Robotica (2005) volume 23, pp. 319–327.
doi:10.1017/S0263574704001341 Printed in the United Kingdom

Using potential energy to measure work related activities for persons wearing upper limb prostheses

Nancy Black*, Edmund N. Biden† and Jeremy Rickards†

SUMMARY

This study presents a novel means of assessing upper limb tasks by using mechanical energy. Potential energy quantifies six work related activities, studied for 20 working-age non-prosthesis users and three powered below elbow prosthesis users. Two marker trajectories on each of the upper arms, forearms, and hands were captured using a 3-camera VICON 140™ system. Task and arm dominance of non-prosthesis users are highly significant ($p < 0.01$) with arm dominance effects being more pronounced for prosthesis users. Qualitative inter-repetition consistency is also concordant with observed increased cumulative trauma disorders among prosthesis users.

KEYWORDS: Prostheses; Upper limb; Mechanical energy.

I. INTRODUCTION

This paper uses a measurement and analysis procedure quantifying mechanical energy of the human upper limbs during work related activities to compare the upper limb mechanical energy levels of individuals with and without an upper limb prosthesis in the context of qualitatively observable movement characteristics. This procedure is shown to be consistent with qualitative observations associated with increased risk of cumulative trauma disorders.

There is a large gap between the qualitative methods commonly used to analyse activities and motions in rehabilitation settings, and the quantitative capacities of motion analysis recording tools. Skilled therapists may compare reliably a benchmark with other populations,[1] however not all prosthetics clinics include occupational therapists. In addition, since amputees can be expected to adapt their work methods over time to match their abilities with those of their prosthesis,[2] long-term follow-up and analysis of actual work methods and movement efficacy are required in order to design working environments which reduce functional limitations of the prosthesis. Past studies of user priorities,[3,4] qualitative observations[5] or strictly time-based comparisons[6] do not directly compare movements and method efficacy of similar prosthesis users and normally limbed adults. Unbiased comparative testing is crucial to designing tools and environments to surmount functional handicaps. Such testing is possible using modern motion recording tools in conjunction with biomechanical modelling.

Despite addition of external power sources and continued research improving control systems for upper limb prostheses, use of a unilateral prosthesis is associated with long term health risks. A recent study found 53% of unilateral upper limb amputees using a prosthesis had pain in their remaining arm, most of which was associated with cumulative trauma disorders.[7] In comparison, upper limb musculo-skeletal disorders account for approximately one-third of the occupational injuries[8] which are estimated to affect annually 50% of the American workforce.[9] Thus unilateral upper limb amputees are roughly three times as likely to suffer repetitive motion or overexertion type injuries as the general workforce. The challenge in avoiding overuse injuries among amputees is that many tasks can be performed with one arm and hand[10] and that a prosthesis cannot be expected to take on more than 30% of the function of a bi-manual task.[11] Since the most common level of upper limb amputation is in the forearm segment ("below-elbow" accounting for 44.6%[3] or 43%[12]), the wrist and hand must be replaced and longitudinal rotation, wrist and finger movements are affected. While prosthesis design continues to evolve, current commercially available upper limb prostheses still limit the functionality of users compared to normally limbed individuals. Most prosthesis designs only allow longitudinal wrist rotation and a pinch grasp opposing "thumb" and "fingers", excluding co-ordinated multi-joint movements. Furthermore, while externally powered limbs may provide good cosmesis, they provide only limited sensory feedback and when users attempt to increase visual feedback to increase functionality,[3] movements or usable postures may be restricted.[13] Unfortunately, how these design limitations affect overall functionality and upper limb movements associated with cumulative trauma disorders is unknown.

II. METHOD

This study was approved by an ethics review board at the University of New Brunswick. The method used will be described in the context of a typical recording session, making special note of the characteristics of participants involved, the recording equipment, and calculations used to analyse the data.

Corresponding author: Dr N. Black.
* Faculté d'ingénierie, Université de Moncton, Moncton, New Brunswick, E1A 3E9 (Canada).
† Institute of Biomedical Engineering, University of New Brunswick, Fredericton, New Brunswick, E3B 5A3 (Canada).

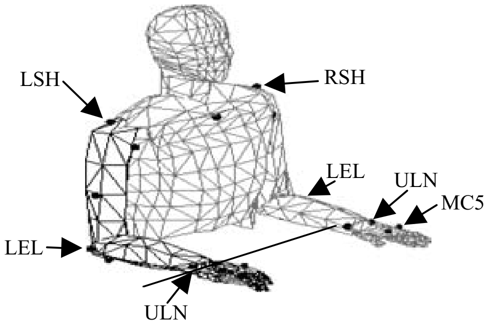

Fig. 1. View of skin-mounted marker locations.

II.1. Recording session method

Subjects read and signed an informed consent form prior to participating. All subjects were screened for current pain or discomfort that might affect their capacity to accomplish work related activities involving the upper limbs. Anthropometric measurements taken included body mass, fingertip reach, upper arm, forearm and hand length as well as circumferences of each of these segments (maximum, proximal and distal values for arm segments and hand width, depth and circumference at the metacarpophalangeal joints). The prosthesis mass and residual limb length were also recorded for prosthesis users.

Wearing a sleeveless top exposing the shoulders and arms, individual spherical retroreflective markers were stuck to the skin over anatomic landmarks at the distal and proximal ends of each upper limb segment, in order to estimate the locations of segmental centres of gravity (see Figure 1). Markers of 1.6 cm diameter were placed over the left and right acromion (LSH, RSH), the lateral epicondyle of the humerus (LEL), and the ulnar styloid (ULN) and a 1.2 cm diameter marker was affixed over the distal end of the fifth metacarpal just proximal of the phalange (MC5). It should be noted that a third marker was placed on each segment although segmental location estimation based on two markers per segment was found to be sufficient for analysis of segmental potential energy, confirming that joint centre of rotation location did not vary greatly during the recorded activities. Marker locations were chosen to minimise skin-movement artefact[14] and allow reliable placement. Marker placement – replacement variation and skin artefact were measured during pilot trials involving four predetermined static postures at two different occasions on each of six individuals. Average variations between marker placement and re-placement were at most 6.7 mm (for the LEL marker on the right side) with combined recording and skin movement error having at most 0.83 mm standard deviation (for the LEL marker).[15]

The experimental set up is depicted in Figure 2. In a relaxed, erect, seated posture, subjects were familiarised with the work related activities to be studied using fixed goal-oriented instructions. Work related activities for this study were selected based on those previously used in functional hand and prosthesis testing.[1,16] Tests involved both hands using a variety of upper extremity joint motion, dexterity, force and velocity requirements and were performed in a seated posture using standard tools (confirmed during videotaped pilot trials). The use of objective measurements of limited duration, standardised tasks involving broad aspects of hand function commonly used in activities of daily living (ADLs) and readily available test materials is consistent with previous hand function tests.[17] The six work related activities chosen are described in Table I. It was not anticipated that these activities would vary greatly between right and left-handed individuals.

A cushioned ergonomic chair was used with seat height adjusted so that the working height was 3 cm below seated elbow height. A footstool was provided, where needed, to ensure full foot support. Tools for task accomplishment were

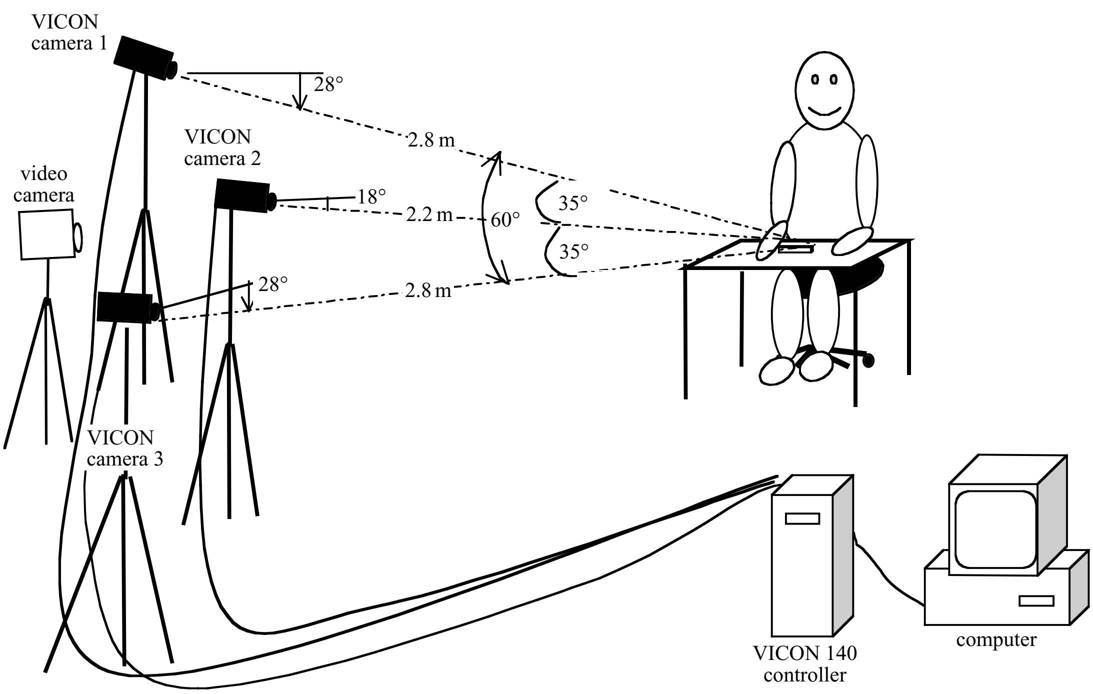

Fig. 2. Schematic of experimental set-up.

https://doi.org/10.1017/S0263574704001341 Published online by Cambridge University Press

Table I. Motion characteristics of the work related activities tested.

Work related activities	Expected motion characteristics
1) zipping and unzipping the main U-shaped zipper on a backpack	– slow movement at relatively constant height – one arm holds while other pulls; roles may reverse; pulling hand may pass above or below holding hand
2) changing a screwdriver bit	– small amplitude, slow movement most frequently involving the fingers, wrist and forearm – similar height to hold screwdriver and manipulate bits
3) cutting two slices of a plasticine "steak", picking them up and placing them on an adjacent plate	– one hand holds fork above steak; cutting arm maintained lower, movement involving entire arm – both arms used to place slices
4) hammering a nail 1.25 cm into a block of wood	– one arm used to hold the nail (or block of wood) while second arm hammers – hammering arm consistently higher, and much more dynamic than holding arm
5) folding a sheet of paper and placing it in an envelope	– similar heights and velocities used for both arms – mostly gross arm movements – insertion orientation and height varied
6) tying a knot and a bow with a string	– both arms maintained at similar heights – mostly finger and horizontal movements

centred in front of subjects at fixed percentages of individual reach. Fifty percent reach was defined as the distance from the table edge to the metacarpophalangeal joint with the forearms resting on the table surface as shown in Figure 1. One hundred percent reach was defined as fingertip reach with the arm extended forward and down with the elbows straight. Task ordering was fixed (as given in Table I) reflecting increasing difficulty for prosthesis users to minimise demotivation. All tasks began and ended with the subject sitting erect with their upper arms relaxed and their hands prone on the work surface at 50% reach, the wrists aligned with the table edge (depicted in Figure 1). Each activity was practiced at least twice prior to recording to verify the constancy of task method. Once confirmed, four repetitions of each task were recorded, two recording the right side, and two the left side, with the ordering of these randomised. The subject and working surface were rotated 90° between the two sides, with recording systems held stationary. A one-second static recording of the subject in the starting position was used to define marker locations relative to known joint angles.

II.2. Participants

Twenty adults with no upper limb disability (ten men and ten women) were studied to provide a reference population of "non-prosthesis users". Three adults (one man and two women) with a below-elbow amputation, using a myoelectrically-controlled externally powered prosthesis were studied ("prosthesis users") (Table II).

While six non-prosthesis users were left handed, handedness did not significantly affect the movements involved in the tasks studied. While it would be expected that the dominant arm after amputation would be the one without the prosthesis, this may not be the arm that was dominant prior to amputation. Among the prosthesis users studied, two considered themselves right handed and used a prosthesis on their left arm, while the third used the prosthesis on his right arm but did not consider himself to have a dominant

Table II. Comparison of subjects by age in years.

	Non-prosthesis users		Prosthesis users	
Age in years	men (n = 10)	women (n = 10)	man (n = 1)	women (n = 2)
Average	38.4	33	54	19
Standard deviation	14.7	14.4	–	2.8
Maximum	62	54	54	21
Minimum	18	18	54	17

arm, either before or after amputation (see Table III). Notably all three amputees had relatively long residual forearms (see Table III).

II.3. Recording equipment

Quantitative movements of the skin-mounted markers were recorded at 60 Hz using a 3-camera VICON 140™ motion analysis system. Prior to recording subjects, the system was calibrated with a three-dimensional cubic calibration object supplied by VICON, centred in the working volume. For the camera positions described, markers moving in the workspace are recorded by the system with an error of approximately 1 mm in each dimension. For angular measures with limb segment lengths typical of adults, this translates into angular uncertainties of approximately 1°.[18] Since three dimensional locating requires that a marker be visible by at least two cameras, camera placement affects marker visibility and inter-camera angle affects locating error. Pilot studies determined that the motions of only one arm at a time could be reliably recorded with the three cameras. Both vertical and horizontal VICON camera placement were adjusted to obtain maximum marker visibility while maintaining reasonable camera separation (see Figure 2). The first and third cameras were oriented 28° below the horizontal 2.8 m from the work origin, while the

https://doi.org/10.1017/S0263574704001341 Published online by Cambridge University Press

Table III. Characteristics of prosthesis users.

Subject code	fp1a	fp2a	mp01
Dominant arm	Right	Right	neither
Prosthetic arm	Left	Left	Right
Dominant arm prior to amputation?	Right	neither	neither
Years using prosthesis	2	15	25
Type of amputation	traumatic	congenital	traumatic
Current prosthesis functions	myoelectric multifunction hand; friction wrist	myoelectric open/close hand grip; friction wrist	myoelectric continuous control hand; friction wrist
Residual arm length (cm)	24.5	21	27.5
Non-prosthesis forearm length (cm)	28	28	28

second camera was equidistant between the other cameras and angled 18° below the horizontal 2.2 m from the work origin. This volume was sufficient for recording the upper limb movements of the expected populations. With this placement of cameras symmetrical about the middle camera, marker visibility was good and bias was avoided when rotating the subject 90° from left to right side recordings. Markers on the test subjects were not removed between trials. When switching from left to right side trials the person being tested was repositioned within the calibrated workspace. Even so, some ADL recordings could not be analysed due to gaps over 10 samples long (0.166 second) in marker trajectories.

A video camera recorded continuously during the work related activities permitting qualitative comparisons across tasks and repetitions.

II.4. Calculations

VICON 140^{TM} work related activity recordings were truncated to include only the period of the actual task. Marker trajectories were filtered using a zero phase shift 4th order Butterworth 6 Hz low pass filter. Prior to filtering, signal power over 5.98 Hz was calculated to be less than 1% for all markers over all recorded tasks. Recorded markers were used to estimate upper limb segment locations and to calculate segmental mechanical energy, ME. ME is defined as:

$$\mathrm{ME} = \mathrm{KE} + \mathrm{PE} \tag{1}$$

$$\mathrm{KE} = \frac{1}{2} * m * v^2 + \frac{1}{2} * I * \omega^2 \tag{2}$$

$$\mathrm{PE} = m^* g^* h \tag{3}$$

where

m = mass (kg)
I = mass moment of inertia about the centre of mass (kg * m^2)
v = translational velocity of the centre of mass (m/s)
ω = angular velocity (rad/s)
h = height of centre of gravity (m)
g = acceleration due to gravity = 9.81 m/s^2

Segmental potential energy, PE, simply requires segmental mass and centre of gravity height. These values were defined from total body mass and segmental lengths based on published sources, as shown in Table IV.[19] Kinetic energy, KE, requires three linear and three angular velocity components, making it more sensitive to displacement recording errors and dependent on visibility of each of three non-collinear markers on each segment. Since preliminary calculations during all six work related activities studied found that KE accounted for less than 2% of total segmental mechanical energy of the upper limb on average[15] and a maximum of 5.5% of changes in PE, PE alone was used to estimate segmental mechanical energy. Potential energy is sensitive to postural changes since the relative height of the limb segments above a neutral surface is the measure of height used to calculate energy. Postures in prosthesis users where humeral elevation is used to position the prosthesis involve increased potential energy over extended periods during the task.

Table IV. Definitions of anthropometry as percentage of total adult human body mass by gender[19].

Segment	Segmental mass by gender: Male	Segmental mass by gender: Female	Location of centre of mass from distal end %
upper arm (both)	6.6%	6.0%	56.4%
forearm (both)	3.8%	3.1%	57.0%
hand (both)	1.3%	1.0%	72.0%

For each recording, average PE was calculated over the entire duration of each activity and normalised on activity length (n):

$$average\_upper\_limb\_\mathrm{PE} = \sum_{segments} \frac{\sum_{i=1}^{n} \mathrm{PE}_{i(segment)}}{n} [J] \tag{4}$$

where *segments* = [upper arm, forearm, hand]
To compare potential energy values across the both upper limbs and between subjects, a composite measure of height was defined, h^*, by normalising total average upper limb PE

https://doi.org/10.1017/S0263574704001341 Published online by Cambridge University Press

on total body weight (mass times acceleration due to gravity):

$$h^* = \frac{\sum_{segments} average\_upper\_limb\_PE}{body\_mass^* g} * 1000\,[\text{mm}] \quad (5)$$

Since h^* ignores the difference between the upper limb masses of women and men, and is based on the weighted composite height of the three upper limb segments, h^* is not an exact height. A second comparative value was defined as *relative* h^* which removed the offset in PE by segment:

$$relative\ h^* = \frac{\sum_{segments} (average\_segment\_\text{PE} - \min(segment\_\text{PE}))}{body\_mass^* g} * 1000\,[\text{mm}] \quad (6)$$

The above calculations are based on mean or relative PE, which removes the effect of time taken to perform the task. By contrast, had the integral of the energy been computed, this would have included a measure sensitive to time. Task duration was not the focus since there was a wide variation in time taken and the ability to perform the task is more critical.[1] Analyses were performed by task across individuals.

III. RESULTS AND DISCUSSION

Results of both quantitative analysis of segmental energy measures and qualitative analysis of movement, motion and material characteristics are presented by task.

III.1. Quantitative results

Comparative boxplots and univariate analyses of variance using SPSS (version 10.05) explore the importance of subject, task, gender, and arm dominance in determining h^*. Note that for the boxplots presented here, the shaded box represents the interquartile range containing 50% of the values, the whiskers (the lines extending from the box) go to the highest and lowest values excluding outliers, and the line across the centre of the box represents the median value. Outliers are defined as data points between 1.5 and 3 box lengths from the upper and lower edge of the box, while extreme values are those more than three box lengths from the upper and lower edges of the box. N refers to the number of data points considered in a given category. All analysable repetitions of each activity for each subject for each side (left and right) were included. Significance was defined as $p < 0.05$. Because of the limited number of prosthesis users tested, data from this group are compared either as a group or individually, relative to the non-prosthesis user population.

Gender: Among non-prosthesis users, gender alone was significant in determining h^*. This was expected given the different percentages of segmental body masses for men and women, and the tendency for men to have larger and longer upper limbs. Indeed, once normalised on both these factors, gender was no longer significant.

Arm dominance: While handedness did not significantly affect h^*, viewing the dominant side (for example, the left view of a left-handed person) was significantly different from viewing the non-dominant side. Comparing dominant and non-dominant h^* values for each of the two populations studied, prosthesis users had a greater difference between the limbs than non-prosthesis users with the dominant side h^* being higher and the non-dominant side being lower relative to non-prosthesis users (see Figure 3). Notably, the non-prosthetic side was assumed to be dominant for prosthesis users.

When comparing h^* values over all analysable recordings by subject, values for prosthesis users were similar to those of the non-prosthesis users. Exploring these population tendencies further, Figure 4 shows that h^* values of prosthesis users relative to non-prosthesis users varied as a function of task and arm dominance. Where files were analysable for both dominant and non-dominant arms for a given task,

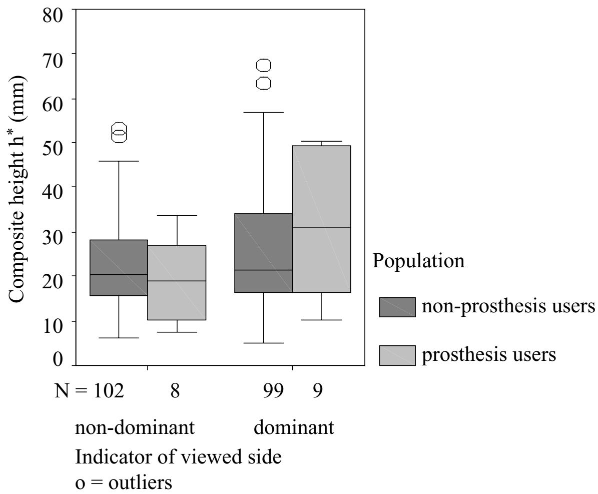

Fig. 3. Boxplot comparing h^* of dominant and non-dominant arms across populations.

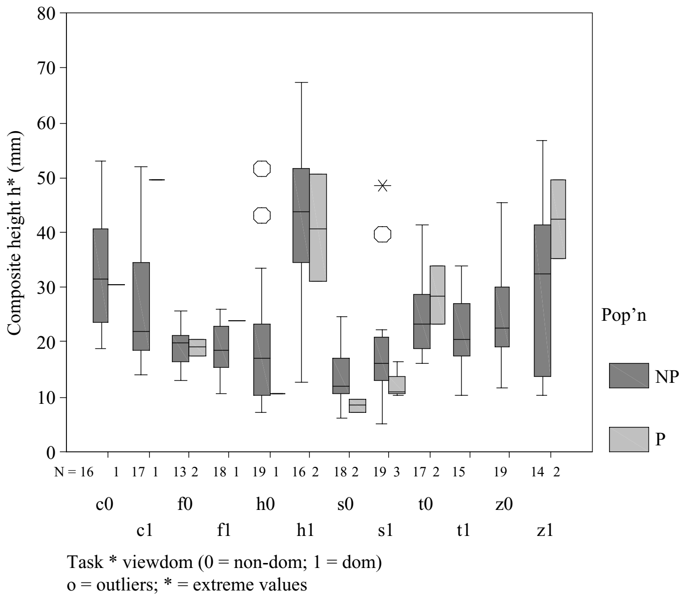

Fig. 4. Boxplot of h^* by task (c = cutting, f = folding, h = hammering, s = screwdriver bit changing, t = tying, z = zipping) and arm dominance (0 = non-dominant; 1 = dominant), grouped by population (NP = non-prosthesis users, P = prosthesis users).

https://doi.org/10.1017/S0263574704001341 Published online by Cambridge University Press

Table V. Tukey pair-wise comparisons of mean difference in *relative* h^* by task for 0.05 significance (for a subset of data with no missing combinations).

Task	Folding	Tying	Zipping	Cutting	Hammering
Screwdriver bit changing[a]	0.32	0.77**	1.11**	1.36**	1.48**
Folding	–	0.45	0.79**	1.04**	1.16**
Tying	–	–	0.34	0.59	0.71*
Zipping	–	–	–	0.25	0.37
Cutting	–	–	–	–	0.12

[a]The minimum mean relative h^* value was 1.63 mm for screwdriver bit changing. Differences are presented in increasing order, left to right.
*Significant ($p < 0.05$); **Highly significant ($p < 0.01$).

prosthesis users consistently had higher h^* values associated with the dominant arm than the non-dominant. Prosthesis users tended to have lower h^* values than non-prosthesis users during the screwdriver bit changing and hammering; the opposite was the case for the dominant arm during folding and zipping, and the non-dominant arm during tying.

In both Figures 3 and 4, some observations for the non-prosthesis users were classified as outliers or extreme values and are shown as such in the figures. These values were included in the analysis.

A univariate ANOVA on h^* including subject, dominant side * task, task and dominant side * subject found all variables highly significant. Interestingly, a univariate ANOVA on *relative* h^* for all available data of non-prosthesis users found subject and dominant arm * subject were no longer significant ($p = 0.09$ and $p = 0.12$, respectively). Further exploration used Tukey Honestly Significant difference pair-wise comparisons of mean differences in *relative* h^* values for non-prosthesis users (Tables V and VI). By task (Table V), screwdriver bit changing had lowest *relative* h^* but was not significantly different from folding. Hammering had greatest *relative* h^* values and was significantly greater than bit changing, folding and tying. When comparing *relative* h^* on task and dominant arm view (Table VI), the dominant arm hammering is significantly greater than all other activities, with the exception of the non-dominant arm during cutting. In contrast, *relative* h^* of the non-dominant arm during hammering is among the lowest values, and is only greater than *relative* h^* of bit changing and folding (both views). The *relative* h^* of the non-dominant arm during the screwdriver bit changing task is lowest of all.

III.2. *Qualitative video analysis*

Video recordings of all sessions were analysed for inconsistencies between repetitions and an Inconsistency Index (ICI) was defined to complement the single energy-based h^* values defined. Data for ICI for both populations studied are presented here by task in Tables VII to XII. For each task characteristic, mean, standard deviation and 95% confidence intervals of ICI are presented for non-prosthesis users, and ICI raw values are reported by subject for prosthesis users.

Zipping task: Inconsistencies among prosthesis users occurred only in the number of pulls used to close the zipper (higher than non-prosthesis users) and zipper stickiness (consistent with non-prosthesis users). Inconsistencies among non-prosthesis users were mostly movement-related (Table VII) and inconsistencies occurred only rarely in the role of each hand (pull and hold, or pull or hold), hand placement, or hesitations. Of the inconsistencies noted among non-prosthesis users, the number of pulls required to open or close the zipper varied most frequently.

Screwdriver bit change task: Prosthesis users showed only movement inconsistencies during the repetitions of this task (twisting the wrong way or hesitating), whereas for non-prosthesis users, eight of the 10 observed bit

Table VI. Tukey pair-wise comparisons of mean difference in *relative* h^* by task and dominant arm view for 0.05 significance.

Task[a] × View domside[b]	N	s1	f0	f1	h0	t1	z0	t0	c1	z1	c0	h1
s0[1]	18	0.45	0.55	0.56	0.63	0.83	1.16*	1.17*	1.31**	1.59**	1.91**	3.00**
s1	19		0.10	0.10	0.18	0.37	0.70	0.71	0.84	1.14*	1.45**	2.54**
f0	13			0.001	0.08	0.27	0.61	0.61	0.75	1.04	1.35**	2.45**
f1	18				0.08	0.27	0.61	0.61	0.75	1.04	1.36**	2.44**
h0	19					0.19	0.53	0.53	0.67	0.96	1.28**	2.36**
t1	15						0.33	0.34	0.48	0.77	1.08	2.17**
z0	19							0.01	0.14	0.43	0.75	1.84**
t0	17								0.14	0.43	0.75	1.84**
c1	17									0.29	0.61	1.69**
z1	14										0.32	1.40**
c0	16											1.09
h1	16											

[a]Task codes: c = cutting, f = folding; h = hammering; s = bit changing; t = tying; z = zipping.
[b]View domside codes: 0 = non-dominant side; 1 = dominant side.
[1]The minimum value of *relative* h^* was 1.39 mm for the non-dominant arm during screwdriver bit changing. Differences are presented in increasing order, left to right.
**Highly significant difference, $p < 0.01$; *Significant difference, $p < 0.05$.
N = number of samples considered.

https://doi.org/10.1017/S0263574704001341 Published online by Cambridge University Press

Table VII. Inconsistency index of qualitative characteristics of zipping task for non-prosthesis users.

Characteristics (yes = 1; no = 0)	class	Non-prosthesis users' summary data				Prosthesis users (normalised ICI)		
		mean	std. dev.	95% CI		fp1a	fp2a	mp01
change in # of pulls to open (/3)	movement	0.12	0.16	0.05	0.20	0	0	0
change in # of pulls to close (/3)	movement	0.10	0.16	0.04	0.18	0.33	0.33	0.33
zipper sticky (/4)	material	0.09	0.12	0.04	0.15	0.25	0.25	0
switch role of hands (pull/hold)	method	0.03	0.11	0.00	0.13	0	0	0
change hand placement as pull	movement	0.05	0.15	0.01	0.16	0	0	0
hesitation (/4)	movement	0.04	0.09	0.01	0.10	0	0	0

Table VIII. Inconsistency index of qualitative characteristics of screwdriver task.

Characteristics (yes = 1; no = 0)	class	Non-prosthesis users' summary data				Prosthesis users (normalised ICI)		
		mean	std. dev.	95% CI		fp1a	fp2a	mp01
front-back uniform orientation	method	0.05	0.15	0.01	0.16	0	0	0
loosen with 2 hands (vs. 1)	method	0.10	0.21	0.03	0.22	0	0	0
loosen with wrist (vs. fingers)	method	0.20	0.30	0.10	0.30	0	0	0
tighten with 2 hands (vs. 1)	method	0.18	0.34	0.05	0.28	0	0	0
tighten with wrist (vs. finger)	method	0.15	0.33	0.03	0.25	0	0	0
place bit separately from screwdriver	method	0.23	0.41	0.08	0.35	0	0	0
twist wrong way	movement	0.33	0.44	0.18	0.48	0.25	0	0.25
hesitate or fumble (/4)	movement	0.11	0.17	0.05	0.18	0	0.25	0.25
change grasp from bit to handle	method	0.10	0.26	0.00	0.18	0	0	0

Table IX. Inconsistency index of qualitative characteristics of cutting task.

Characteristics (yes = 1; no = 0)	class	Non-prosthesis users' summary data				Prosthesis users (normalised ICI)		
		mean	std. dev.	95% CI		fp1a	fp2a	mp01
change in # of sawing motions (/7)	movement	0.26	0.15	0.21	0.32	0.57	0.29	0.29
cut horizontally (vs. vertically)	method	0.05	0.22	0	0.10	0	0	0
change length of cutting motion	movement	0.20	0.30	0.08	0.30	0	0	0
rotate steak between cuts	method	0.03	0.11	0.00	0.12	0.25	0	0
Plasticine sticky (/4)	material	0.56	0.37	0.44	0.70	0	0.25	0.25
non-standard hold of knife and fork	method	0.00	0.00	–	–	0	0	0.5
Hesitations or errors (/4)	movement	0.13	0.19	0.05	0.19	0	0	0
plate slipped	material	0.10	0.26	0.00	0.18	0	0	0.5

changing characteristics were method-based (Table VIII). The greatest observed ICI for non-prosthesis users was associated with twisting the screwdriver the wrong way to loosen it, potentially occurring nearly half the time (48% upper 95% confidence interval limit).

Cutting task: Over the four repetitions of the cutting task, 8 slices were cut, allowing at most 7 changes in the number of sawing motions. Prosthesis users varied the number of sawing motions to cut the slices more than non-prosthesis users (on average 0.36 versus 0.26 ICI), although other aspects were more consistent for prosthesis users (Table IX). The plasticine was stickier for non-prosthesis users, possibly because it had insufficient time to cool between trials with this group. Changes in cutting direction or plate slippage could be absent for both populations. Further, the way in which the knife and fork were held was the same, with the exception if one instance with a prosthesis user.

Hammering task: Inconsistency in number of impacts used and the frequency of hesitations was high among non-prosthesis users and higher still among prosthesis users (Table X). Missing the nail was similarly frequent overall, although regripping (a method variation) was less frequent on average among prosthesis users. Among non-prosthesis users, the number of impacts was significantly more among women than among men (Kruskal-Wallis test of ANOVA on number of impacts by gender, $p < 0.01$).

Table X. Inconsistency index (ICI) of qualitative characteristics of hammering task.

		Non-prosthesis users' summary data				prosthesis users (normalised ICI)		
Characteristics (yes = 1; no = 0)	class	mean	std. dev.	95% CI		fp1a	fp2a	mp01
change in # impacts (/3)	movement	0.73	0.30	0.63	0.85	0.67	1	1
cross hammer/hands at start or end	movement	0.15	0.29	0.05	0.25	0	0	0
regrip hand holding hammer during task	method	0.20	0.30	0.08	0.30	0	0	0.25
regrip hand holding nail during task	method	0.13	0.28	0.03	0.23	0	0	0
hesitate in movement (/4)	movement	0.44	0.36	0.31	0.56	0.5	0	1
miss impact (/4)	movement	0.16	0.25	0.08	0.25	0	0	0.5
wood of different hardness	material	0.05	0.15	0.01	0.16	0	0	0

Table XI. Inconsistency index (ICI) of qualitative characteristics of folding task.

		Non-prosthesis users' summary data				Prosthesis users (normalised ICI)		
Characteristics (yes = 1; no = 0)	class	mean	std. dev.	95% CI		fp1a	fp2a	mp01
fold away-toward (vs. away-away)	method	0.08	0.24	0	0.15	0	0	0
fill envelope flat (vs. angled/vertical)	method	0.05	0.15	0.01	0.16	0	0	0
place filled envelope at 100% reach	method	0.03	0.11	0.00	0.13	0.25	0	0
angle envelope to fold flap (not flat)	method	0.10	0.26	0.00	0.18	0	0	0
hesitations in movement (/4)	movement	0.08	0.12	0.03	0.14	0.25	0	0
extra movements (/4)	movement	0.05	0.13	0.00	0.09	0	0	0

Table XII. Inconsistency index (ICI) of qualitative characteristics of tying task for non-prosthesis users.

		Non-prosthesis users' summary data				Prosthesis users (normalised ICI)		
Characteristics (yes = 1; no = 0)	class	mean	std. dev.	95% CI		fp1a	fp2a	mp01
board moves	material	0.05	0.15	0.01	0.16	0	0	0
grasp string from below/side (vs. above)	method	0.18	0.29	0.08	0.28	0	0	0
segment movement amplitude	movement	0.00	0.00	–	–	0.75	0.25	0

Folding task: Material characteristics caused no inconsistencies during the folding task. Furthermore, this task had relatively low levels of inconsistency, with greatest expected inconsistency at 18% (upper 95% confidence interval limit) occurring when folding the flap (Table XI) for non-prosthesis users. Among prosthesis users, variations were similarly rare, placing the envelope at a reach other than 100%, and hesitating, occurring only once. These two inconsistencies occurred with the same prosthesis user subject, although on average, their frequencies were similar to those of non-prosthesis users.

Tying task: This task showed the smallest number of varying characteristics. Non-prosthesis users most frequently varied the hand orientation used to grasp the string, while prosthesis users showed no inconsistency there (Table XII). The only variation noted for prosthesis users was a change in segmental movement amplitude occurring when the hand of the prosthesis slipped. This grip-based inconsistency happened on average one-third of the time, although 3 of 4 of these observations occurred with the subject who was least familiar with his prosthesis.

IV. CONCLUSIONS

The following conclusions can be made as a result of this research:

(i) The mechanical energy of the upper limbs can be measured and estimated from potential energy alone and the results used to distinguish work related activities. Furthermore, a single composite energy value normalised on body mass and gravity, h^*, can characterise the mechanical energy efficacy of a goal-oriented ADL.

(ii) By incorporating specific adaptations to model the anthropometric differences of prosthesis users, the models for a general (non-prosthesis user) population can be applied for comparative study. For prosthesis users this means locating the prosthesis centre of gravity and adjusting segmental limb mass estimations for the combined residual limb and prosthesis.

(iii) Task, arm dominance, task* arm dominance, subject, and gender are significant determining factors of h^*. Subject ceased to be a significant factor when considering *relative h** and gender significance is eliminated by normalising on segmental mass, total body mass and fingertip arm reach.

(iv) Qualitative analysis found that prosthesis users are more consistent in their methods than non-prosthesis users.

(v) The difference between dominant and non-dominant arm potential energy over the tasks studied was greater among prosthesis users than it was among non-prosthesis users, indicating greater asymmetry in composite upper limb segment height.

Admittedly, three prosthesis users is too small a group to be able to generalise. However this method measures differences in how these individuals perform tasks, and shows that the method can be used to quantify changes. No estimates of statistical significance have been derived from the prosthesis user group.

As a result of the higher degree of inter-repetition consistency and greater dominant – non-dominant arm asymmetry, upper limb prosthesis users are at greater risk of cumulative trauma disorders. By comparing individuals of working age using below-elbow powered prostheses and a similar normally limbed population, differences can be quantified and used to improve prosthesis design and training methods. These methods can help to minimise the additional physiological costs associated with typical working tasks when performed by an amputee, thus minimising fatigue and over-use injuries.

Acknowledgements

Funding for this research was supplied by the Medical Research Fund of New Brunwick, the Association of Professional Engineers and Geoscientists of New Brunswick Advanced Studies Scholarship, and the New Brunswick Women's Doctoral Scholarship. We acknowledge with thanks Dr. Jacques Allard, professor of Mathematics and Statistics at Université de Moncton, for his statistical expertise in calculating confidence intervals for the inconsistencies in qualitative characteristics.

References

1. E. R. Sanderson and R. N. Scott, *UNB Test of Prosthetics function: A test for unilateral upper extremity amputees, Ages 2–13* (University of New Brunswick, Bio-Engineering Institute, September, 1985).
2. R. Ballance, B. N. Wilson and J. A. Harder, "Factors affecting myoelectric prosthetic use and wearing patterns in the juvenile unilateral below-elbow amputee", *Canadian Journal of Occupational Therapy* **56**(3), 132–137 (1989).
3. D. J. Atkins, D. C. Y. Heard and W. H. Donovan, "Epidemiological overview of individuals with upper limb loss and their reported research priorities", *Journal of Prosthetics and Orthotics* **8**(1), 2–11 (1996).
4. P. J. Kyberd, D. J. Beard, J. J. Davey and J. D. Morrison, "A survey of upper-limb prosthesis users in Oxfordshire", *Journal of Prosthetics and Orthotics* **10**(4), 85–91 (1998).
5. A. van Lunteren, G. H. M. van Lunteren-Gerritsen, H. G. Stassen and M. J. Zuithoff, "A field evaluation of arm prostheses for unilateral amputees", *Prosthetics and Orthotics International* **7**(3), 141–151 (1983).
6. I. Gilad, "Objective performance analysis of artificial hands: towards improvement of function", *Ergonomics* **29**(4), 553–561 (1986).
7. L. E. Jones and J. H. Davidson, "Save that arm: a study of problems in the remaining arm of unilateral upper limb amputees", *Prosthetics and Orthotics International* **23**, 55–58 (1999).
8. C. Hartelt, "Ouch!" *Credit Union Management*, Madison **22**(3), 18–21 (1999).
9. J. M. Muggleton, R. Allen and P. H. Chappell, "Hand and arm injuries associated with repetitive manual work in industry: a review of disorders, risk factors and preventive measures", *Ergonomics* **42**(5), 714–739 (1999).
10. H. J. B. Day, "Amputee rehabilitation – Finding the niche (The Knud Jansen Lecture)", *Prosthetics and Orthotics International* **22**, 92–101 (1998).
11. D. J. Atkins, "Adult upper limb prosthetic training", In: (D. J. Atkins and R. H. Meier III, Eds.) *Comprehensive Management of the Upper Limb Amputee* (Springer-Verlag, New York, 1989) Chapter 6, pp. 39–54.
12. P. J. Kyberd, D. J. Beard and J. D. Morrison, "The population of users of upper limb prostheses attending the Oxford Limb Fitting Service", *Prosthetics and Orthotics International* **21**, 85–91 (1997).
13. G. E. Jacques, S. Naumann, M. Milner, W. Cleghorn, G. Bush and S. Hubbard, "Visual feedback and the grasping function of prosthetic hands", *Proceedings of the Myoelectric Control Symposium*, Fredericton, NB: University of New Brunswick (1994) pp. 97–104.
14. J. R. Williams, "Some Aspects of the Biomechanics of the Elbow Joint: Relation to prosthetic design", *Doctor of Medicine Thesis* (Oxford: University of Oxford, 1996).
15. N. L. Black, Mechanical Energy Measurement Applied to Upper Limb Powered Prostheses *Ph.D. Thesis* (Fredericton, NB: University of New Brunswick, 2001).
16. J. E. Edelstein and N. Berger, "Performance comparison among children fitted with myoelectric and body-powered hands", *Archives of Physical Medicine and Rehabilitation* **74**, 376–380 (April, 1993).
17. R. H. Jebsen, N. Taylor, R. B. Trieschmann, M. J. Trotter and L. A. Howard, "An objective and standardized test of hand function", *Archives of Physical Medicine and Rehabilitation* **50**, 311–319 (June, 1969).
18. J. S. Landry and E. N. Biden, "Optimal Fixed Wrist Alignment for Below-elbow Powered Prosthetic Hands", *MEC 02 "The Next Generation" Proceedings of the Myo-Electric Controls / Powered prosthetics Symposium*, Fredericton, NB: University of New Brunswick (2002) pp. 12–14.
19. N. Diffrient, A. R. Tilley and J. C. Bardagjy, *Humanscale 1/2/3 Manual* (A project of Henry Dreyfuss Associates, MIT Press, Cambridge, MA, 1974).