then used as a reference for the angular attenuation (dB) for a given frequency and force level by algebraically solving the equation for the discomfort level 10. Level 10 was used to define the 0 dB angular attenuation at the cut-off frequency.

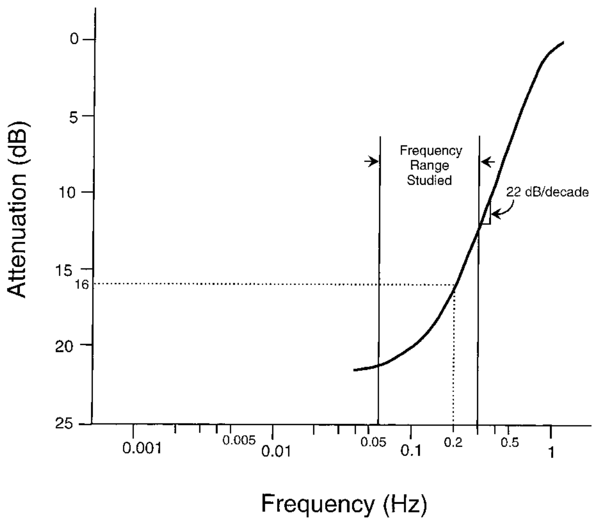

Figure 4. Characteristics of the frequency weighted angle filter.

The filtered angular data $X_E(nT)$ were weighted by a force factor to account for the corresponding force. Angular attenuation of 0·2 Hz according to the discomfort model was 27·5 dB for a 15 N exertion for instance, while angular attenuation at 0·2 Hz according to the filter without adjusting for force was 16·5 dB. Therefore the filtered angular data for a 15 N exertion was adjusted by a force factor of —11 dB (figure 4). The discomfort model was linear and the slope of the filter was constant, therefore the force factor for a given exertion level was constant. The force factor was equal to 0·65 E —21·25 dB, which was derived from solving a linear equation between two exertion levels and the corresponding force factors, where E is exertion in N. The root-mean-square (RMS) of the force and frequency weighted angular data $X_{EF}(nT)$ was then used to represent the relative exposure level $\overline{X_{EF}}\,(nT)$, where

$$\overline{X}_{EF}(nT) = X_E(nT) \times 10^{\left(\frac{0.65E-21.25}{20}\right)}.$$

A block diagram of the process is illustrated in Figure 5.

2.2. *Experiment II: Model validation*

2.2.1. *Apparatus:* A special peg board was designed to control force during peg insertion (figure 6). Ball plungers (Jergens Corp. Cleveland, OH) were used to independently adjust the resistance when each peg was inserted into a hole. The tighter the ball plunger was screwed, the greater the resistance. Each ball plunger resistance was set by adjusting the plunger and slowly pressing a strain gauge load cell against a peg until the peak force reading from the oscilloscope was within ±2%

Table 1. Coefficients for the FIR filter.

k	b_k
0	0·0012
1	0·0009
2	0·0000
3	— 0·0023
4	—0·0072
5	— 0·0155
6	— 0·0272
7	— 0·0419
8	— 0·0582
9	— 0·0742
10	— 0·0877
11	— 0·0967
12	0·8987
13	— 0·0967
14	— 0·0877
15	— 0·0742
16	— 0·0582
17	— 0·0419
18	— 0·0272
19	— 0·0155
20	— 0·0072
21	— 0·0023
22	— 0·0000
23	0·0009
24	0·0012

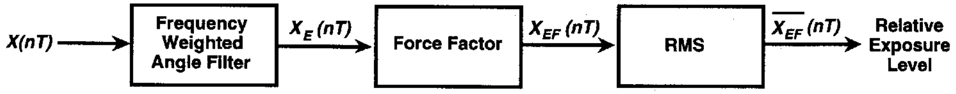

Figure 5. Block diagram for wrist flexion angular data processing.

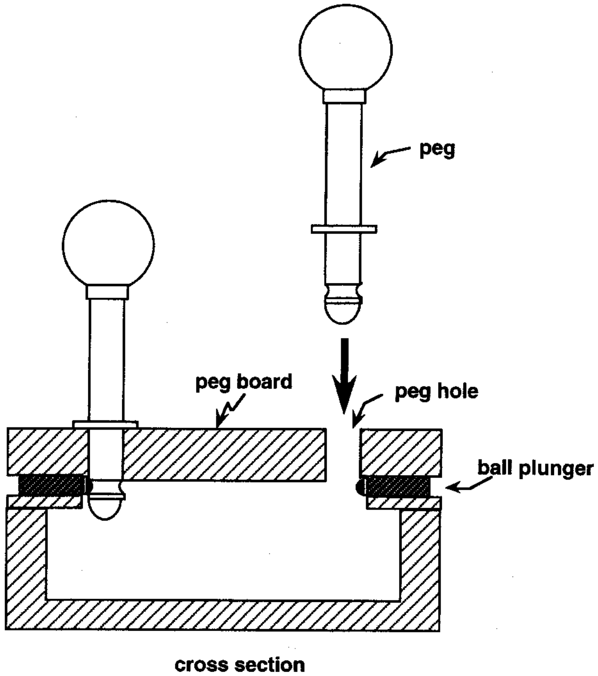

Figure 6. Side view of peg board with controlled resistance.

of the desired resistance. The height of the horizontal bar located in front of the peg board was adjusted in order to control the wrist flexion angle.

Two 2 ×4 peg boards were placed adjacent to each other on an height adjustable table making a 2 row by 8 column matrix of peg holes. The table was adjusted so that the peg board was at seated elbow height. Subjects continuously received a supply of pegs from a chute located next to the seat on their dominant side. They were instructed to insert pegs in a left to right fashion from the top row to the bottom, and to finish one peg board before inserting pegs into another. After a peg board was filled, the experimenter replaced it with an empty peg board, removed pegs from the filled board, and fed the pegs into the slot.

A Penny and Giles Biometrics model M110 strain gauge wrist electrogoniometer was fastened across the dorsal side of the wrist to continuously measure wrist flexion angle. The lateral epicondyle, radial styoid and second metacarpophalangeal joint were used as bony landmarks for aligning the wrist to determine the neutral posture (Schoenmarklin and Marras 1989a). The electrogoniometer was calibrated by setting the zero degree at the neutral wrist flexion. A MacAdios 12-bit analog-digital converter, LabView® software (National Instruments Corp., Austin, TX) and a

Macintosh II/fx microcomputer were used for sampling posture signals from the electrogoniometer and for implementing the digital filter for the posture data. Wrist flexion angular data were sampled at 20 Hz.

2.2.2. *Experimental design and procedures:* The experimental task involved repetitively transferring a peg across the horizontal bar and inserting it into a peg board against a controlled resistance. The experiment consisted of three experimental conditions: (1) 15° wrist flexion at the pace of 6 motions/min against a 15 N resistance, (2) 15° wrist flexion at the pace of 12 motions/min against a 45 N resistance, and (3) 45° flexion at the pace of 15 motions/min against a 45 N resistance. Experimental conditions were presented to each subject in a random order. Only one experimental condition was presented to a subject in a 24-h period. Every condition was performed continuously for 1 h. A 2-min warm-up period was provided at the beginning of each session.

Symptoms of discomfort were the same as defined in Experiment I. Subjects were required to be symptom-free at the beginning of every session. Discomfort was measured using the same visual analogue scale. Linear regression analysis was used with subjective discomfort as the dependent variable and the relative exposure level as the independent variable. Mixed model analysis of variance, with subject as a random effects variable and the experimental condition as the fixed effect, was also conducted.

2.2.3. *Subjects:* Six male subjects were recruited by broadcasting electronic mail announcements and posting signs on the university campus. Age ranged between 19 and 22 years. Five subjects were right-handed and one was left-handed. Subjects were required to have no restriction of hand/arm motion, or history of hand/arm injury. Subjects were paid on an hourly basis.

2.2.4. *Results:* Means and standard deviations of the subjective discomfort ratings and the relative exposure levels for the three conditions are listed in table 2. Results from repeated measures analysis of variance revealed that experimental conditions had significant effects for both the subjective discomfort ratings ($F(2,10) = 33{\cdot}05$, $p < 0{\cdot}05$), and the relative exposure levels ($F(2,10) = 104{\cdot}94$, $p < 0{\cdot}05$). Pearson correlation coefficients between subjective discomfort ratings and the predicted relative discomfort for individual subjects ranged between 0·90 to 1·00 (mean = 0·975, SD = 0·040). A linear regression model was fitted across subjects with subjective discomfort rating as the dependent variable and the relative exposure level as the independent variable. The resulting pooled subject model was:

Table 2. Mean and standard deviation for subjective discomfort ratings and the predicted relative exposure levels ($n = 6$ subjects).

Experimental conditions			Subjective discomfort rating	Relative exposure level
Force (N)	Angle (°)	Pace (motions/min)	Mean (SD)	Mean (SD)
15	15	6	1·70 (1·75)	1·13 (0·22)
45	15	12	4·65 (1·35)	12·88 (3·43)
45	45	15	6·38 (0·92)	18·62 (3·07)

$$D = 2.47 + 0.19\hat{D}$$

($r^2 = 0{\cdot}67$, $F(1,16) = 12{\cdot}79$, $p < 0{\cdot}01$), where D was discomfort (scale of 0 to 10), $\hat{D}$ was the relative exposure levels estimated by the RMS of the frequency and force weighted angular data.

2.2.5. *Validation of the discomfort model using data from Snook* et al. *(1995):* Snook *et al.* (1995) used the method of adjustment to estimate the maximum acceptable force for wrist flexion at various levels of repetition. Subjects worked 7 h daily in that study and were instructed to work as hard as they could without developing 'unusual discomfort' by selecting the maximum acceptable force. Table 12 of Snook *et al.* (1995) summarized the estimated maximum acceptable forces for wrist flexion between 45° above and below the horizontal at 2 motions/min (0·03 Hz), 5 motions/min (0·08 Hz), 10 motions/min (0·17 Hz), 15 motions/min (0·25 Hz), and 20 motions/min (0·33 Hz) by different percentages of the female population studied. Data from table 12 in Snook *et al.* (1995) was entered into the resulting discomfort model from the current study by converting paces into frequencies, treating maximum acceptable forces at exertion levels, and assuming 90° wrist flexion for all conditions. For example, the maximum acceptable force at 0·25 Hz for 90% of the female population was 12 N (Snook *et al.* (1995), table 12), therefore the expected discomfort level was equal to:

$$10^{(-0.183+0.508\log(12)+0.245\log(90)+0.270\log(0.25))} - 1 = 3.80.$$

The expected discomfort levels are summarized in table 3. Differences among the expected discomfort levels between 0·17 Hz and 0·33 Hz were within 0·5 units (on a 10·0 point scale) with a coefficient of variation less than 0·03 for any given percentage of the female population. Equivalent expected discomfort levels were observed among conditions between 0·17 Hz and 0·33 Hz. The mean expected discomfort levels from these conditions may be considered maximum acceptable discomfort levels. The resulting maximum acceptable discomfort was 3·72 (SD = 0·10) for 90% of the female population for frequencies between 0·17 Hz and 0·33 Hz. When comparing expected discomfort levels from conditions of 0·08 Hz and 0·03 Hz, larger differences were observed (table 3).

Table 3. Expected discomfort levels from fitting data from Snook *et al.* (1995) into the discomfort model from the current study (assuming 90° wrist flexion for all conditions) ($n = 15$ subjects).

	Pace of motion				
Percent of population	2/min (0·03 Hz)	5/min (0·08 Hz)	10/min (0·17 Hz)	15/min (0·25 Hz)	20/min (0·33 Hz)
90	2·11	2·98	3·57	3·80	3·78
75	2·90	3·99	4·71	5·00	4·97
50	3·61	4·90	5·74	6·11	6·08
25	4·24	5·70	6·65	7·09	7·04
10	4·74	6·35	7·39	7·86	7·81

3. Discussion

Results of this study showed that the main effects of frequency, force and angle were all significant, and that no significant interaction effects among them were observed. A linear model therefore can be used to describe the relationship among the three factors and their effects on subjective discomfort. These findings are supported by other studies. Dahalan and Fernandez (1993) found that grip force was a significant main effect on maximum acceptable frequency and ratings of perceived exertion in simulated gripping tasks. Marley and Fernandez (1995) found that maximum acceptable frequency decreased as wrist deviation increased. Snook *et al.* (1995) observed that maximum acceptable force levels decreased as repetition increased. Kim and Fernandez (1993) indicated that both force and wrist flexion angle (posture) were significant main effects for maximum acceptable frequency and ratings of perceived exertion for drilling tasks. They also failed to find a significant interaction between force and wrist flexion angle as indicated in this study. Ulin *et al.* (1993b) found that work location (posture) and work pace (frequency) were both significant main effects on ratings of perceived exertion, and that the interaction effect was not significant. Results from Ulin *et al.* (1993c) revealed that tool mass (force) and work location (posture) were both significant effects in determining ratings of perceived exertion in screwdriving tasks, and the interaction effect was not significant. Krawczyk *et al.* (1992) found that frequency was a significant main effect for both preferred weight and perceived exertion, and distance (posture) was a significant effect of preferred load. The interaction effect of frequency and distance (posture) was not significant for both preferred load and perceived exertion.

Data from Snook *et al.* (1995) were entered into the discomfort equation in order to validate the discomfort model under two assumptions. The first assumption was that the additive relationship among force, pace and wrist flexion angle still holds for data from 7 h exposure as it did for the 1 h exposure used in the current study for the range of wrist flexion examined (45° above and below the horizontal). The second assumption was that subjects maintained a fixed reference in determining the maximal acceptable force, so that the selected maximum acceptable forces would result in approximately the same level of discomfort across different paces. Under these two assumptions, the discomfort model could be validated by comparing expected discomfort levels across different combinations of maximum acceptable forces and paces. Equivalent discomfort levels should result across conditions if the current model was valid.

Studies have indicated that the psychophysical approach can be used to define acceptable exposure levels (Snook *et al.* 1995, Marley and Fernandez, 1995). Discomfort models, like the one developed in the current study, may be used for objectively assessing continuous biomechanical data for subjects performing various combinations of force, posture and frequency. Consequently, they are more efficient in providing guidance for task design and evaluation than the method of adjustment unless many different combinations of force, posture and frequency are included. The current research suggests that use of the method of adjustment such as the one used by Snook *et al.* (1995) for determining acceptability, combined with continuous discomfort models, such as the one in this study, together can be used to establish the maximum acceptable discomfort levels for arbitrary tasks that are critical for design and evaluation, since requirements exceeding these levels would result in unusual discomfort.

The maximum acceptable discomfort level was determined in the current study as the relative equivalent expected discomfort levels observed by fitting data from Snook *et al.* (1995) into the discomfort model for conditions between 0·17 Hz and 0·33 Hz. The finding of the relative equivalent expected discomfort levels validated the current discomfort model within the assumptions stated earlier and also indicated that the discomfort model that was established based on data from 1-h exposures could be used for estimating expected discomfort levels for 7-h exposures. A discrepancy against this finding however occurred at 0·08 Hz and 0·03 Hz when the expected discomfort levels for these two conditions deviated from the equivalent expected discomfort levels (table 3). An explanation for this discrepancy may be because of strength limitations.

Hallbeck (1994) showed that mean wrist flexion strength for females was 48·28 N at 45° extension and 59·76 N at 45° flexion. The maximum acceptable force observed in Snook *et al.* (1995) for wrist flexion between 45° above and below horizontal for 50% of the females at 0·08 Hz was 32·3 N, which was more than 66% of the mean flexion strength at 45° wrist extension and more than 50% of the mean flexion strength at 45° wrist flexion observed by Hallbeck (1994). It is possible that during excessive exertions, force itself would be sufficient to cause unacceptable discomfort without regard for repetition (i.e. slower paces may not accommodate for excessive force). The fact that the maximum acceptable forces for 0·08 Hz and 0·03 Hz (Snook *et al.* 1995) were the same further supports the explanation that the maxium acceptable force may have been limited by strength rather than by repetition. This finding also indicated that excessive force requirements, even when they are performed at a relatively low pace, may result in greater discomfort than anticipated by the discomfort model. Task requirements exceeding these forces thus should be avoided regardless of the resulting discomfort levels predicted. Considering limitations from the speed and the range of wrist motion based on the same reasoning, it further suggests that the discomfort model might deviate from linearity when task requirements reach certain extremes of physical exposure, i.e. these extremes of exposures define the boundaries for applications of the linear discomfort model. Further investigations should be conducted to determine these extremes and to investigate how other task parameters may contribute to these limits (e.g. since wrist posture has significant effect on strength, the strength limit will vary for tasks with different ranges of wrist motion).

Radwin *et al.* (1994) studied the relative effects of frequency and wrist flexion angle on subjective discomfort and suggested that while other control variables were set at the same level, discomfort ratings for a task performed at 0·1 Hz and 0·05 Hz were not significantly different. This finding suggests that sufficient recovery may occur for tasks performed at a pace slower than a certain frequency, where discomfort may become insignificant. Future studies should investigate discomfort from repetitive motion and exertion tasks when sufficient recovery occurs.

Based on data from Snook *et al.* (1995) and the discomfort model in this study, and assuming 90° wrist flexion for all conditions, the maximum acceptable discomfort level for repetitive hand intensive tasks for 90% of the female population was 3·72 on the 10-point scale used. Figure 3 shows the equal discomfort strata for discomfort level 4 with exertion contours plotted against frequency and angle. Any combination of frequency and wrist flexion angle above the 28 N contour for tasks requiring 28 N or higher exertions would result in unacceptable discomfort.

Although general applications of this research are premature, this study explored the relative effects among the three factors for discomfort and took the preliminary step in modelling the quantitative relationship between these factors and subjective discomfort. Results from the validation experiment (Experiment II) showed that the discomfort model established in Experiment I had significant capability in predicting subjective discomfort for a repetitive hand intensive task. The strong correlation between subjective discomfort ratings and the relative exposure levels estimated by the RMS of the force and frequency weighted angular data demonstrated the feasibility of implementing force and frequency weighted angle filters in an electronic biomechanical exposure assessment instrument. Such an instrument could be used to record and integrate continuous multi-factor biomechanical data into a single value proportional to psychophysical discomfort level. The large constant term in the regression between subjective discomfort ratings and relative exposure levels may have been due to differences in the tasks performed in the model development experiment and in the validation experiment. The model development experiment used a highly-controlled task that isolated the wrist motion while the validation experiment had minimal restrictions on motions of the upper extremity, consequently involving different muscle groups. The main concern of this study was the relative differences between conditions used for establishing the filter attenuation slope.

There are several limitations of the discomfort model developed. The study only investigated wrist flexion motion with a power grip from the neutral position, which may not be representative for tasks involving different grip postures (e.g. pinch grip) and wrist motions (e.g. ulnar and radial deviations, wrist rotations, and sustained postures and exertions). Only a relatively small group of subjects participated in establishing the model at this preliminary stage. There were only two control levels within each factor. More control levels within each factor are desirable for further exploring the effect of individual factors on discomfort. Force was assumed to be a constant level in this study, while in most cases it varied by the corresponding task elements and should be measured as another entry of continuous biomechanical stress like the wrist flexion angles. The boundaries for applications for the linear discomfort model were not defined. Future studies involving longer exposure and more levels of control, including the sustained level for each risk factor, are necessary for refining the discomfort model and for defining its boundaries.

This research investigated the short-term effects of repetitive motions and exertions. Subjective discomfort was selected as a short-term response to physical stress in this study. Discomfort is significant since discomfort in the workplace may be associated with risk of injuries and should be reduced. Discomfort was considered also because it can be measured within a reasonable amount of time and discomfort can provide guidance in work design and evaluation, particularly when the quantitative exposure criteria for reducing the risk of CTDs is still unavailable. When such data becomes available, morbidity data could be treated in a similar approach as the response for establishing frequency weighted filters. Although an additive model was apt for subjective discomfort, it is not yet clear if a linear model will suffice for morbidity data (Silverstein *et al.* 1986).

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