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Assessing the physical loading of wearable computers

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Abstract

Wearable computers enable workers to interact with computer equipment in situations where previously they were unable. Attaching a computer to the body though has an unknown physical effect. This paper reports a methodology for addressing this, by assessing postural effects and the effect of added weight. Using the example of arm-mounted computers (AMCs), the paper shows that adopting a posture to interact with an AMC generates fatiguing levels of stress and a load of 0.54 kg results in increased level of stress and increased rate of fatigue. The paper shows that, due to poor postures adopted when wearing and interacting with computers and the weight of the device attached to the body, one possible outcome for prolonged exposure is the development of musculoskeletal disorders.

Keywords: Arm-mounted computer; Ergonomics; Fatigue; Musculoskeletal disorders; EMG; Posture; Perceived exertion; Pain; Discomfort

1. Introduction

Wearable computers are intended to be devices that are attached to the body in such a way that they are considered as being part of the wearer, both by the wearer and by others (Mann, 1997). Wearable computers should be designed to exist within the corporeal envelope of the user (Bass, 1997) and behave as extensions of the user's mind (Mann, 1997). These definitions are useful in terms of understanding the interactional and symbiotic relationship that wearing a computer is intended to have with the wearer, but they fail to define the 'wearing' aspect, with respect to its inherent physical nature.

In defining the term 'wearable' Gemperle et al. (1998) demonstrated the physicality of wearable computers, by highlighting the use of the human body as a support for some product. They went on to state that, the long-term use of wearable computers has an unknown physiological effect on the human body and that it will be important to test their [wearable computers] effect on the human body.

Studies with wearable computers have shown that the weight and positioning of the devices can have a significant effect on the wearer's comfort (Gemperle et al., 1998; Bodine and Gemperle, 2003; Knight and Baber, 2005). Here, the comfort may be measured in terms of the direct physical sensation of the device on the body, which may include some level of pain, but may also include psychological responses such as embarrassment and anxiety. In addition, wearable systems can make the wearer feel awkward, such that they make conscious modifications to their movement (Knight et al., 2002; Knight and Baber, 2005).

Wearable computers may force the wearer to adopt specific postures, firstly so that they are able to view and interact with the wearable device, and secondly as a consequence of the device inhibiting the wearer from assuming their normal posture. Certain postures have been shown to result in localised muscle fatigue (Chaffin, 1973), where long term exposure can have a degenerative effect and result in the development of musculoskeletal disorders (Hagberg et al., 1995; European Agency for Safety and Health at Work, 1999; NIOSH, 1997). It is therefore important to assess the postural effects of wearing and interacting with wearable computers.

There are a number of studies that have demonstrated physiological and biomechanical costs when carrying or wearing loads (Bobet and Norman, 1984; Knapik et al., 1996; Pandolf, 1997). This research however, often focuses on large masses (e.g. >10 kg, 10–40% body weight).

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Wearable computer systems may not weigh as much as this, but as they may be worn on sites other than the trunk (Lin, 1994; Gemperle et al., 1998; Legg, 1985) it is possible that they can still have a detrimental effect on the wearer, especially if they are worn for prolonged periods of time. It is therefore important to determine what weights of wearable devices will not have a detrimental effect on the wearer, and as such be deemed acceptable.

In the first instance, an appropriate weight for a wearable computer may be one that does not result in a significant increase in musculoskeletal loading (Knight and Baber, 2004). In addition, muscle fatigue is considered to be a major factor in the development of muscle damage (Hagberg et al., 1995; European Agency for Safety and Health at Work, 1999). Therefore, a weight of wearable computer that results in an earlier onset, or increased rate of fatigue should also be deemed inappropriate.

2. Risk assessment of wearable computers: applied to arm-mounted computers

Based on the two risk factors of posture and weight of the wearable computer, this paper proposes a seven-stage risk assessment to determine if, and how, wearable computers pose risks for developing musculoskeletal disorders. In discussing these assessment activities, the paper uses an example of an assessment of arm-mounted computer (AMC) technologies.

Stage 1. *Review of the wearable computer*: The first aspect of the assessment involves a review of the type of wearable computer in question. This includes a description of where and how the technology is worn, how it is used and what it is used for. For arm-mounted computers this review highlights that the technology appears to be developing along two fronts. The first is to fit as much technology as possible into products with the form factor of a wristwatch (e.g. Narayanaswami and Raghunath, 2000). The second is to take desktop or portable computer devices and adapt them to be worn on the arm (e.g. TekGears forearm keyboard, Orang-Otang Computer Inc.'s 'Peel-It' which mounts a Palm Pilot^{TM} on the wrist, and BT's 'Office on the arm'). In both cases, the technology is designed to be around the wrist of one arm (usually the non-dominant arm) so that it can easily be brought into view and enable interacting with the other hand.

Stage 2. *Review of the biomechanics of the body segment to which the wearable computer is attached*: This review includes a discussion of the anatomy of the body segment to which the device is attached and highlights the muscles involved in movement and helps determine the structures that might be stressed by the attachment of a wearable computer. For AMCs this review involved understanding the movements of the arm about the shoulder, including the movements of the shoulder girdle in flexion, extension, abduction, adduction, lateral flexion and extension, and medial and lateral rotation. It also included the articulation of the arm about the elbow, as well as pronation and supination of the forearm. For an assessment of muscular control the review highlighted the muscles involved in arm movement, specifically highlighting the prime movers for arm flexion and abduction (e.g. the different aspects of the deltoid) elbow flexion (e.g. biceps brachii, brachialis and brachioradialis) and pronation (pronator quadratus).

Stage 3. *Literature review of the postural risk factors of the body part or segment to which the wearable computer is attached*: A review of the postural risk factors for the area of the body to which the wearable is attached will highlight the postures that have been found to result in localised muscle fatigue and incidences of musculoskeletal disorder. For the arm, considerable research has shown that abducted and flexed arm positions are detrimental and should not be adopted for prolonged periods of time (Chaffin, 1973; Herberts et al., 1980; Bjelle et al., 1981; Hagberg, 1981; Kilbom et al., 1986; Aarås et al., 1988; Järvholm et al., 1988; Wiker et al., 1989). In addition, pronation of the forearm can result in significant increases in forearm muscle activity (Zipp et al., 1983) and discomfort (O'Sullivan and Gallwey, 2005). This review in stage 3 will relate considerably to stages 4 and 5.

Stage 4. *Determine factors inherent in the wearable computer that requires the adoption of specific or altered postures*: This paper proposes that wearable computers have the potential to generate postural risk factors by two methods. The first is by forcing the user to adopt postures so that the wearer is able to view or interact with the device. The second is as a consequence of the device inhibiting normal performance in some way such that the wearer has to modify their posture or action.

For AMCs a quickly apparent postural effect is that the wearer will need to adopt a specific posture to enable interaction with the device. For a device worn around the wrist or on the forearm, an AMC interaction position will require the wearer to bring their arm out in front of their body so that they can see and display and interact with it. This will invariably involve the arm being abducted and flexed at the shoulder, the elbow flexed and the arm rotated so that the forearm is horizontal and parallel to the chest. It may also involve some forearm pronation to turn the display to the face.

Stage 5. *Determine if these postures (from stage* 4*) are pathomechanical*: By combining stage 3 with stage 4 it may be possible to determine if the postures adopted when using the wearable have been found to exert a significant postural stress on the body.

A posture that could be adopted to interact with an AMC is one that involves up to 45° of arm flexion and abduction, with the elbow flexed at 90°, and the upper arm medially rotated so that the forearm is horizontal and lies parallel to the chest. With this posture, the level of upper arm flexion and abduction is one that is greater than the levels that have been found to result in physically detrimental effects as determined by Chaffin (1973), Kilbom et al. (1986), Aarås et al. (1988), Järvholm et al. (1988) and Dul (1988). However, it is below the 60° level proposed as a possible border of acceptability by the European Agency for Safety and Health at Work (1999) and NIOSH (1997).

However, it may be that the postures required to interact with a wearable computers are novel with respect to those assessed in the musculoskeletal disorder research. For AMC interaction, the posture adopted requires some degree of arm flexion and abduction and so places some stress on the musculoskeletal system of the shoulder and arm. However, as the elbow is flexed at around 90° the hands are closer to the chest, which reduces the adductor and extensor torques generated by the weight of the arm about the shoulder, compared to the torques generated when the arms are straight. In addition, the possible pronation of the forearm may also impart some significant stress. Indeed, pronation of the forearm may affect the effectiveness of the biceps brachii as an elbow flexor, shifting much of this load to the brachialis and brachioradialis (Rasch and Burke, 1978).

None of the studies referenced above have analysed specific postures similar to the proposed AMC interaction position. It may be that the interaction posture is unique with respect to those currently used when working. Therefore, determining whether it places the wearer under a level of musculoskeletal stress that will fatigue the system and possibly result in the development of disorders based on the literature may not be appropriate.

Stage 6. *Assess the effects of added weight to the body segment to which the wearable computer is attached*: In an AMC interaction position, Fig. 1 shows that the weight of the upper arm (Wa) and forearm (Wf) generate extensor and adductor torques about the shoulder that must be counteracted by flexor and abductor muscles, such as the anterior or frontal aspect of the deltoid (Mdf) and the lateral or medial aspect of the deltoid (Mdl), and that the weight of the forearm also generates an extensor torque about the elbow, which must be counteracted by flexor muscles, such as the biceps brachii (Mb). Attaching weight to the arm in the form of an AMC (Wl) increases the workload on these muscles. In addition, as an AMC is attached at the distal end of the arm (i.e. wrist, hand), the moment arms of the load (Wl) about the centres of rotation at the shoulder and elbow are maximised, and as such, so are the adductor and extensor torques generated by the weight of the AMC, against which the abductor and flexor muscles must counteract. A simple design solution for this situation would be to move the load further up the arm, which would reduce the torques generated by the AMC, however in this position, it may not be easy to see or interact with the device.

Stage 7. *Determine what weight results in a significant increase in musculoskeletal loading and increased rate of fatigue*: A review of the literature may suggest acceptable weights for loads attached to the body, specifically for loads attached to the back (Knapik et al., 1996) and head (Knight and Baber, 2004). For AMCs, previous research has shown that wrist weights ranging from 0.45 to 4.8 kg have resulted in a significant increase in energy expenditure when walking and running (Bhambhani et al., 1989; Miller and Stamford, 1987; Claremont and Hall, 1988; Graves et al., 1988). Studies on hand tools have shown that increased hand held weight results in an increase in musculoskeletal stress (Ulin et al., 1993, Putz-Anderson and Galinsky, 1993) and has resulted in recommendations that hand tools should not be heavier than 0.4 kg (Wiker et al., 1989) and 0.45 kg (Chaffin, 1973). However, as highlighted by Putz-Anderson and Galinsky (1993), of more importance than the weight of the tool is the force required to perform the task. In that, the user often has to impart considerable force to the tool in addition to that required to maintain it in position (e.g. when screwing or hammering). For AMCs though, it is assumed that the magnitude of force exerted by the muscles of the arm supporting the AMC is determined predominantly on the level required to hold it in position. The additional force required when interacting with the device (i.e. during typing or pressing on a touch screen) is assumed to be negligible. Therefore, basing recommendation of AMC weight on tool weight data is inappropriate.

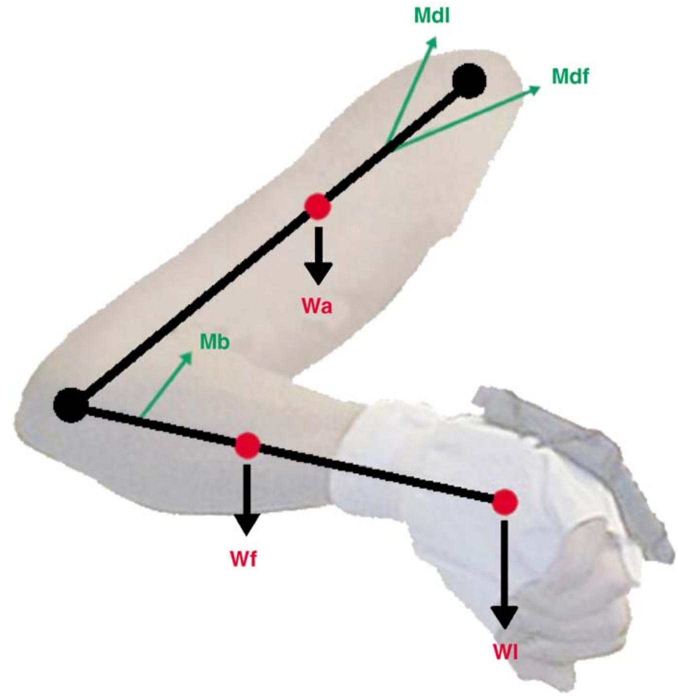

Fig. 1. Model of the AMC interaction position. Wa and Wf represent the weight of the upper arm and forearm, respectively, and Wl represents the weight of the AMC. These generate, extensor and adductor moments at the shoulder that must be counteracted by shoulder flexors Mdf (e.g. anterior aspect of the deltoid) and shoulder abductors Mdl (e.g. lateral aspect of the deltoid), respectively. Wf and Wl also generate elbow extensor moments that must by counteracted by elbow flexors Mb (e.g. biceps brachii).

3. Assessment of AMC interaction position and AMC weight

The risk assessment of AMCs highlighted that because the AMC interaction position is novel with respect to working postures, and because issues of mounting weight on to the forearm have not been previously addressed, determining the effect of wearing and interacting with AMCs is difficult with current knowledge. To address these issues, two experiments were carried out. The first examined an assumed AMC interaction position, and the effect of adding weight to the forearm. The second experiment measured muscle activity, arm posture and perceived pain and discomfort when users interacted with AMCs of different weights.

3.1. Study 1: AMC interaction position and added forearm load

A method of normalising postural load is to refer the level of loading to a maximal level that can be voluntarily achieved. Using maximal voluntary contraction (MVC) methods, attempts have been made to determine a load level that will allow an unlimited duration of contraction without risk of injury. Jörgensen et al. (1988) for example, demonstrated that muscle fatigue might be elicited by sustained isometric contraction at 5–10% MVC and Jonsson (1982) has suggested that work load should not be greater than 2–5% MVC for a prolonged period of time.

The aim of this study was to determine the level of muscle activity generated to hold the arm in a static AMC interaction position by referring magnitudes of test EMG to those elicited when performing a MVC. In addition, this study aimed to assess the affect of adding weight to the forearm and determine the weight at which the increase in muscle loading becomes significant and exceeds the recommendations of Jörgensen et al. (1988) and Jonsson (1982).

Participants: Seven postgraduate research students (5 male, 2 female) from the University of Birmingham participated in the study. They had a mean age of 27 ± 3 years and weight of 73 ± 11 kg. None of the subjects reported any symptoms of musculoskeletal discomfort in the upper body prior to testing.

Experimental set-up: Each participant sat on a standard chair without arm rests. Surface EMG electrodes were placed over the belly of the biceps brachii, anterior aspect of the deltoid (deltoid-anterior) and medial aspect of the deltoid (deltoid-medial) of the right arm. The electrodes were bipolar, silver, surface electrodes (10 mm diameter, 17 mm centre to centre), which were part of a skin-mounted pre-amplifier ($\times 1000$) encapsulated in araldite (Johnson et al., 1977). The electrodes have a common mode rejection ratio at 50 and 500 Hz of 100 and 80 dB, respectively, with a filter pass-band from 10 to 1000 Hz. The noise to signal ratio of the EMG signals was $<5\%$. EMG signals were fed via a variable amplifier to separate channels of an analogue to digital converter (Computer Boards, Inc. A–D Board C10-DAS801). The signals were sampled at 2 kHz and fed into Microsoft® Excel by means of a DASWizard computer package on a Pentium II computer for storage and analysis. Further processing of the signals took place off-line.

Prior to testing each subject performed maximal voluntary contractions (MVCs) for each of the three muscles. For each MVC condition, EMG was recorded for 8 s during which the subject was requested to maintain the MVC for a minimum of 3 s and a maximum of 5 s. MVCs for the biceps and deltoid-anterior were performed as the elbow was flexed at 90° and the forearm was positioned horizontally in front of the chest parallel to the frontal plane at a flexed arm angle of approximately 45°. EMG recordings of the biceps and deltoid-anterior were recorded as the subjects attempted to flex the elbow and flex the arm, respectively, with maximal force against a fixed resistance. MVCs for the deltoid-medial were performed as the elbow was flexed at 90° and the forearm positioned horizontally parallel to the sagittal plane at an abducted arm angle of approximately 45°. EMG was recorded as the subject attempted to abduct the arm with maximal force against a fixed resistance. A rest period of 2 min between MVCs was used to allow for recovery (De Luca, 1997).

To determine a value for the MVC, the EMG data samples were full wave rectified and smoothed with a sliding window of 1-s using a macro written in Microsoft® Excel. The greatest value of the rectified and smoothed data was taken as the MVC (De Luca, 1997).

Conditions and protocol: The testing procedure involved the participants holding their arm in a static AMC interaction position for 10 s, during which samples of EMG were recorded. The AMC interaction position involved the participants holding their forearm horizontally, in front of the body and parallel to the chest. The upper arm was flexed and abducted by 45°. The participants adopted this test position under 7-arm load conditions, which attempted to replicate the weight of a proposed AMC. The seven load conditions were: no added load, 0.34, 0.68, 1.02, 1.36, 1.70 and 2.04 kg. The loads used were lead weights sewn into padded wrist belts attached around the wrist with Velcro. The order of the load conditions was randomised with an inter-condition rest period of 5 min.

EMG analysis: The test EMG was processed using a root mean square (RMS) macro written in Microsoft® Excel 97 SR-1. A normalised EMG value was then calculated for each muscle under the 7 load conditions by expressing the RMS value as a percentage of the MVC.

3.1.1. Results

With an unloaded forearm, holding the arm in an AMC interaction position required similar levels of muscle activity for the biceps ($7 \pm 3\%$MVC) and the deltoid-medial ($8 \pm 3\%$MVC), which were greater than the 5% cut-off of

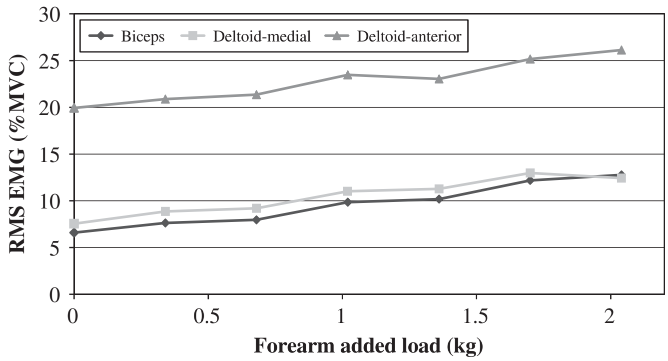

Fig. 2. Arm EMG as the arm is held in the AMC interaction position with added forearm load.

Jonsson (1982). For the deltoid-anterior the level of activity was $20 \pm 6\%$MVC with an unloaded forearm, greater than double the level of the other muscles and greater than the 10% cut-off of Jörgensen et al. (1988).

Adding weight to the forearm resulted in a linear increase in EMG for all three muscles (Fig. 2), where the level of contraction can be approximated to: $3.14 \times$ added armload + 6.4 MVC for the biceps, $2.62 \times$ added armload + 7.8 MVC for the deltoid-medial and $3.03 \times$ added armload + 19.8 MVC for the deltoid anterior. A load of 1.02 kg resulted in the levels of EMG for the biceps and the deltoid-medial joining the deltoid-anterior in exceeding the 10%MVC cut-off. For all three muscles the increase in muscle activity due to added forearm load became significant with a load of 1.7 kg [biceps: $t(6) = 3.4$, $p < 0.05$; deltoid-medial: $t(6) = 2.8$, $p < 0.05$; deltoid-anterior: $t(6) = 3.9$, $p < 0.05$].

3.2. *Study 2: responses to interacting with AMCs of different weights*

Study 1 showed that, in an assumed AMC interaction position, fatiguing levels of muscle activity were generated, and that the levels of activity became significantly greater when a load of 1.7 kg was added to the forearm. A second experiment was designed to measure the effects of actually interacting with AMCs of different weights. The aim was to determine, if when interacting with these AMCs, participants demonstrate any symptoms of fatigue. In addition, it was perceived by the experimenters, that a forearm load of 1.7 kg was excessive for a cut-off for AMC load. Therefore, the experiment aimed to determine if, using the criterion of an increased rate of fatigue resulted in a lower cut-off weight.

Participants: Fifteen postgraduate students (8 male, 7 female) from the University of Birmingham participated in the study. They had a mean age of 24 ± 3 years and weight of 69 ± 11 kg. None of the subjects reported any symptoms of shoulder or arm discomfort prior to testing.

Protocol: During each testing session the participant sat on a standard chair without arm rests and was required to perform a simple data input task onto an AMC. The AMC was a Palm^{TM} VIIx attached to the back of a glove and worn on the participants' non-dominant hand enabling the dominant hand to be free for AMC interaction.

The data input task required the participants to view a computer monitor onto which a series of single words was displayed. Within the series of words were numbers displayed in text (e.g. 'one', 'fifteen', etc). When a number was recognised, the participant had to tap the number onto the touch screen numeral pad of the palm pilot, using a stylus, which was held in the dominant hand. The words used came from a list of 100, and included the numbers 'zero' up to and including 'twenty', and 79 other non-numeral words. The words were displayed in a randomised order, grouped into sets of 120, and presented with an inter-word refresh rate of 1 s. Each testing session consisted of 5 sets. Each set lasted 2 min, with an inter-set rest period of 30 s.

The data input task was performed under five conditions, to simulate different weights of AMC. These conditions were the Palm^{TM} VIIx (weighing 0.20 kg) only, and the Palm^{TM} VIIx with wrist weights to give total forearm loads 0.54, 0.88, 1.56 and 2.24 kg. The wrist weights used were of the same design as those used in study 1. The order the conditions were undertaken was randomised with testing sessions performed on separate days.

Measures of fatigue: As a result of increased action potential firing rate or the recruitment of additional motor units, increases in EMG amplitude over time is often used as an indicator of fatigue (Herzog et al., 1994). The Borg CR-10 score has been reliably used in assessing perceptions of pain and discomfort due to loading of joint structures (Harms-Ringdahl et al., 1986) and by assessing changes with time has been used as an index of fatigue (Oberg et al., 1994, Knight and Baber, 2004).

Fatigue has been defined as the inability to maintain the expected force or power output (Edwards, 1981) and as any reduction of the force generating capacity of the total neuromuscular system, regardless of the force required in any given situation (Bigland-Richie, 1984). As such, so that the arm is less flexed and abducted and the torques generated by the weight of the arm are less, lowering the arm over time could be inferred as being a postural response to a fatiguing musculoskeletal system.

This study therefore used increases in the amplitude of shoulder and arm EMG and perceived pain and discomfort, and decreases of arm flexion and abduction angles, over time, as indices of fatigue.

Electromyography: As well as the same muscles analysed (biceps brachii, deltoid-anterior, deltoid-medial) the same equipment and protocol was used for recording EMG in this study as used in study 1.

Due to equipment availability, EMG could not always be recorded from all three muscles in the same testing session. The outcome being that for each muscle 8 participants provided samples of EMG.

Due to the concern of the motivational factor required in generating a MVC, and as test conditions were performed on separate days, a sub-maximal contraction of fixed force was chosen to elicit a sample of EMG that could be used to normalise the test EMG. Prior to each testing session, a calibration contraction for the biceps was recorded as the participant held a 2.2 kg dumbbell at 90° of elbow flexion, with the upper arm against the side of the body, for approximately 5 s. Calibration contractions for the deltoid-medial and deltoid-anterior were recorded as the participant held the 2.2 kg dumbbell at 90° of shoulder abduction and shoulder flexion, respectively, for approximately 5 s.

Samples of EMG were recorded in 8-s blocks at the start (in the initial 0–10 s) and end (110–120 s) of each set, and subsequently normalised. The test EMG was processed using a root mean square (RMS) macro written in Microsoft® Excel 97 SR-1. A normalised EMG value was then calculated, for each muscle under the 5 load conditions, by expressing the RMS value as a percentage of the calibration EMG.

Arm posture: Arm position was measured with respect to angular displacement at the shoulder and the elbow using a Biometrics XM180 bi-axial strain gauge goniometer. The goniometer was positioned over the shoulder such that the proximal end lay above the shoulder parallel to the frontal plane and the distal end lay along the long axis of the upper arm parallel to the frontal plane. In this position, one axis of the goniometer measured angular displacement in the frontal plane as the arm abducted. The other axis though, measured movement in a non-specific plane. If the arm is held by the side, the goniometer measures the angular displacement of the arm through flexion. However, if the arm were abducted by 90°, the goniometer now measures angular displacement as the arm horizontally flexed. The angle measured by the second axis of the goniometer was thus the angular displacement of the arm in a plane perpendicular to the abducted arm position in the frontal plane. This new angle involves displacement in the sagittal plane, and as such it can be used to determine displacement of the arm in front of the body. Therefore, although it is not a measure of pure arm flexion, it was still considered to be a measure of flexion. Prior to testing, the goniometer was calibrated to zero, when the arm was held tight along the side of the body.

To measure the elbow angle, the distal end of the goniometer was attached along the long axis of the upper arm and the proximal end attached along the long axis of the lower arm. The goniometer was calibrated to zero, when the elbow was held at full extension. For both the shoulder and the elbow, the ends of the goniometer were attached to the skin using surgical tape.

Placing a biaxial electro-goniometer over the shoulder to measure abduction and flexion angles offers a potential source of error. Due to medial rotation of the upper arm, when adopting the AMC interaction position, the goniometer twists along its long axis. This means that the abduction and flexion angles given by the goniometer may not be pure angular displacements in one plane and may incorporate an aspect of movement in a perpendicular plane. With respect to this source of error, the angular displacement must be interpreted cautiously.

Due to equipment availability the participants were split into two groups, one group of 8 participants had measurements of angular displacement about the shoulder taken, the other group of 7 participants had measurements of angular displacement about the elbow taken. Measurements of angular displacement were taken from a data reader at the start (in the initial 0–10 s) and end (110–120 s) of each set.

Perceived pain and discomfort (PPD): While performing the tests, the participants were asked to rate their perceptions of PPD around the shoulder and arm using the Borg CR-10 scale. Borg values for the forearm, upper arm, shoulder and neck were taken during the 30 s rest period between each set and at the end of the test.

3.2.1. Results

Using the 0.20 kg condition as a baseline, effects due to AMC weight were determined by paired comparisons with the other AMC load conditions. Effects due to time were determined on two levels. Between set effects, used the first set as a baseline and compared it with subsequent sets. Within set effects, compared the results recorded at the start and end of each set. Statistical significance was set at $p<0.05$.

Electromyography: The EMG data shown in Table 1 represents the group mean value for the percentage change in RMS EMG for each data point with respect to the RMS EMG value for the 0.20 kg AMC condition at the start of set 1. In representing the data this way, relative comparisons between the AMC weight conditions are easier. It must be noted that statistical analysis was carried out on the normalised data only, before it was adjusted to represent a relative change.

Increases of AMC weight resulted in increases in RMS EMG for all three muscles, though they only became significant from the 0.20 kg AMC condition when the AMC weight was 1.56 kg, and only for the deltoid-anterior and the biceps muscles.

There were apparent increases in RMS EMG due to time, between and with sets, for all three muscles, specifically for the higher load conditions. However, these increases were only significant for within-set time effects for the deltoid-medial when the AMC weight was 0.54 kg or greater.

Arm angle: Across all conditions and throughout the tests, the average arm position was one where the upper arm was abducted by $25\pm7°$ and flexed by $17\pm7°$, and the elbow was flexed by $95\pm6°$. For arm position there were no significant effects due to AMC weight, but there were for time effects (Table 2). Between sets and within sets, the general trend was for the arm abduction and flexion angles to decrease due to time. The greater the weight of the AMC, the more the arm angles decreased.

Perceived PPD: For all four areas of the body assessed (forearm, upper arm, shoulder and neck), an increase of

Table 1
RMS EMG when interacting with an AMC

	0.2 kg AMC				0.54 kg AMC				0.88 kg AMC				1.56 kg AMC				2.24 kg AMC			
	Start		End		Start		End		Start		End		Start		End		Start		End	
Set	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Biceps													*L*				*L*			
1	0	0	0	14	27	27	22	36	27	70	23	51	28	39	59	71	140	84	134	106
2	−2	12	−3	14	17	25	17	32	15	68	28	57	31	43	71	72	128	88	142	129
3	1	19	−2	18	24	43	19	35	20	62	35	43	24	37	68	63	146	80	217	128
4	−3	17	−5	21	13	28	29	63	17	57	19	43	33	31	80	67	185	101	224	120
5	0	9	−1	13	13	28	22	34	13	56	21	46	51	57	85	63	215	111	248	142
Deltoid–medial																				
1	0	0	4	6	33	75	38	78	39	55	54	62 *w*	61	120	74	96	84	55	123	69 *w*
2	−7	10	−3	12	16	63	42	85 *w*	38	70	59	78 *w*	20	39	62	68 *w*	77	80	84	41
3	2	22	7	18	16	58	39	77 *w*	45	79	72	87	18	44	59	70 *w*	70	80	138	105 *w*
4	−2	17	−3	25	19	63	83	112 *w*	56	109	92	111 *w*	15	40	62	70 *w*	81	96	141	90 *w*
5	−5	16	10	10	13	46	48	68 *w*	64	116	100	125 *w*	42	55	87	85 *w*	88	104	164	110 *w*
Deltoid−anterior													*L*				*L*			
1	0	0	4	10	28	36	27	34	21	54	29	65	47	32	36	34	49	40	41	49
2	−11	15	−4	11	25	28	29	34	12	64	19	60	40	30	39	36	32	27	45	41
3	−6	15	−1	17	34	36	32	43	16	57	22	52	48	42	53	51	25	31	34	51
4	2	24	−4	24	46	48	48	57	11	50	16	51	44	45	46	48	34	34	57	62
5	−2	20	−1	16	38	57	33	54	16	48	27	66	65	48	47	64	41	39	59	61

Table 2
Arm posture when interacting with an AMC

Set	0.2 kg AMC				0.54 kg AMC				0.88 kg AMC				1.56 kg AMC				2.24 kg AMC			
	Start		End		Start		End		Start		End		Start		End		Start		End	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Abduction																				
1	28	4.9	27	4.6	29	6.5	28	7.8	31	7.0	29	5.9 *w*	28	9.1	25	8.7 *w*	26	3.5	22	1.9 *w*
2	27	4.4	26	4.2	27	7.0	26	6.9 *b*	28	7.1	27	7.3 *b*	25	9.7	22	8.1 *b*	23	2.5	20	3.4 *w*
3	26	5.1	26	5.2	27	6.9	25	7.2 *w*	27	7.2	25	6.5 *bw*	25	11.2	24	10.4	20	2.1	18	3.1 *bw*
4	26	5.4	26	6.0	27	6.9	25	7.1	25	7.2	23	7.5 *bw*	25	11.0	21	9.6	20	4.4	17	5.2 *bw*
5	25	5.6	24	6.4 *w*	25	7.3	24	7.9	24	8.4	24	8.8 *b*	24	10.9	20	8.1 *w*	19	5.7	17	6.0 *bw*
Flexion																				
1	22	6.4	20	8.5	18	6.2	16	5.1 *w*	20	5.7	18	5.2	19	4.7	16	4.4 *w*	21	3.6	15	3.7 *w*
2	23	8.0	21	8.6	16	6.0	15	5.6 *w*	19	4.8	17	5.0 *w*	17	3.9	12	4.9 *bw*	18	3.4	15	3.6
3	21	7.3	20	8.3	17	6.3	14	7.0 *w*	18	3.1	15	2.7 *w*	15	4.1	11	3.6 *bw*	17	4.9	13	4.7 *bw*
4	21	7.4	20	7.0	16	7.2	14	7.1	19	4.0	16	3.9 *w*	15	5.0	11	4.5 *w*	17	2.9	13	6.4 *w*
5	21	4.9	20	6.2 *w*	16	8.6	13	8.5 *w*	17	2.7	15	2.8	15	4.8	11	4.8 *b*	16	3.3	11	2.8 *w*
Elbow																				
1	95	5.4	95	4.4	93	5.5	96	5.5	95	6.0	96	6.4	96	6.1	97	6.1	98	6.0	97	4.7
2	94	6.3	95	6.4	93	5.7	94	5.5	95	5.2	97	5.4	96	5.9	96	5.4	99	5.3	100	4.9
3	95	6.4	94	6.6	93	6.0	91	7.0	95	6.4	96	5.4	96	6.1	96	5.7	97	5.3	98	6.3
4	93	6.0	93	5.9	94	5.9	95	7.0	95	5.8	96	5.7	94	6.8	96	5.2	97	5.8	97	4.9
5	95	6.5	94	6.3	94	5.5	96	5.3	96	6.1	96	7.1	96	5.1	97	5.2	97	5.6	98	5.1

b indicates a significant increase between sets, from set 1; *w* indicates a significant increase within the set.

AMC weight from 0.20 to 0.54 kg resulted in a significant increase in Borg CR-10 score (Table 3). Further significant increases due to AMC weight came when the AMC weight was 1.56 kg for the forearm, upper arm, and the shoulder, and 2.24 for the neck. In the 0.20 kg AMC condition, the only significant increase in Borg CR-10 score due to time was the increase given for the upper arm by the end of the third set. In the 0.54 kg AMC condition, there were

Table 3
Perceived pain and discomfort when interacting with an AMC

Set	0.2 kg AMC		0.54 kg AMC *L*1		0.88 kg AMC		1.56 kg AMC *L*2		2.24 kg AMC	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Fore arm			*L1*				*L2*		*L3*	
1	0.5	0.7	1.1	1.0	1.0	1.1	1.9	1.2	2.7	2.1
2	0.6	0.7	1.3	1.1	1.5	1.4 *t*	2.4	1.4 *t*	3.2	2.4
3	0.7	0.9	1.6	1.4	1.8	2.0	2.4	1.9	3.5	2.5
4	0.8	0.8	1.7	1.5	1.8	1.6	3.3	2.3	3.5	2.6
5	0.9	0.9	1.7	1.4	1.6	1.5	3.3	2.8	3.8	2.9
Upper arm			*L1*				*L2*		*L3*	
1	0.7	0.7	1.5	1.2	2.0	1.6	2.5	1.4	3.5	2.1
2	0.8	0.5	2.2	1.5 *t*	2.2	1.5	3.0	1.8 *t*	4.4	2.2
3	1.1	1.0 *t*	2.5	1.6	2.8	2.0 *t*	3.9	1.8	4.9	2.6
4	1.3	0.9	3.0	1.9	3.4	2.5	4.3	1.7	5.1	2.7
5	1.5	1.0	2.9	1.6	3.7	2.3	4.6	2.3	5.5	2.8
Shoulder			*L1*				*L2*			
1	0.5	0.7	0.8	0.9	1.5	1.2	2.1	1.1	3.2	1.9
2	0.6	1.0	1.3	1.4 *t*	2.1	1.2 *t*	2.9	1.1 *t*	3.7	1.9
3	0.7	1.0	1.5	1.5	2.3	1.9	3.6	1.9	4.5	2.6
4	0.8	1.1	1.8	1.4	3.0	2.5	4.0	1.7	4.4	2.5
5	0.8	1.0	2.3	1.5	3.1	1.5	4.4	1.8	5.0	2.4
Neck			*L1*						*L2*	
1	0.0	0.1	0.3	0.6	0.3	0.5	0.6	1.1	0.8	1.1
2	0.1	0.3	0.7	0.9 *t*	0.7	0.9	1.3	1.7 *t*	1.7	1.8
3	0.2	0.5	0.9	1.1	1.1	1.5 *t*	1.7	1.9	2.1	2.6
4	0.4	0.9	1.0	1.2	1.4	1.4	2.2	2.4	2.3	2.6
5	0.4	0.9	1.2	1.3	1.5	1.2	2.6	2.7	2.5	2.7

*L*1 indicates a significant increase from the 0.2 kg AMC condition. *L*2 indicates a significant increase from *L*1. *L*3 indicates a significant increase from *L*2. *t* indicates a significant increase from set 1.

significant increases for all areas except the forearm. For the greater AMC weight conditions, the general trend was for the increases in Borg CR-10 score to become significant by the end of the second set.

3.3. Discussion of results

Wearing and interacting with an AMC has the potential to affect the musculoskeletal system of the whole arm and shoulder girdle. In the current study, participants experienced pain and discomfort in the neck, shoulder, upper arm and forearm. These perceptions increased due to time, as did levels of muscle activity in the shoulder and upper arm muscles. In addition, participants made postural adjustments, lowering the arm as time went on. These responses suggest that, in these conditions, the musculoskeletal system was fatiguing.

In the palm pilot only (0.20 kg) condition there were few signs of fatigue, though there were some. The EMG data did not suggest fatigue, but a significant decrease in abduction angle and increased upper arm Borg CR-10 score suggest that perceptions of fatigue were being experienced. If the palm pilot only condition displayed signs that could tentatively imply that the musculoskeletal system was fatiguing, then the 0.54 kg condition showed signs that clearly indicate that the system was fatiguing. With an AMC of 0.54 kg, deltoid-medial RMS EMG increased within sets, arm flexion and abduction angles decreased within sets, and between sets for abduction, and perceptions of exertion increased with time in the upper arm, shoulder, and neck. When the AMC weight increased to 0.88 kg the signs became more numerous, including signs of fatigue developing in the forearm region.

It was not until the AMC weight was 1.52 kg that a significant increase in EMG activity between conditions became significant. These results show that using the criterion of a significant increase of EMG as a cut-off for AMC weight may induce type II errors. Using this criterion, AMC weights of up to 1.5 kg would be considered safe, whereas using a significant increase in the rate of fatigue as a criterion, suggests that the weight of an AMC should be less than 0.54 kg.

4. Generating guidelines for wearable computer musculoskeletal risk assessment

Workers who are predisposed to mechanical stresses, due to the posture constraining and load inducing nature of their working environments, are often recommended to take frequent rest breaks to allow their bodies to recuperate

Table 4
Summary review of risk assessment of arm mounted computers

Risk assessment	Activity	Wearable computer
Stage 1	Review of the wearable computer	Arm Mounted Computers (AMCs)
Stage 2	Review of the biomechanics of the body segment to which the wearable computer is attached	Shoulder and arm musculoskeletal system
Stage 3	Literature review of the postural risk factors of the body part or segment to which the wearable computer is attached	Arm abduction and flexion linked to disorders
Stage 4	Determine factors inherent in the wearable computer that requires the adoption of specific or altered postures	Adoption of an AMC interaction position
Stage 5	Determine if these postures (from Stage 4) are pathomechanical	AMC interaction posture exerts fatiguing level of stress
Stage 6	Assess the effects of added weight to the body segment to which the wearable computer is attached	Increased arm abductor and flexor stress and elbow flexor stress
Stage 7	Determine what weight results in a significant increase in musculoskeletal loading and increased rate of fatigue	0.54 kg AMC results in a significant increase in rate of fatigue from a 0.2 kg AMC

(Dul, 1988). The current philosophy within the wearable computer community however, may not allow for this. The intertwining tenet of wearable computers dictates that the technology maintains operational and interactional consistency (Mann, 1997), meaning that they are always worn. This philosophy, of person and machine in perfect harmony, will have to be changed. Guidelines may have to be drawn up, limiting computer exposure, solely to allow the wearer's physical body to recover.

The stages, presented in this paper for a risk assessment of wearable computers, attempt to develop a basis for generating guidelines for wearable computer usage. For AMCs, a summary of risk assessment might look like Table 4. In Table 4 the findings are general to any type of AMC. However, to fully establish what effect wearing a computer will have on the wearer, an understanding of how and where the computer will be used, is also needed.

Knowledge of body posture has been shown to be important in determining the physical effect to the wearer. There are a number of methods for assessing working postures, from detailed 3D motion capture and video analysis, to simple paper based posture targeting (Li and Buckle, 1999). Once working postures are known, it should be determined whether, when wearing and interacting with the computer, the wearer is required to adopt a modified posture. The biomechanical structures that are stressed can be determined from simple biomechanical modelling. Laboratory studies, followed up with field-based studies, can measure the responses of these structures to working with the wearable computer in actual work situations. In addition to the wearable computer interaction postures, it must be remembered that the devices may also be worn in situations where other tasks are being performed. So, to produce a risk assessment that encapsulates the totality of risk to which the wearer is exposed, a complete assessment of all working postures will have to be undertaken, including postures adopted when the computer is not being used but is still being worn.

Recommending safe weights for wearable computers may prove difficult. This paper attempted to classify a weight as being safe, if it did not result in a significant increase in musculoskeletal loading or result in an increased rate of fatigue. These criteria, based on responses in muscle activity, generally passed greater weights than those that passed as determined by perceived pain and discomfort. This suggests that the criterion should not be based solely on muscle activity responses. Indeed, as perceived pain and discomfort results tended to be more sensitive, these are recommended over muscle activity.

Ultimately, the only method to determine if wearing and interacting with a wearable computer results in long term problems, such as the development of musculoskeletal disorders, is by prospective epidemiology (Sorock and Courtney, 1996). In prospective epidemiology studies, participants enter the study disease free and have no prior history of disease. They are then categorised by exposure status; in this case this will be a posture analysis, a record of where the wearable computer is attached, the weight of the computer, and the activities carried out by the wearer. They are then followed up with periodic examinations and interviews to assess the development of any symptoms of disorder. Over time a picture will then emerge as to how the wearer develops wearing the computer. With comparison to persons not wearing a computer, causal relationships between risk factors and new cases of disorder can be determined. It is proposed that, as wearable computers become incorporated into the work environment, prospective epidemiology studies be carried out.

5. Conclusion

Gemperle et al. (1998) stated that the long-term use of wearable computers has an unknown physiological effect on the body and that it will be important to test their effect. This paper has shown that the postures adopted, to interact with wearable computers, can place a fatiguing level of stress on the body and that the weight of the device can have a significant effect. In the short term, these stresses can result in sensations of pain and discomfort. In the long term, it is proposed that prolonged exposure to such stresses may have serious pathomechanical effects, including the possibility of developing musculoskeletal disorders. Therefore, as wearable computers become incorporated into the work environment, their physiological effect should be monitored.

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