Validation of Endurance Model for Manual Tasks*

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Abstract— **Physical fatigue in the workplace can lead to work-related musculoskeletal disorders (WMSDs), especially in occupations that require repetitive, mid-air movements, such as manufacturing and assembly tasks in industry settings. The current paper endeavors to validate an existing torque-based fatigue prediction model for lifting tasks. The model uses anthropometrics and the maximum torque of the individual to predict the time to fatigue. Twelve participants took part in the study which measured body composition parameters and the maximum force produced by the shoulder joint in flexion, followed by three lifting tasks for the shoulder in flexion, including isometric and dynamic tasks with one and two hands. Inertial measurements units (IMUs) were worn by participants to determine the torque at each instant to calculate the endurance time and CE, while a self-subjective questionnaire was utilized to assess physical exertion, the Borg Rate of Perceived Exertion (RPE) scale. The model was effective for static and two-handed tasks and produced errors in the range of [28.62 49.21] for the last task completed, indicating the previous workloads affect the endurance time, even though the individual perceives they are fully rested. The model was not effective for the one-handed dynamic task and differences were observed between males and females, which will be the focus of future work.**

An individualized, torque-based fatigue prediction model, such as the model presented, can be used to design worker-specific target levels and workloads, take inter and intra individual differences into account, and put fatigue mitigating interventions into place before fatigue occurs; resulting in potentially preventing WMSDs, aiding in worker wellbeing and benefitting the quality and efficiency of the work output.

Clinical Relevance— **This research provides the basis for an individualized, torque-based approach to the prediction of fatigue at the shoulder joint which can be used to assign worker tasks and rest breaks, design worker specific targets and reduce the prevalence of work-related musculoskeletal disorders in occupational settings.**

I. Introduction

Work-related musculoskeletal disorders (WMSDs) are the main reason people are absent from work globally [1]; they affect an organization's overall health care costs [1]; they result in massive economic burdens [1]; and are the second highest contributor to disability globally [2]. The most common WMSD conditions in industry include carpal tunnel syndrome, tendonitis, herniated disk disease, trigger thumb [3] and gorilla hand syndrome [4]. Since there is a link between fatigue and decreased performance, reliability, efficiency, increased accidents [5], and quality deficits [6], it is within the interests of the worker, industry organization and governing bodies to assuage the issue.

Ergonomic assessment tools are often used to assess the risk associated with repetitive work and are generally based on joints' angles, and the distances and heights a load is lifted. The Strain Index [7], REBA (Rapid Entire Body Assessment) [8], RULA (Rapid Upper Limb Assessment) [9], and NIOSH Lifting Equation [10] are a few examples, although they are not without disadvantages as some authors have questioned their thoroughness and risk classification (for example, [11, 12]). Other risk assessment and fatigue prediction equations present endurance-based approaches, oftentimes combined with electromyography (EMG) or inertial measurement unit (IMU) sensing. Developments in compact wearable sensors have led to a new dimension of tackling fatigue as they enable real-time monitoring of the worker. Human centric sensor data, combined with individual parameters, are also being utilized to create a tailored and personalized approach to mitigate fatigue and WMSDs. For example, the Threshold Limit Values [13] assesses the maximum voluntary contraction (MVC) and duty cycle. Maximum Endurance Time models, which consider static muscular contractions until exhaustion, have been developed for differing applications [14, 15], as well as models based on endurance and resumption times in repetitive tasks [16]. K-Score [17] combines joint angles and muscle fatigue by means of IMU and EMG sensors, and assesses fatigue-related ergonomic risk. Ergonomic risk and biomathematical models provide differing assessment results depending on the type of work [12], which, combined with differences between age groups [18], static and dynamic tasks [19], and load [20] demonstrates that they are highly task- and worker-specific, making it difficult for one method to be used for a single occupation and a diverse cohort of individuals.

The Hincapie-Ramos endurance model [4] predicts the time to fatigue in mid-air repetitive interactions based on the maximum torque that can be produced by an individual around the shoulder joint and the torque at a given moment. The model has been validated for simple unloaded hand interactions, and

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it may allow for the complexity and workload of manual tasks to be accounted for by considering joint angles and weights of the objects being lifted. It provides an individualized, non-gendered, and task-specific indication of the time to fatigue. Consequently, by validating the Hincapie-Ramos model for industry tasks and, if necessary, by adapting the model for industry applications, a time to fatigue can be predicted for industry workers.

Fatigue studies require validation measures to ascertain whether the individual is in fact fatigued. While EMGs have been used for such purposes [17, 21, 22], subjective self-assessment questionnaires are usually employed to serve as a reference measure. Questionnaires are generally focused on either physical (e.g., Borg RPE: Rate of Perceived Exertion [23]; Borg CR-10 [24]) or mental fatigue (e.g., KSS: Karolinska Sleepiness Scale [25]; and Stanford Sleepiness Scale [26]) and have been validated against physiological signals such as heart rate. Although questions have been raised regarding the validity of subjective questionnaires in ergonomics [27, 28], these examples are considered the 'gold-standard' as reference measures in current fatigue studies. The most commonly used questionnaires in manufacturing and industry-based tasks with wearables sensors are the Borg RPE and KSS scales [29]. While the Borg RPE denotes the physical state of the individual, it can additionally give an insight into mental and emotional states, and the motivation to complete a task, considering its subjective nature. Many authors such as Balkin (2011) [30] recommend the use of a hybrid approach to the fatigue problem, incorporating a variety of measurements and modelling methods.

A hybrid approach is employed in the present study, which aims to validate the Hincapie-Ramos model in the realm of individualized and task-specific manual tasks with the use of hand-held items. Twelve subjects complete three lifting tasks with weights, both static and dynamic, while wearable sensors are used to collect physiological data, and subjective questionnaires act as a reference measure. It is hypothesized that the model predicts the time to fatigue for each subject, for both static and dynamic tasks. It is important that the model is validated for simple lifting movements under static and dynamic conditions before it is applied to complex and intricate tasks, and tasks that are applicable to a specific occupation. The broad approach taken in the present study means that it is applicable to any sector that requires the use of endurance data, such as sports, rehabilitation, ergonomics, and occupational health and safety.

II. Methodology

Twelve healthy participants (7 female, Table 1) took part in the study and were recruited by email advertising and word of mouth. Participants were excluded if they reported any musculoskeletal disorders; all subjects provided written informed consent. The study had ethical approval from the university's ethical committee (CREC Review Reference number: ECM 4 (p) 6/7/2021). The mean age of the cohort was 30 years in the range of [23 62]. Participant's weight ranged between [56.3 102.7] with a mean of 72.23 kg. Anthropometric measurements were recorded, and the force of MVCs for flexion of the shoulder was measured by means of a NK-500 force

TABLE I. Subject Details

Parameters	***All Subjects***	***Group A***	***Group B***
Sex	7 females, 5 males	4 females, 2 males	3 females, 3 males
Age (years)	30 ± 10	33 ± 14	26 ± 3
Weight (kg)	72.23 ± 13.18	71.52 ± 12.30	72.95 ± 13.96
Height (cm)	1.72 ± 0.06	1.70 ± 0.07	1.74 ± 0.05
Upperarm Length (m)	0.33 ± 0.02	0.34 ± 0.02	0.33 ± 0.02
Forearm Length (m)	0.26 ± 0.02	0.27 ± 0.02	0.26 ± 0.02
Hand Length (m)	0.20 ± 0.01	0.19 ± 0.01	0.20 ± 0.02
Upperarm Radius (m)	0.05 ± 0.01	0.05 ± 0.01	0.05 ± 0.01
Forearm Radius (m)	0.04 ± 0.00	0.04 ± 0.00	0.04 ± 0.00
Hand Radius (m)	0.03 ± 0.00	0.03 ± 0.00	0.04 ± 0.00
Maximum Shoulder Joint Torque (Nm)	49.61 ± 19.86	45.97 ± 20.52	53.25 ± 18.47

gauge. During the measurement, the arm was outstretched straight ahead in supination with the shoulder, wrist and elbow joints in alignment. Participants were seated to minimize influence from the lower body, with their back upright, thighs horizontal to the floor and feet on a footrest attached to the seat. The seat height was adjustable to enable the participant's shoulder to be at the same height as the force gauge handle which was at the center of their palm as they pushed upwards, moving their shoulder in flexion. MVCs and anthropometric data were used to calculate maximum shoulder joint torque in flexion. Subjects were divided into two groups, A and B, depending on the order they completed the tasks in the study. Subjects' details within each group are shown in Table 1.

Participants wore the XSENS Awinda shirt, glove and straps to adhere the IMUs (XSENS MTw Awinda), sampling at 100Hz. IMUs were placed on the dominant (self-reported) side of the body, to the sternum, shoulder, upper arm, forearm and the palm, according to the manufacturer's recommendations (Figure 1).

Three tasks were selected. A static task with the shoulder of the dominant arm held at 90 degrees in flexion, that is, in the same position as the MVC exercise, while holding a weight (*Static Task*); a dynamic task with the dominant hand moving the shoulder joint from 0 to 90 degrees in flexion while holding a weight (*Dynamic Task*); and finally, the *Dynamic Task* with both hands holding the weight (*Two-handed Task*). The MVC force was used to calculate the weight a subject lifted during the tasks, with participants approximately 30%-45% of their MVC force. The hand was in pronation for the tasks in the

Figure 1. IMU placement on the shoulder, upper arm, forearm, hand and torso.

study: this was done to isolate the shoulder muscles and reduce the influence of the biceps for the shoulder tasks. During the study, the exertion of subjects was tracked throughout the task by means of the subjective Borg RPE scale. The scale ranges from 6 to 20 and categorizes physical exertion into 4 categories: no exertion (6-7), light (8-11), hard (12-16), maximal (17-20). The Borg RPE score was recorded before the task, and ten and twenty seconds after starting the tasks, and every twenty seconds thereafter. The tasks were completed until the subject was fatigued, verbally reporting they were maximally exerted on the Borg RPE scale, or when they were unable to maintain the task position within approximately 5 degrees for *Static Task*, or when they were unable to maintain the prescribed rate of movement (i.e., 60 beats per minute played on a metronome) for the dynamic tasks. The final Borg rating was also noted at the end of each recording. Individuals took ten minutes of rest before they began the subsequent task; according to upper extremity recovery models [31-33] no more than six minutes is required for recovery after a fatiguing interaction, and ten minutes was sufficient to reduce subject's Borg rating to 6 or 7, demonstrating that they were rested. Fatigue is highly task-dependent; therefore, subjects were divided into two groups with different task orders: Group A's task order was *Static Task*, *Dynamic Task*, *Two-handed Task*; while Group B's task order was the reverse.

Endurance as a function of torque as described in [4] is:

$$E(T_{shoulder}) = \frac{1236.5}{\left(\frac{T_{shoulder}}{T_{max}}\times 100 - 15\right)^{0.618}} - 72.5 \qquad (1)$$

Where T_{max} is the maximum shoulder joint torque the individual can produce and $T_{shoulder}$ is the sum of the torques acting on the frontal axis of the shoulder joint, represented by Equation 2 in static conditions and Equation 3 in dynamic conditions. The torque was calculated for each frame based on the joint angles and the centre of mass (CoM) position.

$$\|\vec{T}_{shoulder}\| = \|\vec{r} \times m\vec{g}\| \qquad (2)$$

$$\|\vec{T}_{shoulder,t}\| = \|\vec{r} \times \overrightarrow{acc}_t \times m - (\vec{r} \times m\vec{g} + I_t \vec{\alpha}_t)\| \qquad (3)$$

The m is the mass of the arm and weight, $\overrightarrow{acc}$ is the acceleration of the CoM, found by double differentiation of the distance of the CoM from the shoulder, $\vec{r}$, found with Equations 4 and 5. Angular acceleration α was found using $\frac{\overrightarrow{acc}}{\|\vec{r}\|}$. The mass of the weight was halved for the *Two-handed* task.

$$D = B + \frac{WrHaWe_{mass}}{EbWr_{mass} + WrHaWe_{mass}} \overrightarrow{BC} \qquad (4)$$

$$CoM = A + \frac{EbWr_{mass} + WrHaWe_{mass}}{ArmWe_{mass}} \overrightarrow{AD} \qquad (5)$$

Where A, B and C are the CoMs of the upper arm, forearm and hand respectively, found as described in [34] and joint angles extracted from IMUs (Figure 2). D is the CoM of the forearm (EbWr), hand and weight (WrHaWe). The mass of the arm segments from [35] were used. The mean of the Hincapie-Ramos endurance time was extracted as the predicted endurance time. The I denotes the inertia, which is found for each frame using the Parallel-Axis Theorem, as described in [36]. The inertia of the segments are summed with the inertia of the dumbbell, which is approximated as a sphere in this case.

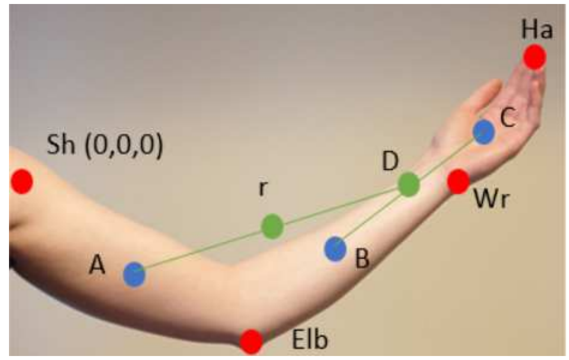

Figure 2. Locations of center of mass of the arm.

TABLE II. RESULTS

Endurance Measurements and Predictions				
		Static Task	***Dynamic Task***	***Two-handed Task***
Measured Endurance Time (s) (mean ± std)	All Subjects	63.25 ± 21.13	87.75 ± 34.67	138.42 ± 49.28
	Females	59.14 ± 19.79	93.86 ± 40.40	125.43 ± 39.71
	Males	69.00 ± 21.60	79.20 ± 21.79	156.60 ± 55.27
Predicted Endurance Time (s) (mean ± std)	All Subjects	56.01 ± 6.78	165.12 ± 4.47	172.30 ± 3.02
	Females	54.40 ± 3.25	165.61 ± 5.61	171.56 ± 1.86
	Males	58.27 ± 9.32	164.43 ± 1.74	173.35 ± 3.89
Endurance Time Absolute Error %	All Subjects	28.62 ± 23.27	115.45 ± 70.42	49.42 ± 38.72
	Females	35.74 ± 25.85	111.26 ± 81.23	54.98 ± 45.55
	Males	18.66 ± 13.91	121.32 ± 51.03	41.13 ± 24.13

$$\sum I = m\rho_0^{\ 2} + mx^2 \qquad (6)$$

Where m is the mass of the segment, ρ_0 is the radius of gyration around the center of gravity and x is the distance from the shoulder joint. The mean of the Hincapie-Ramos endurance time was extracted as the predicted endurance time.

III. RESULTS

Table 2 shows the experimental and predicted endurance times and the endurance time error. Measured endurance times for *Dynamic Task* are slightly longer than that of *Static Task*, likely due to the reduced torque as the shoulder angle decreases throughout the repetitive movement, however, this is not reflected in the mean predicted endurance times. This task shows the highest errors, regardless of task order and gender, and the predicted endurance time was consistently underestimated. Gender affected the endurance time measured, with the average measured endurance time greater for males in *Static* and *Two-handed* task and only slightly less than females for the *Dynamic* task, however, greater variability is observed for females in this task. The predicted endurance times for males and females are all relatively close, indicating the model did not take the greater variability and generally lower maximum forces (Table 1) of females into account. Fig.

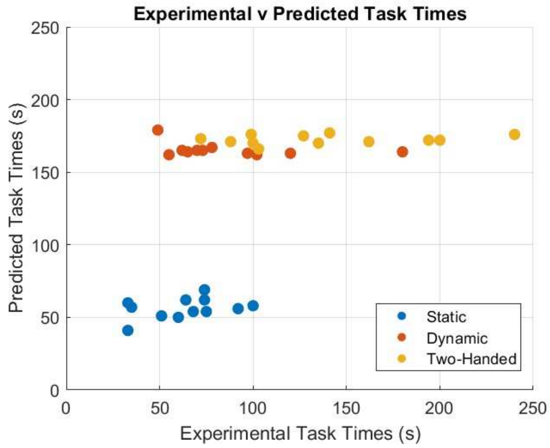

Figure 3. Experimental and predicted endurance times.

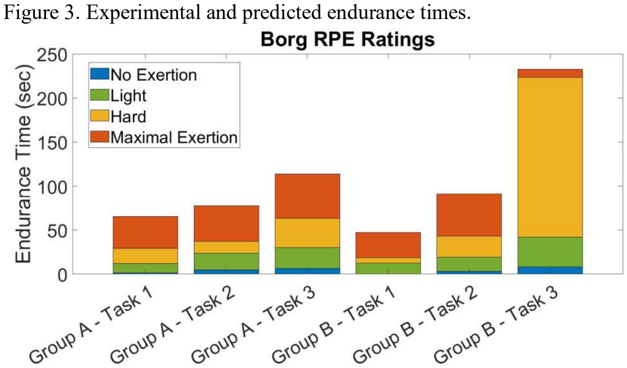

Figure 4. Mean length of time (seconds) participants spent in each Borg RPE category.

3 and shows the *Dynamic* predicted endurance time is consistently higher than the measured endurance time, while there is no such pattern for *Static* and *Two-handed* tasks.

It is also possible that group, therefore task order, affected the measured endurance times. Group A completed *Two-handed Task* last and Group B completed *Static* first, resulting in lower measured endurance times for these group-task combinations. Group A's mean experimental time for the *Static* task was 77s as opposed to Group B's 50s, while Group A and B's mean experimental time for the *Two-handed* task were 126s and 151s, respectively. This is also demonstrated in Fig. 4 which shows the mean time spent in each Borg RPE exertion zone. Group A participants tended to be more exhausted by the final task than Group B, spending most of their time in the maximally exerted zone. Likewise, Group B spent most of the *Static Task* maximally exerted.

IV. Discussion

Task order and gender were found to affect the endurance times of subjects, indicating that gender, previous tasks, their workloads, and duration must be taken into account in fatigue prediction, even if the individual reports that they are fully rested from any previous activity. Errors for *Dynamic Task* were highest across all participants, regardless of sex, torque or task order. The number of subjects, the narrow age range of the cohort and the completion of all three tasks on the same day are limitations to the study. Individuals were tested between 30-45% of their MVC, however, a range of intensities (i.e. 10-20%, 20-30%, etc.) in different directions should be tested to gain a full understanding of how the model performs.

The results of the present study illustrate that the Hincapie-Ramos endurance model can provide a valuable tool for estimating the endurance time of static and dynamic lifting tasks about the shoulder joint under specific boundary conditions. The choice of Hicapie-Ramos et al. to use torque instead of force, like in Romhert's model [37], better reflects the change of CoM during the movement, but makes any error in the anthropometric data (length and radius of the body segments of the arm) to propagate faster. Future work will involve the reduction of errors and modification of the model to deal with the aforementioned shortcomings. For example, modifications may comprise of adjusting the model to be applicable to other joints (e.g. elbow and wrist), other directions (e.g. internal and external rotation) while using one or two hands at a range of standardised intensity levels and considering the torques in more than one plane. Other considerations, such as the influence of body composition parameters, exercise level and task type and order, can be addressed in future models. Additional wearable sensors, such as EMG, galvanic skin response and photoplethysmography, can be integrated into future studies, where non-subjective physiological data can be used to compare the physical state of the subject to their consumed endurance. Real-time endurance time predictions in combination with real-time physiological sensing provides a strong basis for fatigue prediction, detection and mitigation of fatigue. The work in the study provides a basis for the model to be applied to industry specific tasks, enabling workers to benefit from the advantages of the Hincapie-Ramos endurance curve and CE model. It may also provide the basis for a more individualized approach to ergonomic risk assessment questionnaires or workload prescription, such as Snook's (1991) table of maximum acceptable weights and forces [38]. An individualized fatigue prediction model can be used to design worker-specific target levels and workloads, take inter and intra individual differences into account, and allow fatigue mitigating interventions to be put in place before fatigue occurs; resulting in potentially preventing WMSDs, aiding in worker wellbeing and benefitting the quality and efficiency of the work output.

V. Conclusion

Overall, the torque-based Hincapie-Ramos model has been found to predict the endurance times of a loaded and flexed shoulder joint, in one-handed static conditions and two-handed dynamic conditions with errors of 28.62% and 49.21% respectively. Gender was found to affect the predicted endurance time, especially for the *Static* task, the error 18.66% for males and 35.74% for females. The model significantly overestimated one-handed dynamic tasks across the cohort, regardless of task order.

More subjects and a wider range of abilities and ages are required for future work which may involve the integration of wearable sensors to gain unbiased physical exertion measurements and the modification of the Hincapie-Ramos endurance model to adjust different joints under load during static and dynamic conditions, with one and both hands and at a range of intensities and directions.

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