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Agreement between a frequency-weighted filter for continuous biomechanical measurements of repetitive wrist flexion against a load and published psychophysical data

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A previous pilot study demonstrated that a force and frequency-weighted filter network could be developed for processing continuous biomechanical measures of repetitive wrist motions and exertions. The current study achieves the objective by modelling subjective discomfort for repetitive wrist flexion using controlled posture, pace and force. A three-level fractional factorial experiment was conducted involving repetitive wrist flexion (2 s/motion, 6 s/motion, 10 s/motion) from a neutral posture to a given angle (10°, 28°, 45°) against a controlled resistance (5 N, 25 N, 50 N) using a Box Behnken design. Ten subjects participated. Discomfort was reported on a 10 cm visual analogue scale. Results of response surface regression analysis revealed that main effects of force, wrist flexion angle, and repetition were all significant ($p < 0.05$) and that no second-order effects were observed. Linear regression analysis on these factors established a discomfort model on which the filter characteristics were based. The pure error test model revealed no significant lack of fit ($p > 0.05$). The continuous model was compared and agreed with discrete psychophysical data from other published studies. The model was used for generating parameters for a force and frequency-weighted digital filter that weighs continuous wrist postural signals with corresponding force in proportion to the equal discomfort function as a function of frequency of repetition. These filters will enable integration of large quantities of biomechanical data in field studies.

1. Introduction

Psychophysical assessment of subjective discomfort is often used as a short-term response to biomechanical stress in order to provide guidance for ergonomic work design and evaluation. Discomfort in the workplace is significant because it is undesirable, it is sometimes considered a precursor to musculoskeletal disorders, and because it may distract workers and lead to adoption of inappropriate or unsafe working methods. Hunting *et al.* (1994) showed that more than 15% of the participating electricians, who suffered from discomfort symptoms that did not occur frequently enough to be qualified as musculoskeletal disorders, would miss work, require medical care, or switch to light-duty jobs.

Psychophysical measurements have been shown to be sensitive to variations in task parameters and tool configurations (Schoenmarklin and Marras 1989, Ulin *et al.* 1990, Krawczyk and Armstrong 1991, Ulin *et al.* 1993a, b, Kihlberg *et al.* 1993, Genaidy *et al.* 1995), and they have been used to evaluate tasks and workstation designs (Genaidy and Karwowski 1993, Ulin *et al.* 1993a, b, Saldaña *et al.* 1994,

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Graf *et al.* 1995). Studies have indicated that psychophysical measurements are supported by objective physiological measures, including EMG, heart rate, blood pressure, strength, anthropometric predictions, and biomechanical data (Ulin *et al.* 1992, Dahalan and Fernandez 1993, Kim and Fernandez 1993, Öster *et al.* 1994, Genaidy *et al.* 1995, Marley and Fernandez 1995).

In addition, subjective discomfort has been used as a reference to establish maximum acceptable exposure. Kim and Fernandez (1993) used the method of adjustment to determine the maximum acceptable frequency for drilling tasks at different forces and wrist flexion angles. The same approach was used by Dahalan and Fernandez (1993) to determine the maximum acceptable frequency for gripping tasks at different force and duration. Snook *et al.* (1995) also used the method of adjustment to determine the maximum acceptable force for various paces, grip postures and types of motions. Subjects in that study were instructed to select the maximum acceptable forces that would not cause unusual discomfort after a 7-h work period. Maximum acceptable frequency for a sheet metal drilling task were estimated by Marley and Fernandez (1995) using the same method. Few studies, however, have explored the individual and combined effects of repetition, force and posture on resulting subjective discomfort, or have attempted to model the combined relationship among them. Furthermore, most psychophysical methods are limited in that they are subjective rather than objective.

Force and frequency-weighted filters have been shown to be promising for processing and integrating large quantities of objective continuous biomechanical measurements into a single value that accounts for physical stress exposure (Radwin *et al.* 1994, Lin 1995, Lin *et al.* 1997). The objective of the current study is to refine and broaden the scope of the pilot study (Lin *et al.* 1997) by expanding the range of exposure and increasing the control levels within each factor to examine individual and combined effects, in order to generate parameters for filters to assess biomechanical stress exposure in proportion to discomfort for a specific motion. This study used response surface modelling for establishing a quantitative model that associates physical stress of force, posture and repetition with relative discomfort. The resulting discomfort model is validated by incorporating data from other published studies that used various psychophysical methods. Theory and assumptions for these validation approaches are explored. A frequency-weighted angle filter was developed based on this complete model for processing continuous biomechanical measurements for arbitrary tasks.

Methods and results

2.1. *Model development*

2.1.1. *Methods*: A fixture for controlling wrist flexion range and hand force was used in this experiment (Lin *et al.* 1997). Torque resistance from an electromagnetic brake was controlled by adjusting the input current. A Plexiglas handle (2 cm diameter and 13 cm long) was attached perpendicularly to one end of a bar while the other end of the bar was fixed to the brake spindle. The handle was adjusted so that while grasping the handle, the wrist joint centre of rotation was aligned with the spindle. A clutch was installed on the brake spindle so that subjects flexed against resistance, but extended against nominal resistance (0·015 Nm). Wrist flexion angle was monitored using a Penny and Giles model M110 electrogoniometer (Penny and Giles Biometrics Ltd., Gwent, UK) and an angle display unit.

The experimental task involved repetitively flexing the pronated wrist from a neutral posture to a pre-determined angle against a controlled resistance while holding a handle with the dominant hand using a power grip. Subjects were instructed to exert just enough force to overcome the resistance of the brake, and to relax between exertions. An 1-h training session was provided to assure that subjects learned the task properly. A 25 N resistance, 28° wrist flexion and 6 s/motion pace were used for demonstrating and practising the task during training.

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

This experiment used a Box and Behnken (1960) three-level fractional factorial design in order to examine the combined and individual effects of force, wrist flexion angle, and pace on subjective discomfort. This design was chosen because it needed only 15 experimental conditions (including three repeated centre conditions) to examine first- and second-order effects, therefore greatly reducing the number of times subjects had to return for the experiment. This design was formed by combining a balanced incomplete block of the 2^2 factorial design with three centre points that allowed estimation of pure error. The three-level fractional factorial design matrix and the corresponding experimental conditions are presented in table 1. These experimental conditions covered more than 60% of the mean static wrist flexion strength at 45° wrist flexion (Hallbeck 1994) and 60% of the female range of wrist flexion (NASA 1978). The three identical central conditions (conditions 13, 14 and 15 in table 1) were presented to each subject as the first, eighth, and last conditions; the rest of the experimental conditions were presented in a random order. Only one condition was presented to a subject on a given day. A 2-min warm-up period was provided at the beginning of every session.

Table 1. Design matrix of the Box and Behnken (1960) three-level fractional factorial design for three factors and the corresponding experimental conditions.

Condition	Design matrix			Force (N)	Angle (°)	Pace (s/motion)
1	–	–	0	5	10	6
2	+	–	0	50	10	6
3	–	+	0	5	45	6
4	+	+	0	50	45	6
5	–	0	–	5	28	10
6	+	0	–	50	28	10
7	–	0	+	5	28	2
8	+	0	+	50	28	2
9	0	–	–	25	10	10
10	0	+	–	25	45	10
11	0	–	+	25	10	2
12	0	+	+	25	45	2
13	0	0	0	25	28	6
14	0	0	0	25	28	6
15	0	0	0	25	28	6

Subjects were advised that discomfort in this study included sensations like fatigue, soreness, stiffness, numbness, or pain. They were required to be free of any of these sensations at the beginning of every session. Discomfort was measured using a 10 cm visual analogue scale anchored between 'none' at 0 cm, to 'very high' at 10 cm, where 'none' meant that none of the discomfort sensations were experienced during the experiment and 'very high' meant that a very high level of these sensations were experienced. A thin 0·5 cm vertical line was located at the mid-point of the scale. Subjects drew a vertical line across the horizontal scale to indicate discomfort.

Subjects performed the task continuously for 1 h. Discomfort was assessed at the conclusion of the 1-h period. Subjects then performed the task for another 5 min after a 1-min break. Another discomfort rating was assessed after the 5-min period. Data for the 5-min period was disregarded. This helped to assure that subjects would pay attention to rating discomfort at the conclusion of 1 h by suppressing anticipation of leaving.

Response surface regression analysis and analysis of variance for linear, square and interaction effects were used to determine the significance of the first-order and second-order effects. Discomfort ratings were logarithm-transformed for all regression and variance analysis. Data from the pilot study (Lin *et al.* 1997) and the current study were pooled for modelling the relative effects of force, wrist flexion and repetition on subjective discomfort. Examination of residuals was conducted to investigate the group effect. A pure error test was used to test the goodness of fit. The resulting regression model was then used to specify parameters for a force and frequency-weighted angle filter (Radwin *et al.* 1994, Lin 1995).

Ten subjects (four males and six females) ranging between 20 and 25 years in age were randomly recruited from the university campus. All subjects were right-handed. They were required to have no history of hand or arm injuries and no restriction of motion. Subjects were paid on an hourly basis for their participation.

2.1.2. *Results*: Mean discomfort ratings for the 15 conditions of force, wrist flexion angle and pace are shown in figure 1. Analysis of variance of the log transformed discomfort ratings revealed that the linear effect was statistically significant ($F(3,5) = 66·05$, $p < 0·001$) and that no significant second-order effects were observed ($p > 0·05$). Response surface regression analysis showed that all main effects of force ($t_{(5)} = 11·89$, $p < 0·05$), wrist flexion angle ($t_{(5)} = 4·44$, $p < 0·05$) and pace ($t_{(5)} = 6·09$, $p < 0·05$) were statistically significant. On a scale from 1 to 10, mean discomfort increased from 1·47 (SD = 0·68) for 5 N exertions, to 2·94 (SD = 1·04) for 25 N exertions, and to 4·85 (SD = 1·07) for 50 N exertions. Increasing wrist flexion angles increased discomfort from 2·44 (SD = 1·52) for 10° flexion, to 3·05 (SD = 1·68) for 28° flexion, and to 3·68 (SD = 1·56) for 45° flexion. Discomfort was elevated from 2·28 (SD = 1·38) for 10 s/motion, to 2·90 (SD = 1·44) for 6 s/motion, and to 4·13 (SD = 1·72) for 2 s/motion. Tukey *post-hoc* test comparisons among mean discomfort ratings for all conditions are summarized in table 2.

No significant differences in discomfort among the three repeated conditions (experimental conditions 13, 14 and 15 in table 1) were observed ($F(2,27) = 0·89$, $p > 0·05$). Ratings for these conditions were therefore considered as replicates and were used for estimating pure error.

The discomfort model was used to develop a frequency-weighted angle filter, mean transformed discomfort ratings were fitted against the dependent variables of logarithm of frequency, exertion and wrist flexion angle. The discomfort data from the current experiment were pooled with data acquired from a previous pilot study (Lin *et al.* 1997). The resulting residuals revealed that residuals from the pilot data were significantly greater than residuals from the current study ($p < 0{\cdot}05$). A group

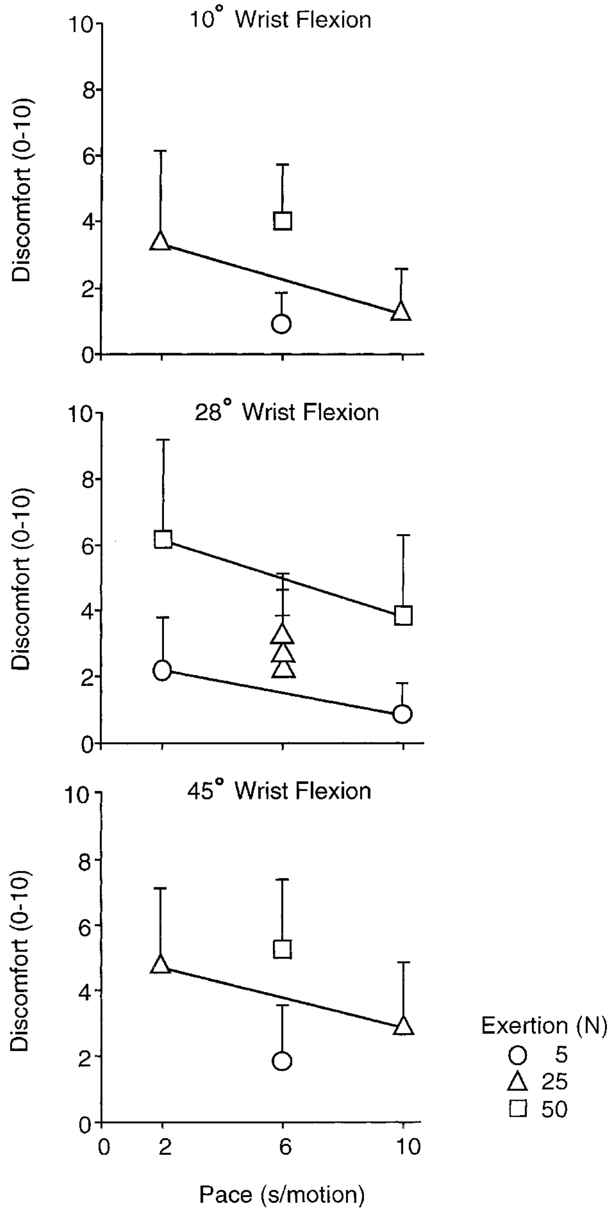

Figure 1. Mean discomfort ratings for all conditions (10 subjects). Error bars indicate one standard deviation.

Table 2. Matrix of pairwise differences.

	Condition											
Condition	1	2	3	4	5	6	7	8	9	10	11	12
2	3·15*											
3	0·97	– 2·18										
4	4·31**	1·16	3·34**									
5	– 0·02	– 3·17*	– 0·99	– 4·33**								
6	3·00	– 0·15	2·03	– 1·31	3·02*							
7	1·30	– 1·85	0·33	– 3·01	1·32	– 1·70						
8	5·29	2·14	4·32**	0·98	5·31**	2·29	3·99**					
9	0·50	– 2·65	– 0·47	– 3·81**	0·52	– 2·50	– 0·80	– 4·79**				
10	1·98	– 1·17	1·01	– 2·33	2·00	– 1·02	0·68	– 3·31*	1·48			
11	2·47	– 0·68	1·50	– 1·84	2·49	– 0·53	1·17	– 2·82	1·97	0·49		
12	3·82**	0·67	2·85	– 0·49	3·84**	0·82	2·52	– 1·47	3·32*	1·84	1·35	
13,14,15†	1·81	– 1·34	0·84	– 2·50*	1·83	– 1·19	0·51	– 3·48**	1·31	– 0·17	– 0·66	– 2·10

*Significant at 0·05 level.
**Significant at 0·01.
†Pooled data from experimental conditions 13, 14 and 15.

effect variable G was therefore introduced into the regression analysis, with data from the current study coded as -1 and the pilot data coded as 1. The resulting discomfort model was:

$$D = 10^{(-0{\cdot}046+0{\cdot}393\ \log E+0{\cdot}225\ \log A+0{\cdot}275\ \log F+0{\cdot}047\ G)} - 1,$$

($r^2 = 0{\cdot}930$, $F(4, 18) = 60{\cdot}196, p < 0{\cdot}01$), where D was discomfort (on a scale between 0 and 10), E was exertion level (N), A was wrist flexion angle (°), and F was frequency (Hz), which is the inverse of pace. The pure error test revealed that there was no significant lack of fit for this linear regression model (F (16,2) = $0{\cdot}574$, $p > 0{\cdot}05$).

Equal discomfort strata were determined by solving the discomfort model for wrist flexion angles as a function of frequency and exertion at a given discomfort level (Radwin *et al.* 1994). The strata for a discomfort level of 3·5 are shown in figure 2 with exertion contours plotted against frequency and wrist flexion angle. Tasks with combinations of exertion, wrist flexion and frequency corresponding to these curves will result in the same level of discomfort.

2.2. *Model validation using published data*

2.2.1. *Data from Snook* et al. *(1995)*: Snook *et al.* (1995) used the method of adjustment to estimate maximum acceptable force for a repetitive wrist flexion task. Fifteen female subjects from an industrial population were instructed to work as hard as they could without developing 'unusual discomfort' at the end of a 7-h session by selecting the maximum acceptable force for wrist flexion between 45° above and below the horizontal at various levels of repetition. Data from that study were substituted into the current study discomfort model by (1) converting paces into frequencies, (2) using maximum acceptable force as the exertion levels, and (3) assuming a 90° wrist flexion for all conditions. Since the discomfort model is logarithmic, expected discomfort levels were estimated by taking the geometric mean discomfort levels among groups. The expected discomfort levels are summarized in table 3.

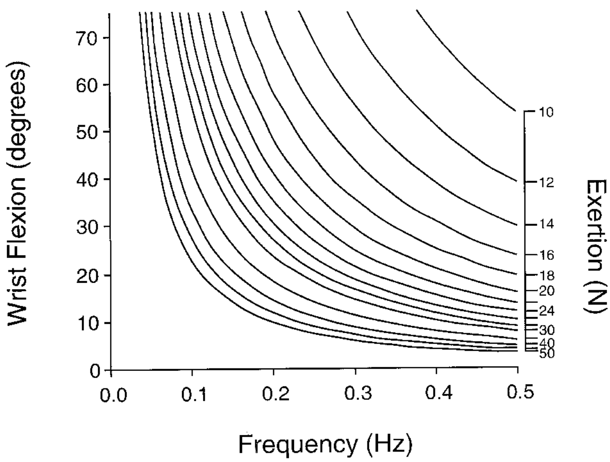

Figure 2. Equal discomfort strata for discomfort level 3·5. Maximum acceptable discomfort strata for 90% of the female subjects in Snook *et al.* (1995) and Marley and Fernandez (1995).

Differences among the expected discomfort levels for conditions between 0·17 and 0·33 Hz, for any given percentage of the female population represented by the 15 female subjects, were within 0·6 units (on a 0 to 10 scale) with coefficients of variation less than 0·05. Mean expected discomfort levels for these conditions were considered the maximum acceptable discomfort (figure 3). The resulting maximum acceptable discomfort for 90% of the female subjects was 3·42 (SD = 0·15). When examining expected discomfort at 0·08 and 0·03 Hz, some discrepancies were observed (table 3).

2.2.2. *Data from Marley and Fernandez (1995)*: Marley and Fernandez (1995) used the method of adjustment to estimate the maximum acceptable frequency (MAF) (motions/min) for a drilling task. Twelve female subjects performed the task for 25 min for each condition and were instructed to work as hard as they could by selecting a 'reasonable' MAF for a presumed 8-h workday with typical rest breaks. Maximum acceptable frequency for wrist flexion of 0°, 25° and 50° were estimated. Subjective ratings of perceived exertion (RPE) were also obtained using the Borg scale. Since sustained exertions were not included in the current study, estimates for the neutral wrist position were not used for validating the current discomfort model.

Table 3. Expected discomfort levels (0 – 10 scale) from fitting data from Snook *et al.* (1995) into the discomfort model (assuming 90° wrist flexion for all conditions).

Percentage of the female subjects	Pace of motion				
	2/minute (0·03 Hz)	5/minute (0·08 Hz)	10/minute (0·17 Hz)	15/minute (0·25 HZ)	20/minute (0·33 HZ)
90	1·81	2·61	3·21	3·49	3·56
75	2·34	3·30	3·99	4·33	4·41
50	2·81	3·90	4·68	5·08	5·18
25	3·20	4·41	5·27	5·72	5·82
10	3·51	4·81	5·73	6·21	6·32

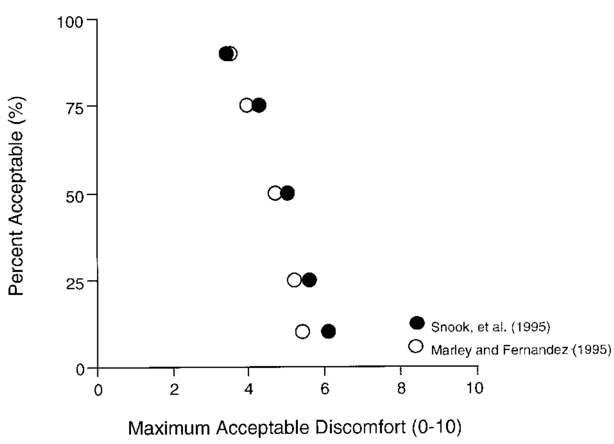

Figure 3. Maximum acceptable discomfort levels estimated for different percentage of the female subjects by fitting data from Snook *et al.* (1995) and from Marley and Fernandez (1995) into the discomfort model.

Maximum acceptable frequencies were estimated for 10, 25, 50, 75, and 90% of the female subjects by assuming that MAF had a normal probability distribution for each condition and adjusting the means with the corresponding standard deviation. Data from their study were substituted into the discomfort model by converting acceptable pace (motions/min) into frequency (Hz) and using the controlled drilling force (54 N) as the exertion level for all conditions. Expected discomfort levels were estimated by taking the geometric mean discomfort levels among groups. The expected discomfort levels are shown in table 4.

Relative equivalent expected discomfort levels were observed between 25° and 50° wrist flexion for the 10th percentile and the 25th percentile female subjects with coefficients of variation less than 0·04. Although expected discomfort levels between these two conditions deviated more for greater percentiles, the differences remained less than 1 on the 10·0 scale (see table 4). Mean expected discomfort levels between the two conditions were used to estimate the maximum acceptable discomfort levels (figure 3).

2.2.3. *Data from Kim and Fernandez (1993)*: Kim and Fernandez (1993) used the method of adjustment for estimating MAF (motions/min) for performing a simulated sheet metal drilling task. Fifteen female subjects performed the task for 25 min for each condition and were instructed to work as hard as possible without 'overexerting' or 'overtiring' themselves by selecting a 'reasonable' MAF for a presumed 8-h shift. Subjective ratings of perceived exertion (RPE) were documented using the Borg scale. Estimates of MAF for various combinations of drilling force (27, 53, 80, and 107 N) and wrist flexion (10° and 20°) were fitted into the current discomfort model by converting acceptable pace (motions/min) into frequency (Hz) and using the required drilling force as the exertion level.

Expected discomfort levels, estimated by taking the geometric mean discomfort levels among groups, and the corresponding RPE are shown in figure 4. A linear regression model was fitted with RPE as the dependent variable and expected discomfort levels as the independent variable. The resulting model was:

$$\mathrm{RPE} = -0.146 + 3.597\hat{D},$$

($r^2 = 0{\cdot}941$, $F(1,6) = 95{\cdot}395$, $p < 0{\cdot}01$), where RPE was subjective ratings of perceived exertion, and $\hat{D}$ was expected discomfort predicted by the discomfort model from the current study.

Table 4. Expected discomfort levels predicted by the discomfort model based on data from Marley and Fernandez (1995) (assuming 54 N exertion for all conditions).

	Wrist flexion angles	
Percentage of the female subjects	25°	50°
90	3·51	3·52
75	3·80	4·11
50	4·36	5·03
25	4·77	5·67
10	4·94	5·90

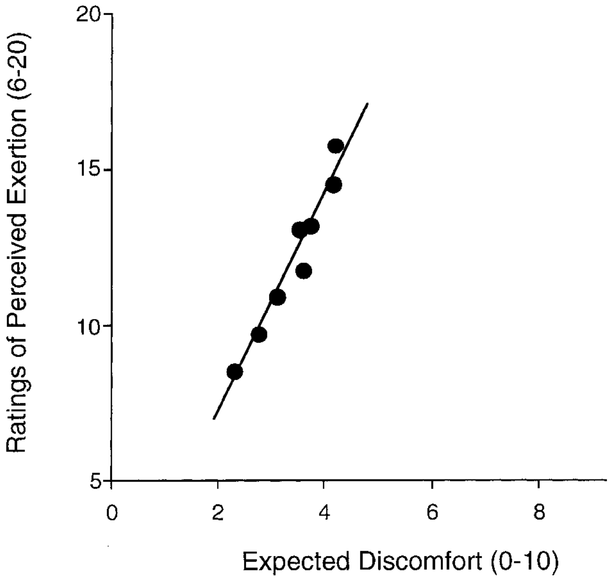

Figure 4. Expected discomfort levels estimated by fitting data from Kim and Fernandez (1993) into the discomfort model and the corresponding subjective ratings of perceived exertion.

2.3. *Derivation of digital filter coefficients and force factor*

2.3.1. *Frequency-weighted filter*: A frequency-weighted angle filter weighs postural signals by the corresponding frequency in proportion to the equal discomfort function, so that the filtered angular data accounts for the relative discomfort associated with motion and repetition. The attenuation slope for the frequency-weighted angle filter is determined by the relative relationship between angle and repetition and can be obtained by algebraically solving the discomfort model at a given exertion and discomfort level. Similarly, a frequency-weighted force filter weighs force signals by the corresponding frequency in proportion to the equal discomfort function and its attenuation slope can be determined by solving the discomfort model at given angle and discomfort level. Based on the discomfort model developed, the attenuation slope is 24 dB/decade for the frequency-weighted angle filter and 14 dB/decade for the frequency-weighted force filter.

The frequency-weighted angle filter was modelled using MATLAB™ (The MathWorks Inc., Natick, MA, USA) as a finite impulse response (FIR) high-pass filter with a 24 dB/decade attenuation slope in the linear region. The difference equation for the FIR filter is:

$$X_F(nT) = \sum_{k=0}^{N} b_k X(nT - kT)$$

where $X_F(nT)$ 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 $X_F(nT)$ of the frequency-weighted filter is the weighted sum of the input for the current sample, $X(nT)$, and the input values for the preceding N samples. Coefficients for the resulting difference equation are listed in table 5. The characteristics of the frequency-weighted angle filter are illustrated in figure 5. The cut-off frequency for

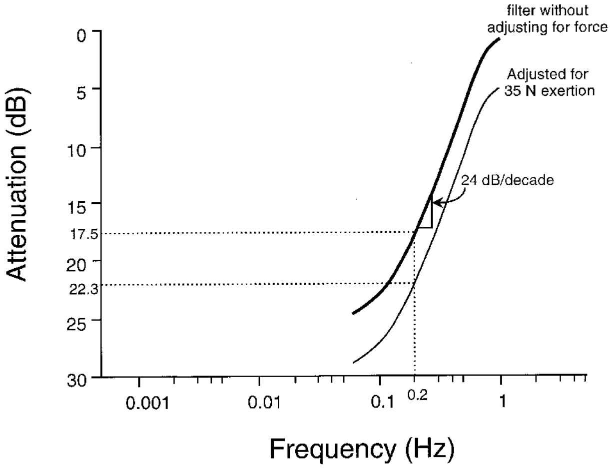

Figure 5. Characteristics of the frequency-weighted filter.

Table 5. Coefficients for the FIR filter.

k	b_k
0	0·0016
1	0·0015
2	0·0012
3	− 0·0000
4	− 0·0031
5	− 0·0089
6	− 0·0179
7	− 0·0301
8	− 0·0448
9	− 0·0607
10	− 0·0760
11	− 0·0888
12	− 0·0972
13	0·9015
14	− 0·0972
15	− 0·0888
16	− 0·0760
17	− 0·0607
18	− 0·0448
19	− 0·0301
20	− 0·0179
21	− 0·0089
22	− 0·0031
23	− 0·0000
24	0·0012
25	0·0015
26	0·0016

this filter was arbitrarily set at 1 Hz because of limitations in the filter design algorithm that was constrained by the attenuation slope and the desired bandwidth. The details for determining the attenuation slope for frequency-weighted filters are discussed in Lin *et al.* (1997).

2.3.2. *Force factor*: Since the frequency-weighted filter does not account for force, filtered angular outputs must be adjusted by a force factor in order to comply with the discomfort model. The adjustment (dB) for a given exertion E was determined by:

$$a_E = A_{E,F} - A'_F$$

where a_E is the adjustment (dB) for exertion E (Newtons), $A_{E,F}$ is the angular attenuation level (dB) for exertion E and frequency F (Hz) based on the discomfort model for a given discomfort level and wrist flexion angle as a reference, and A'_F is the angular attenuation level (dB) for frequency F of the frequency-weighted filter (table 6). The 75° wrist flexion was used as a reference in order to cover the female range of wrist motion (NASA 1978). The difference between $A_{E,F}$ and A'_F can be determined by:

$$a_E = A_{E,F} - A'_F = 20 \times \log\left(\frac{X_{FE}}{X'_F}\right)$$

where X_{FE} was the angle from the discomfort model for frequency F and exertion E, and X'_F was the filter angle without accounting for force. Since the discomfort model is linear and the filter attenuation slope is a constant, the necessary adjustment (dB) for a given exertion is constant and independent of frequency. The linear equation between a given exertion level and its force adjustments for discomfort level 10 was:

$$a_E = 34.88 \log E - 58.67.$$

The adjustment a_E was derived by solving a linear equation between the points ($\log E_1$, a_{E1}) and ($\log E_2$, a_{E2}), where a_{E1} (dB) was the force adjustment for E_1 (N) and a_{E2} (dB) was the force adjustment for E_2 (N).

The force factor for a given exertion level was determined by converting the force adjustment (dB) into a ratio:

$$\mathrm{f}_E = \frac{X_{FE}}{X'_F}$$

Substituting,

$$a_E = 20 \times \log(\mathrm{f}_E)$$

and solving,

$$\mathrm{f}_E = 10^{\left(\frac{a_E}{20}\right)}.$$

Therefore the force factor f_E for exertion level E was equal to:

$$\mathrm{f}_E = 10^{\left(\frac{34.88 \log E - 58.67}{20}\right)}$$

3. Discussion

This investigation showed that main effects of force, wrist flexion angle, and repetition on discomfort were all significant and that no interaction effects were

Table 6. Force adjustment (dB) for given exertion at frequency of 0·2 Hz and discomfort level 10.

Exertion (N)	$A_{E,\ 0\cdot2}$	a_E (dB) = $A_{E,\ 0\cdot02}$ − (− 17·5)
5	− 51·8	− 34·3
10	− 41·3	− 23·8
15	− 35·1	− 17·6
20	− 30·8	− 13·3
25	− 27·4	− 9·9
30	− 24·7	− 7·2
35	− 22·5	− 4·8
40	− 20·3	− 2·8
45	− 18·5	− 1·0

observed. These findings are supported by other studies (Krawczyk and Armstrong 1991, Krawczyk *et al.* 1992, Dahalan and Fernandez 1993, Kim and Fernandez 1993, Ulin *et al.* 1993a, Radwin *et al.* 1994, Marley and Fernandez 1995, Snook *et al.* 1995, Lin *et al.* 1997). The results also indicated that no significant second-order individual effects were observed, suggesting that a linear model was appropriate for describing the individual and combined effects of these factors on subjective discomfort. Subjective ratings of discomfort were shown to be reliable since the three repeated central conditions were conducted more than 1 week apart and no significant difference was observed among subjects.

Lin *et al.* (1997) conducted a 2^3 pilot experiment using fewer experimental conditions and a more limited range. Independent variables in that study were wrist flexion (15° and 45°), pace (3 s/motion and 15 s/motion) and force (15 and 45 N). Seven subjects participated. Since results from that study and the current study were mutually supportive and the two studies used similar designs, data from the two studies were pooled for modelling the relative effects of force, wrist flexion and repetition on subjective discomfort.

The training session that familiarized subjects with the experimental task for all paces could be one aspect of the current study that contributed to the significant difference between the two subject groups. Subjects in the current study were also explicitly instructed to relax between exertions and they were observed to be in compliance. Another difference that may have contributed to the group effect is the additional 5-min session that was included in the current study. Furthermore the current study investigated a wider range of forces. Since the frequency-weighted filters are relative and the relative differences between experimental conditions were consistent among the two groups, these differences should not cause a problem.

The method of adjustment has been used in previous studies in order to establish psychophysically acceptable exposure for various task parameters (Dahalan and Fernandez 1993, Kim and Fernandez 1993, Marley and Fernandez 1995, Snook *et al.* 1995). These psychophysical estimates were used to test the discomfort model under the assumption that subjects maintained a fixed reference (i.e. they worked equally hard for all different conditions) in determining the maximum acceptable exposure levels, so that the selected maximum acceptable exposure would result in approximately the same level of discomfort across different conditions. Based on that assumption, equivalent expected discomfort levels should result across conditions if the current model was valid.

Results from testing the model using data from Snook *et al.* (1995) supported the discomfort model based on the above assumption. Relative equivalent expected discomfort levels were observed among conditions between 0·17 and 0·33 Hz. A notable discrepancy occurred however at 0·08 and 0·03 Hz (table 4) where expected discomfort levels deviated from the equivalent expected discomfort levels. This discrepancy may be due to strength limitations and that excessive exertion alone are sufficient to cause unacceptable discomfort regardless of pace. This explanation was supported by the finding that maximum acceptable force for the two conditions were identical. This result also indicated that the discomfort model might deviate from linearity when certain extremes of physical exposure are reached.

Substituting data from Marley and Fernandez (1995) into the discomfort model showed that expected discomfort levels between 25° and 50° wrist flexion were relatively equivalent for the 10th percentile and the 25th percentile subjects, but expected discomfort started to deviate for greater percentiles. The corresponding RPE revealed that although no significant difference ($p > 0{\cdot}05$) was observed between 25° and 50° wrist flexion, ratings for 50° flexion (mean = 12·58, SD = 1·78) were on the average one level greater than for 25° flexion (mean = 11·50, SD = 2·20) on the Borg scale. However, because of the larger standard deviation in RPE for 25° wrist flexion, after adjusting the mean RPE with the corresponding standard deviation, the difference diminished for the 10th and the 25th percentile subjects and increased for the 75th and the 90th percentile subjects. This observation showed that expected discomfort levels between the two conditions were relatively equivalent when RPE was consistent, and deviated in the same direction as RPE varied. This finding supported the discomfort model by showing that expected discomfort levels were comparable with subjective ratings of perceived exertion.

Since RPE was significantly different among conditions in Kim and Fernandez (1993), the assumption that subjects maintained a fixed reference for MAF was violated. Therefore, instead of validating the discomfort model by identifying equivalent expected discomfort levels, the discomfort model is supported by the high correlation between the expected discomfort and RPE. This validation approach is based on the rationale that a high correlation between the expected discomfort and RPE indicates that the discomfort model is able to predict relative psychophysical response based on biomechanical conditions. This finding agreed with the result from testing the model using data from Marley and Fernandez (1995).

Maximum acceptable discomfort levels suggested by data from Snook *et al.* (1995) and Marley and Fernandez (1995) are very consistent (figure 4). If experimental settings in these two studies represented typical industrial settings for the types of tasks that they investigated, it may be suggested that discomfort level 3·5 may be used to approximate the maximum acceptable discomfort level for 90% of their subjects for repetitive wrist flexion tasks. According to the regression equation between expected discomfort and RPE from testing the model using data from Kim and Fernandez (1993), the maximum acceptable discomfort level for 90% of the subjects for an 8-h workday would result in an RPE of 12, between 'fairly light' and 'somewhat hard' on the Borg scale.

Equal discomfort strata are composed of exposures to combinations of exertion, joint angle and frequency that will result in an equivalent level of discomfort. Equal discomfort strata for the maximum acceptable discomfort for 90% of the female subjects in Snook *et al.* (1995) is shown in figure 2. A task that requires a certain exertion level should have combinations of frequency and wrist flexion on or below the contour for that exertion level.

While information from subjective surveys in the workplace can only be obtained after workers have performed the task for a period of time, it is often biased and confounded with other factors irrelevant to the task itself. A discomfort model, on the other hand, may be used to predict psychophysical responses for a task based solely on biomechanical measurements. Consequently it may be used as an objective method for assessing task design and for potentially identifying problematic tasks. Since the discomfort model describes the individual and combined effects of the three factors on discomfort, it may prove useful for providing quantitative guidance in developing engineering solutions for reducing discomfort in the workplace.

Frequency-weighted filters have been shown to be an effective way for reducing large quantities of biomechanical data into a single value (Radwin *et al.* 1994, Lin *et al.* 1997). The discomfort model proposed here was used for generating parameters for force and frequency-weighted filters. The resulting filter network weighs the magnitude of wrist postural signals recorded by electrogoniometers or other means by the corresponding force and frequency in proportion to the equal discomfort function and enables continuous biomechanical measurements to be filtered and integrated into a quantity that is proportional to relative discomfort. The cut-off for the frequency-weighted filter was arbitrarily set to 1 Hz using a simple filter design. A more sophisticated filter design algorithm might achieve similar frequency characteristics for a greater cut-off frequency.

Although industrial subjects and student subjects might have different perceptions of discomfort (Snook 1978), data from Snook *et al.* (1995) indicated that this model in the current study was consistent with data obtained for industrial subjects. Since the design of frequency-weighted filters depends only on the relative effects, consistent perceptual discrepancies should be insignificant. Future studies will be conducted to verify the current discomfort model using subjects from industrial populations.

While validation of this discomfort model has shown promising results, there are some limitations. The discomfort model was established and tested using a power grip during wrist flexion, which may not be representative for many tasks involving different grip postures (i.e. pinch) or wrist motions (e.g. ulnar and radial deviation). Results from testing the model using data from Snook *et al.* (1995) indicated that the discomfort model deviates from linearity for certain extremes. The discomfort model is also limited because since it is only based on data from 1-h exposure, it is not yet certain if relative discomfort response is linear for different durations of exposure. Future studies will be conducted in order to investigate the duration effect. While a linear model was adequate for describing the relationship among these factors and short-term effects, long-term health effects also need to be investigated.

4. Conclusions

This investigation showed that force, wrist flexion angle, and repetition were all significant main effects for a discomfort model, and that no significant second-order effects among them were observed. A linear model was found to be appropriate for describing the relative relationship among the three factors and discomfort. No significant differences were observed among the three repeated conditions, indicating that the method for subjective ratings of discomfort was relatively reliable.

The discomfort model agreed with data from other studies. This model can be used to shape force and frequency-weighted filters for processing continuous biomechanical data for objective exposure assessment. Although general application of this research is limited to wrist flexion against a resistance using a power grip, the discomfort model describes the individual and combined effects of these factors and may be shown to be useful for quantifying ergonomic conditions and interventions for reducing discomfort in the workplace.

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