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DEFINITION OF A PROTOCOL FOR GEOMETRIC AND KINEMATIC MEASUREMENTS TO ASSESS WHEELCHAIR PROPULSION

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Abstract − We developed a protocol for the assessment of manual wheelchair set-up and propulsion in a common clinical motion analysis laboratory. We also designed a device to detect hand contact on handrim.

In a first phase, we took anthropometric and wheelchair geometric measures. Later, subjects propelled the wheelchair and, based on the experimental data on subjects' movements and their effect on wheelchair velocity, we identified a number of indicators for performance in wheelchair push. We show that these indicators clearly distinguish between experienced and novice users.

Keywords (up to three): wheelchair, movement analysis

1. INTRODUCTION

Wheelchair propulsion is as essential for wheelchair users as walking is for unimpaired subjects. Due to the repetitive and strong load that wheelchair users have to face while pushing [1], they often experience upper limb injuries, such as carpal tunnel syndrome, impaired lower median nerve function or shoulder pain. The later often force them to immobility and therefore reduce their autonomy.

Wheelchair propulsion involves a *push* and a *recovery* phase. To prevent long term injuries, it is usually recommended that the relative duration of push is as large as possible, while the push frequency should be kept to a minimum[2].

Wheelchair therapists may achieve this goal by optimizing the position of seat and backrest, and by providing appropriate user training. Quantitative methods would be potentially capable of guiding therapists in both these aspects.

Manual wheelchair propulsion is typically assessed by kinetic measurement devices [3], e.g. instrumented wheels which measure forces, torques, and speed (through an encoder) but do not take into account users' kinematics. Such dedicated instruments require replacement of the original wheels, which is not always practical in clinical settings. In contrast, optoelectronic devices appear very practical for this application, as they allow to quickly take geometric, anthropometric and kinematic measurements.

While there are standard protocols for gait analysis, based on 3D stereophotogrammetry, no such protocols are currently available for wheelchair propulsion.

In order to optimise seat and backrest configuration, not only user posture but also wheelchair geometry should be monitored and assessed quantitatively. Therefore, a movement analysis protocol should provide static measurements, both user-related (e.g. length of body segments) and wheelchair-related (e.g. shoulder height, which depends on seat height).

Currently, performance in wheelchair propulsion is measured by the fraction of effective force (FEF) and by the amount of oxygen uptake [4].

Propulsion can be characterized in terms of arm and hand kinematics, as a sequence of five sub-phases[5]: (i) early push, (ii) late push, (iii) follow-thru, (iv) hand recovery and (v) pre-push. During pre-push, the hand moves forward in order to prepare for a new contact with the handrim. Detailed kinematic measurements of this sequence, and in particular the relation between hand and handrim coupling and wheelchair speed could potentially allow a closer look at the determinants of propulsion efficiency and subjects' skill.

The kinematics of wheelchair push has been frequently studied in altered conditions, e.g. through the use of simulators, ergometers or putting the wheelchair over rolls. However, these methods do not account for the contribution to propulsive torque due to head, arm and trunk movements during the recovery phase, as reported by [6] in the analysis of a single athlete sprint start on a racing wheelchair.

For these reasons, we designed a protocol in which the wheelchair had to actually move within the workspace of a gait analysis apparatus. We measured over-ground wheelchair speed and the intra-push velocity profile of the wrist in a population of able-bodied or spinal cord injury (SCI) subjects, operating the wheelchair at their preferred speed.

2. MATERIALS AND METHODS

2.1 Subjects

A total of twenty subjects (13 M, 7 F, age 37 ± 11) participated in this study. Twelve were able-bodied users, eight had SCI with different level of lesions.

Able bodied wheelchair therapists (5 subjects) and users with more than six months of experience in wheelchair propulsion (5 subjects) were classified as experts (E group), while able-bodied subjects without any previous experience

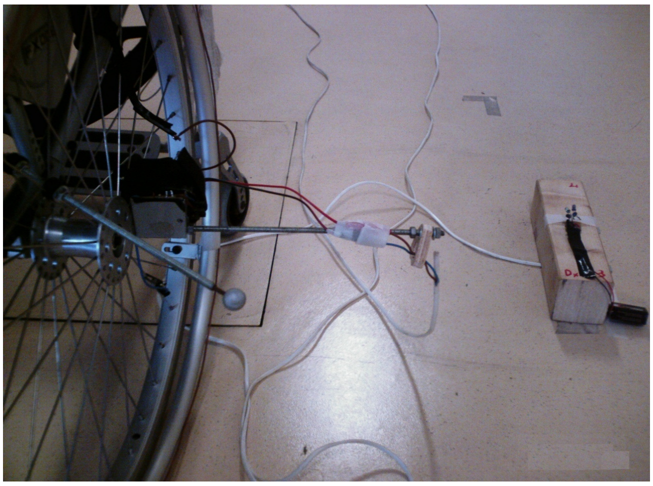

Fig 1 Capacitive sensing touch switch device mounted on a wheel, controlling a IRLED (on the right). A thin conductive wire, applied on the edge of the handrim, is used as touch sensor

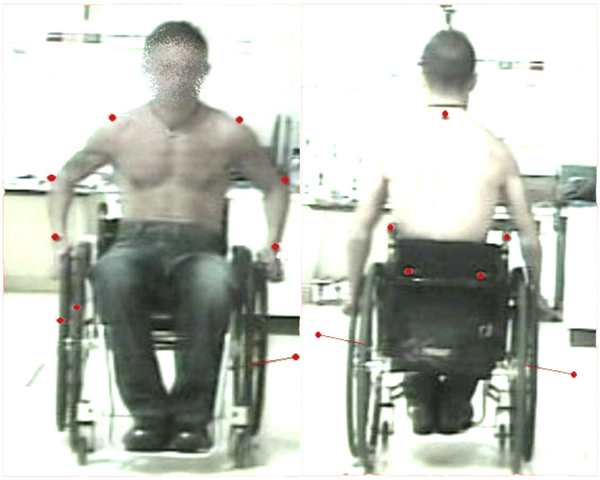

Fig. 2 Markers positioning on subject body and wheelchair. Four markers on the backrest identify a local reference system.

in manual wheelchair propulsion (7 subjects) and SCI subjects with less than 6 months of experience (3 subjects), formed the novice users group (N).

2.2 Experimental apparatus

All able-bodied subjects used the same ultra-light wheelchair in the same set-up, while spinal cord injured people used their own, fitted wheelchair.

We used a motion analysis system (Elite, BTS srl, Milan), with passive markers and 6 IR cameras, with an acquisition volume of about 3 × 1 × 1 m and a sampling frequency of 100 Hz.

We also developed a capacitive sensing device to detect contact between skin and the handrim. Depending on the wheelchair electrical characteristics (handrim and wheel conductivity), we had to place a thin conductive wire over the handrim to make the device work properly. Finally, the *touch switch* output was connected to two IRLEDs (one per each side) which were recognized as markers, thus appearing only during the push phase; this guaranteed the synchronization with the motion analysis system. We assumed push start as the handrim contact and push end as handrim release.

2.3 Geometric and anthropometric measurements

All subjects first performed a *static* test, with markers placed on their upper limbs (ulnar styloid, lateral epicondyle and acromion), trunk (C7), and on the wheelchair (four on the backrest, two on the axles, one on the seat, one on a handrim and one on a wheel).

Subjects were asked to spread their arms and then touch the handrims three times. We measured arm length as the maximum distance between the acromion and ulnar styloid marker and subject relative-to-ground height, as the maximum height of the C7 marker. We also measured wheelchair dimensions (seat absolute and relative to axle height and depth, backrest height, handrim and wheel diameters).

We measured elbow extension angle (on the second and third contact) too, since it is recommended that this value should be in a range of 80°-100° with the hand staying on the top dead center of the wheel[7].

2.4 Kinematic measurements

Due to touch switch incompatibility with the handrim and wheel electrical characteristics of some of the wheelchairs, handrim contact was only recorded in seven subjects in E group (five wheelchair therapists and two SCI subjects) and six (all able-bodied) in the N group.

In this phase, we left only four wheelchair backrest markers, which identified a local reference frame, whose origin was the mid-point between the lower markers on the backrest. The three axes were directed, respectively, toward the lower right marker on the backrest, toward the midpoint of the upper markers on the backrest, and orthogonal to these in anterior direction. We then expressed all subsequent movements of the subject during wheelchair propulsion with respect to this reference frame.

Subjects operated the wheelchair at a preferred velocity, by starting and stopping within about 2 m outside the acquisition space. The exercise was repeated for 6 times (the first was for familiarization).

Marker trajectories were low-pass filtered with a 4th order Butterworth filter, with a cut-off frequency of 7 Hz. We then estimated wheelchair speed as the 1st derivative of the antero-posterior component of the trajectory of the middle point of the backrest.

Wheelchair propulsion analysis is usually performed at a specified, fixed wheelchair speed. However, propulsion kinematics varies depending on speed [8]. In these experiments we measured the average speed of the wheelchair, s. Moreover, to check whether subjects were still accelerating or rather they were maintaining a steady state velocity, we also looked at wheelchair speed when entering, s_{in}, and exiting, s_{out}, the acquisition space.

Thus, deceleration during the push meant inefficacy, and

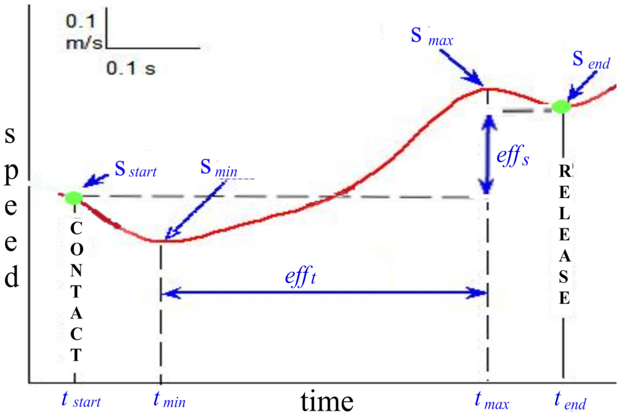

Fig. 3 Intra-push velocity profile. Note that because of their definition, both efficacy indicators are minor or equal to 1, and will further be indicated as %

we defined push efficacy in time (*eff$_t$*) as the ratio between the period in which the velocity was effectively increasing and the total push duration:

$$eff_t = \frac{t_{\max} - t_{\min}}{t_{end} - t_{start}} \quad (1).$$

Furthermore, we defined push efficacy in speed (*eff$_s$*) as the ratio between the variation in speed due to push (final minus starting velocity) and the difference between maximum and minimum speed recorded during the push:

$$eff_s = \frac{s_{end} - s_{start}}{s_{\max} - s_{\min}} \quad (2).$$

We calculated the wrist speed as the 1st derivative of the antero-posterior component of the trajectory of the ulnar styloid process marker and took its magnitude just before contact (*s$_{wr}$*).

We reasoned that expert users exhibit a better hand coupling (thus having greater *s$_{wr}$*), and a greater efficacy in both speed and time. Also, we assumed that *s$_{wr}$* is an indicator of wheelchair braking.

We used one-way ANOVA to test for between-group differences.

3. RESULTS AND DISCUSSION

3.1 Geometric and anthropometric measures

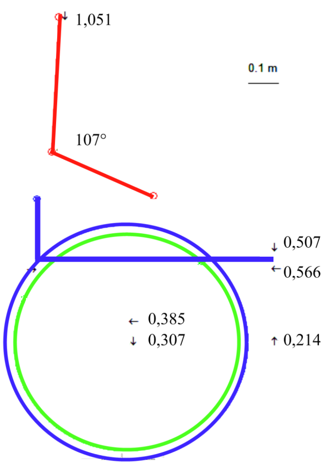

Fig. 4 Graphical report of user's posture and the wheelchair dimensions and setup.

Fig.4 shows the measures obtained for a spinal cord subject on his own fitted wheelchair. We reported backrest and seat coordinates relative to the axle-hub, since both their positions are typically adjustable by the wheelchair therapists. For instance, seeing that the elbow extension angle has a value higher than the recommended 80-100°, one should lower the seat to achieve this.

3.2 Kinematic measures

Table 1 mean speed values and efficacy indicators

	E	NE
s_{in} $[m \cdot s^{-1}]$	1,2 ± 0,28	1,1 ± 0,11
s_{out} $[m \cdot s^{-1}]$	1,44 ± 0,35	1,21 ± 0,11
s $[m \cdot s^{-1}]$	1,36 ± 0,28	1,19 ± 0,11
eff_t [%]	71,3 ± 6,8	56,4 ± 6,7
eff_s [%]	74,2 ± 8,2	44,1 ± 10,4
s_{wr} $[m \cdot s^{-1}]$	0,16 ± 0,41	-0,56 ± 0,32

We collected an average of 12 pushes per subjects. We first looked at the overall speed of the wheelchair.

Comparing the speed recordings in Tab.1 to the 1,06 [m/s] needed to safely cross an intersection which has already been used as threshold in wheelchair propulsion analysis [9], we found that all subjects except one (non-expert) overcame that value, already at the start of the acquisition. This appears to confirm the feasibility of wheelchair propulsion assessment in a common gait analysis laboratory.

Both efficacy in time and efficacy in speed were significantly higher in the E group ($p=0{,}0001$ and $p=0{,}002$).

Also, experts approached the handrim with a higher speed ($p=0{,}005$), and this suggests that experience is predictive of an effective hand coupling. Novice subjects approached the handrim with a negative wrist speed, that is to say without a pre-push phase.

Speed of the wrist at contact is also likely to be a major determinant of *eff$_s$*. In fact, we found a strong, significant correlation ($R=0{,}90$, $p<0{,}05$) between wrist speed and speed efficacy; see Fig. 5.

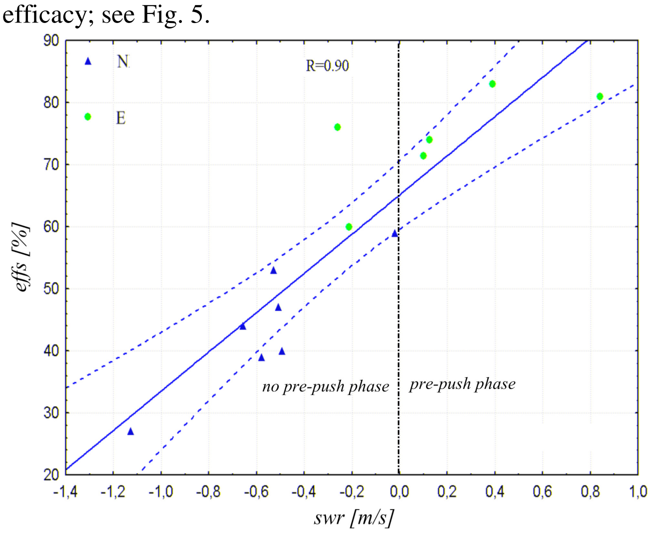

Fig 5 Correlation between wrist speed at contact and speed efficacy.

4.CONCLUSION

We developed a measurement protocol to acquire a set of *static* measures and kinematic indicators of propulsion using optoelectronic systems, in order to support wheelchair therapists. The proposed simple kinematic indicators, namely wrist speed at contact, push efficacy in time and push efficacy in speed are capable of discriminating between expert and novice users.

In [10] a braking axle moment was measured at contact occurrence, only in multiple sclerotic subjects, while spinal-cord injured did not show this. Their experiment was however conducted on a dynamometer. We instead observed that wheelchair velocity decreases during the push (*braking effect*). Moreover, this effect relates to poor hand coupling at handrim contact.

We therefore suggest that handrim grasping should be more investigated, since the initial phase of push seems to be crucial to its efficacy. Also, considering that a wrong hand movement during the pre-push phase leads to stronger collisions with the handrim, further investigations should be conducted to verify if this could be a cause for the wrist pathologies we mentioned in introduction as more frequent in wheelchair users populations.

Therefore, we conclude that 3D movement analysis is a suitable method to analyze wheelchair set-up and evaluate propulsion mechanics.

Furthermore, we applied capacitive sensing technology to manual wheelchair propulsion to identify the exact timing of the push phase. We suggest that the same technology could be easily applied to wheelchairs during daily activity to monitor the temporal features of wheelchair propulsion.

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