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INFLUENCE OF A HYBRID MANUAL–ELECTRIC WHEELCHAIR PROPULSION SYSTEM ON THE USER'S MUSCULAR EFFORT

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Abstract: Self-propelled wheelchairs favour the rehabilitation process, forcing the user to be physically active. Unfortunately, in most cases, the manual propulsion is not adapted to the individual needs and physical capabilities of the user. This paper presents the results of operational tests of a wheelchair equipped with a hybrid propulsion system in which the muscle strength generated by the user is assisted by two independent electric motors. The research aimed to investigate the influence of the applied control algorithm and the assistance factor (W) on the value of the muscular effort (MA) while propelling the wheelchair with the use of push rims. A modified ARmedical AR-405 wheelchair equipped with two MagicPie 5 electric motors built into the wheelchair's hubs with a power of 500 W was used in this research. The tests were carried out on a wheelchair test bench simulating the moment of resistance within the range of 8–11 Nm. Surface electromyography was employed for the measurement of MA, specifically, a four-channel Noraxon Mini DTS apparatus. The research was carried out on five patients from the group of C50 anthropometric dimensions. The effort was measured for four muscles: deltoid–anterior part, deltoid–posteriori part, and triceps brachii and extensor carpi radialis longus. The effectiveness of the hybrid propulsion system was observed based on the extensor carpi radialis longus muscle. In this case, for the standard wheelchair, the MA ranged from 93% to 123%. In contrast, for a wheelchair equipped with the hybrid propulsion system, at W = 70%, the MA was within the range of 43%–75%.

Key words: assistive technology, wheelchair, electromyogram (EMG), electric propulsion, control

1. INTRODUCTION

The wheelchair is the basic device enabling the movement of people with physical disabilities. Over the years, wheelchairs have developed in many directions, increasing their functionality through the ability to overcome obstacles (stairs [1,2], thresholds [3], hills [4]) or improving basic functions such as locomotion. The locomotive function of the wheelchair is strongly dependent on its structure and was developed towards reducing weight and size [5], strengthening the frame [6] and adjusting the motion control function depending on the person's disability [7]. Structural improvement increasing the safety of movement [8,9] and systems facilitating manual propulsion of a wheelchair (e.g. through the use of gears) are also subject to research and analysis [10]. However, the main development trend is the use of electric propulsion systems [11,12]. Among the many disadvantages of wheelchairs powered solely by electric motors (greater weight [13], like the control learning problems [14–16], and difficulties in transporting electric wheelchairs in public means of transport [17], the greatest disadvantage is the reduction of the user's motor activity. This leads to a limitation of physical rehabilitation, which in turn may predispose the user to multiple long-term health problems [13]. Therefore, more work is currently being devoted to designing hybrid drive systems. When it comes to wheelchairs, a parallel hybrid drive (manual–electric) is the most advantageous [18]. In the literature, these solutions are also described as assistive technology solutions [19]. The main task of these systems is to assist the movement of people with physical disabilities, and not to completely exclude their physical activity.

Wheelchairs equipped with assistive technology solutions can assist the wheelchair users' effort by adding tractive torque of a specific value transferred by electric motors to the wheelchair's drive wheels. This value can be defined by the user [18,20,21], or set by the control algorithms based on signals from additional sensors, for example, gyroscopic systems recognising slopes [18,20]. Although solutions enabling the setting of the wheelchair drive assistance level are presented in works by Cooper et al. 2002, Oh and Hori 2013, Oh et al. 2014, no studies have shown the effect of these settings on the muscular effort parameters.

Assistive technologies make it possible to reduce the force required to generate by the operator of a machine or device, and are used and popular in many different areas of life. Examples of such systems are the steering assist system [22–24] and the brake pedal control system [25–27] in motor vehicles, electric bicycle drive systems [28–30] or rehabilitation support devices [31]. Too low value of the supporting force may hinder or prevent the performance of tasks, and in extreme cases lead to overloading the operator's musculoskeletal system. On the other hand, too great a value of supporting force may also be problematic, leading to a loss of control precision or the ability to assess the operation of the system whose control process is being carried out. Excessive support in muscle-driven propulsion systems can also overload the operator's musculoskeletal system. This paper presents the influence of the propelling force assistance factor W settings on the muscular effort of the upper limbs of the wheelchair user. In the work, the propelling force assistance factor defined the value of the maximum flow rate of electric current supplied to the electric motors installed in the wheelchair's wheels. The test was carried out for various wheelchair movement resistances, thus mapping various conditions of use, such as driving uphill or driving on unpaved surfaces.

2. METHODS AND MATERIALS

2.1. Researched patients

Five men took part in the tests (Tab. 1) and they were categorised according to height, weight, age, maximum push force of the upper limb and wheelchair experience. To ensure comparability between patients, their dimensions reflected the 50th percentile of anthropometric dimensions according to the European standard NEN-EN 979. In order to ensure that people with a similar physical condition were selected for the study, participants in the same age range and having a similar value of the force generated by the upper limb were selected. The measurement method for the push force was formalised in terms of methodology. A special stand was used on which the user, in a seated position, pushed a handle connected to a strain gauge towards his knee (Fig. 1). The evaluation of the participants' experience was based on a five-point scale. This assessment was performed by the examined patients, taking into account the time of using the wheelchair, the variety of places of its use and the general confidence in moving in a wheelchair. Each patient was familiarised with the test procedure and had to fill out a consent form to participate in the research. The authors decided to conduct the research on patients without physical disabilities due to the use of prototype propulsion system solutions in difficult terrain conditions. The research and experimental protocols have been positively evaluated by the Bioethical Commission at the Karol Marcinkowski Medical University in Poznań Poland, Resolution No. 513/21 of 24 June 2020, under the guidance of Prof. M. Krawczyński for the research team led by M. Kukla. The authors obtained written consent of the researched person for publication of the research test performed with their participation. The data were presented in such a way as to ensure complete anonymity. The measurement method and data acquisition were carried out following the directives of the Bioethics Commission at the Karol Marcinkowski Medical University in Poznań Poland, which are in line with the guidelines of the Helsinki Declarations.

Tab. 1. Comparison of anthropometric features and the level of experience in wheelchair operation of the test subjects. Mean values determined with the 95% confidence interval (p = 0.05)

	Height	Weight	Age	Push force	Experience
	cm	*kg*	*Years*	*N*	*[-]*
Subject 1	183	90	32	364	●●●●●
Subject 2	179	88	33	322	●●●●●
Subject 3	175	110	31	298	●●●●○
Subject 4	178	96	30	309	●●●○○
Subject 5	171	93	33	306	●●●○○
average	176 ± 3	86 ± 9	33 ± 2	302 ± 23	–

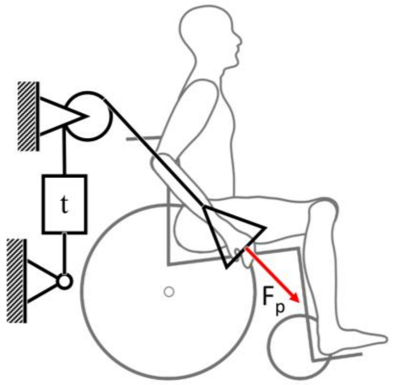

Fig. 1. Schematic illustration of the push force measurement system, where t is strain gauge force sensor and Fp is push force

2.2. The tested wheelchair

The tests were carried out using an ARMedical AR-405 wheelchair equipped with a prototype hybrid propulsion system module (parallel manual–electric hybrid) (Fig. 2) (Patent in the Patent Office of the Republic of Poland, no. exclusive rights PL 239350, 2021) and a conventional version of the AR-405 wheelchair.

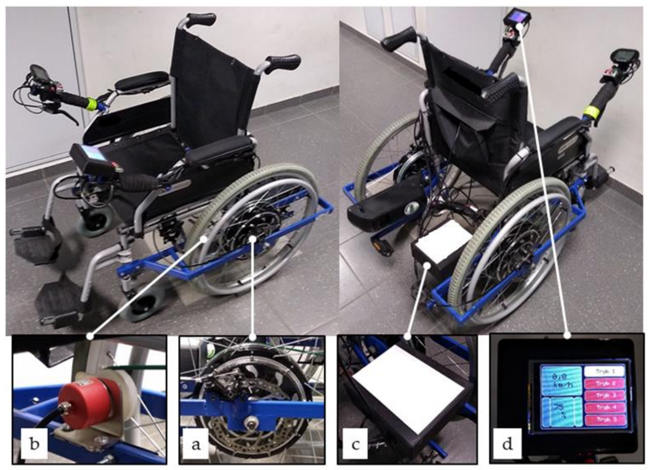

Fig. 2. The tested AR-405 wheelchair is equipped with a hybrid propulsion module. a – power of 500 W installed in the drive wheel hubs; b – incremental encoders; c – proprietary control system with a gyroscope; d – touch screen controller

The modified wheelchair was equipped with two BLDC MagicPie 5 motors by Golden Motor, with a power of 500 W installed in the drive wheel hubs (a), as well as incremental encoders (b), a proprietary control system with a gyroscope (c) and a touch screen controller (d). The screen was used to select the assistance mode and the value of the assistance gain factor W. The wheelchair control system was based on a 32-bit STM32F407 microcontroller operating at a frequency of 100 MHz. The main task of the control system was to control the rotational speed of the two BLDC motors connected with the wheel rims. The speed was measured by two incremental encoders connected to the digital inputs configured in counter mode. The signal generated by the encoder was counted in quadrature, which increased the accuracy of the measurement and allowed to determine the direc-

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tion of rotation. The speed value was determined with a base of 30 ms, using an independent counter.

The heart of the hybrid propulsion system was the algorithm reducing the resistance to movement of the wheelchair resulting from changing operating conditions. To move the wheelchair, a driving torque M_N (1) was generated, being the sum of the electric motor torque M_{SE} and muscular strength M_M. The M_M was only part of the total driving torque required to overcome the resistance to motion:

$$M_N = M_{SE} + M_M \tag{1}$$

In the adopted resistance to motion reduction algorithm, the driving torque generated by the incorporated electric motors results from the value of the constant assistance controlled using the assistance factor *K* and the slope angle (Fig. 3).

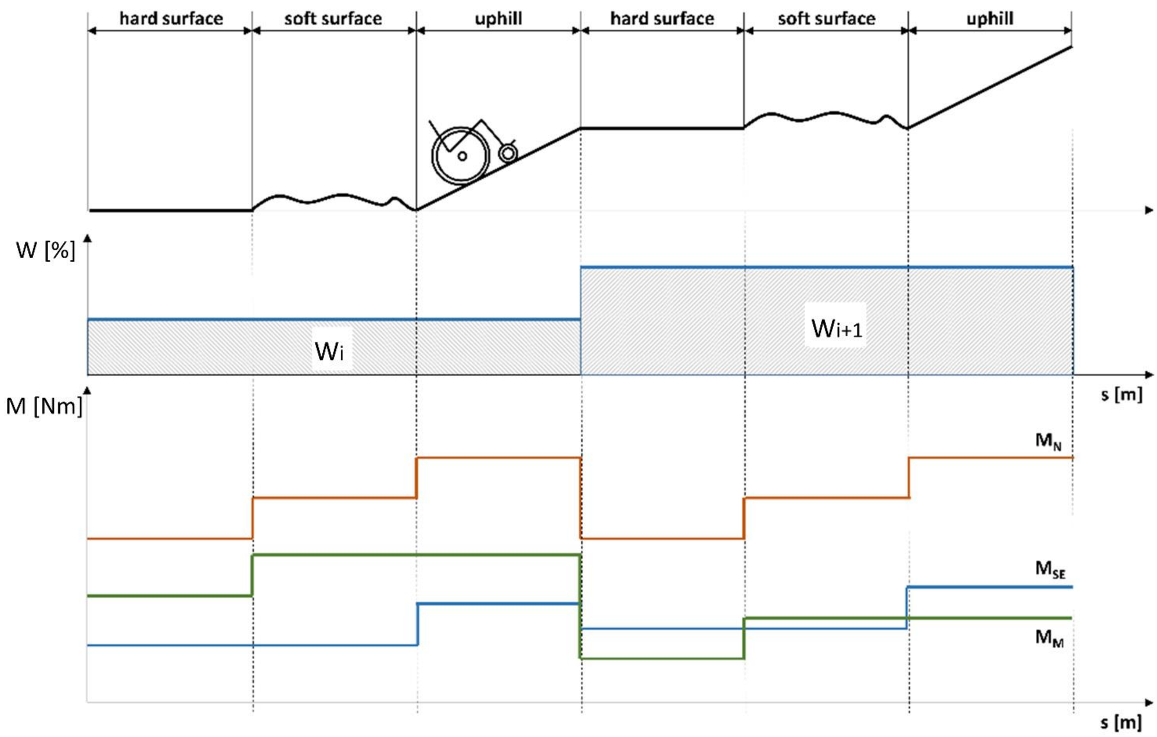

Fig. 3. Schematic diagram of the driving torque MN, muscle strength moment MM, and torque of the electric motor MSE, depending on the assistance factor w and the terrain conditions

In this mode, the system itself adjusts the reduction of resistance resulting from the slope of the terrain, whereas the user determines the level of reduction of resistance resulting from the type of the surface. The resistance reduction mode does not allow the wheelchair to be propelled solely by electric motors. To induce movement of the wheelchair in this mode, it is required to supply muscular force to the push rings of the wheelchair.

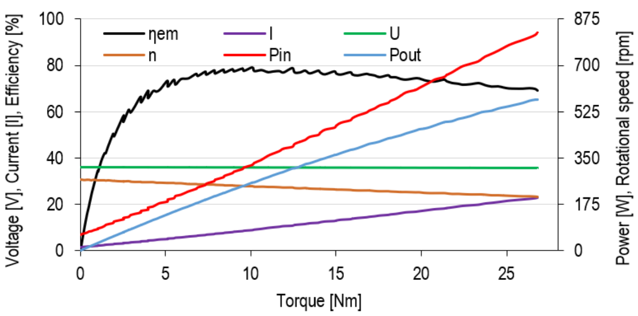

Fig. 4. Characteristics of the MagicPie 3 system by Golden Motor; based on the manufacturer's data; η_{em} is the efficiency of electric motor, n is rotational speed, I is electric current, P_{in} is input power, P_{out} is output power and U is voltage

According to the above algorithm, the share of torque provided by muscles depends on the assist factor *W* set by the user. The user can set the *W* factor in a range from 0% to 100%, where 0% meant no electric motor was involved in propelling the wheelchair. The value of assistance factor *W* was used to determine the value of the electric motors' control current. The control current was calculated based on the equation taking into account experimentally determined correction factors (2). It should be noted that the formula for the control voltage *u* was determined experimentally and allowed to calculate the value of the voltage in the range from 0 V to 3 V. In this equation, the variable is the angle of inclination of the wheelchair α and the assistance factor *W*. In this system, the driving torque of the engine (Fig. 4) is proportional to the value of the control signal *u*.

$$u = \alpha \cdot c + r_3 \cdot W + \delta, \tag{2}$$

where *u* is the signal sent to the BLDC motor controller, *α* is the wheelchair angle [°], c is the angle coefficient (0.0075), r_3 is the gain coefficient in mode 3 (0.0016), *W* is the assistance factor [%] and *δ* is the offset (1.102) [V].

2.3. Research methodology

The tests were carried out on the proprietary dynamometer for wheelchairs [32] which was used to simulate the resistance of movement of the wheelchair and its angle of inclination. During one measurement test, the participants performed 30 driving phases for 9 different configurations of the tested wheelchair. Participants performed driving phases to adjust their speed of propulsion to the pace set by the metronome at a rate of 45 beats per minute (BPM). Muscle activity (*MA*) performed during such propulsion phases was recorded by the myoMOTION software and then transformed and statistically processed.

The MA measurement that allowed to estimate the effort of the muscular system of the upper limb was performed using a Noraxon miniDTS surface electromyography apparatus, equipped with four measurement channels. The analysis and recording of the MA signal were performed with the supplied Noraxon MR3 software. The muscular effort analysis (*MA*) concerned four muscles that are involved in propelling the wheelchair: deltoid muscle anterior MA_1 and posteriori MA_2, triceps brachii MA_3 and extensor carpi radialis longus MA_4. These muscles were selected because, as shown in previous studies, they show the greatest activity when propelling a wheelchair [33].

Additionally, to determine the effort of the entire upper limb MA_{arm}, the measured effort of all measured muscles was averaged as [3]

$$MA_{arm} = \frac{\sum_{i=0}^{n} MA_i}{n} \cdot 100\%, \quad (3)$$

where MA_{arm} is the effort of the entire upper limb, MA_i is the muscular effort of the *i*-th measured muscle of the upper limb and *n* is the number of muscles measured on the upper limb.

Before starting the measurement of *MA*, each patient underwent a normalisation procedure following the guidelines of the electromyogram (EMG) apparatus manufacturer. Round electrodes covered with gel (20 mm in diameter) were used for the tests. They were placed in the central part of the examined muscle heads. The measurements were made at a frequency of 1,500 Hz. The research procedure assumed the performance of a normalisation procedure for each patient to check the value of the maximum voluntary contraction (MVC_{max}) of the examined muscles during the static test [33]. This procedure was performed 1 h before the actual *MVC* test. The purpose of the procedure was to determine the reference value (MVC_{max}) necessary for further calculations.

The measured EMG signals were rectified and then smoothed using RMS algorithms with a window width of 150 ms. The *MVC* test was then performed. This post-processing method utilises a reference value to normalise subsequent EMG data series [4]. The output is displayed as a percentage of the *MVC* value, which can be used to establish a common ground when comparing data between repetitions and individual patients:

$$MA = \frac{MVC}{MVC_{max}} \cdot 100\%, \quad (4)$$

where *MA* is the muscle effort, *MVC* is measured during the measurement test and MVC_{max} is the maximum voluntary contraction measured during the normalisation procedure.

The test procedure assumed that the participant would perform five pushes with his right hand at a frequency of 30 BPM. The entire test procedure was performed on a wheelchair dynamometer on which the resistance to motion was simulated. During the test, three values of the torque (*m*) mimicking the moment of the resistance force M_R were simulated: 8.14 Nm, 9.67 Nm and 11.19 Nm.

3. RESULTS

The research was performed on a group of five participants representing the same group of anthropometric dimensions. Each of the participants had the same four muscles tested, for which the average muscle effort is presented in Tab. 2. Muscle effort of the entire upper limb is also included in these tables. The estimation of the entire upper limb muscle effort MA_{arm} was a compilation of the average values of the muscular effort of all four measured muscles. To determine MA_{arm}, the confidence interval p = 0.05 was calculated using the *t*-Student test for the significance level.

To compare the results obtained for the wheelchair with the results obtained for the conventional wheelchair, the tests of the same muscles for the wheelchair were repeated for the wheelchair with a conventional manual propulsion system – also included in Tab. 2.

Tab. 2. Average results of MA from five participants for the tested values of the assistance factor W and the moment of resistance to movement m , where: MA_1 – MA of deltoid – anterior part, MA_2 – MA of deltoid – posterior part, MA_3 – MA of triceps brachii, MA_4 – MA of extensor carpi radialis longus, MA_{arm} – muscular effort of entire limb averaged over the measurement of four muscles. The value of MA_{arm} was determined with the confidence interval for the probability level p = 0.05 (MA, muscle activity)

W	m	MA_1	MA_2	MA_3	MA_4	MA_{arm}
-	*[Nm]*	[%]	[%]	[%]	[%]	[%]
0%	8.14	66	126	86	59	84 ± 14
	9.67	94	164	109	87	113 ± 16
	11.19	78	205	126	106	129 ± 26
25%	8.14	47	132	82	53	78 ± 18
	9.67	73	161	100	80	104 ± 19
	11.19	78	199	118	101	124 ± 25
30%	8.14	54	120	80	49	75 ± 15
	9.67	77	151	94	81	101 ± 16
	11.19	70	177	112	96	114 ± 21
40%	8.14	51	129	91	47	79 ± 18
	9.67	81	157	88	72	100 ± 18
	11.19	76	169	108	93	111 ± 19
50%	8.14	48	130	84	53	79 ± 18
	9.67	92	155	82	66	99 ± 18
	11.19	85	166	102	89	110 ± 18
60%	8.14	51	127	92	59	82 ± 16
	9.67	70	141	82	64	89 ± 16
	11.19	81	151	95	83	102 ± 15

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70%	8.14	42	151	90	62	86 ± 22
	9.67	66	147	88	60	90 ± 19
	11.19	78	122	87	82	92 ± 10
75%	8.14	50	129	87	66	83 ± 16
	9.67	67	141	96	72	94 ±1 6
	11.19	71	147	93	89	100 ± 15
Standard	8.14	66	135	53	92	87 ± 16
	9.67	90	140	83	113	107 ± 18
	11.19	83	143	87	127	110 ± 20

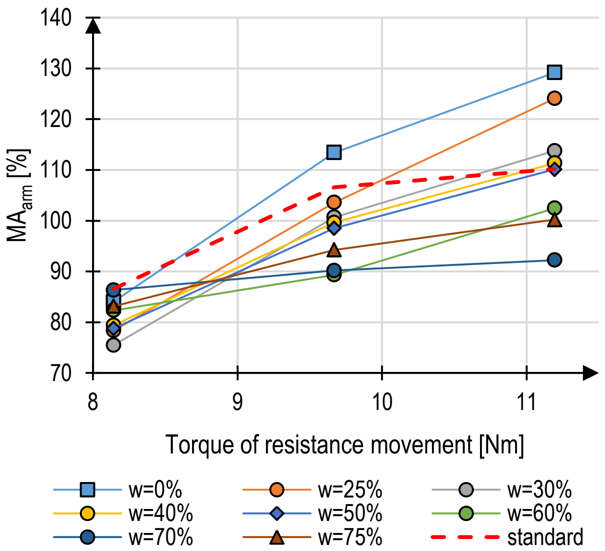

Fig. 5. Diagram showing the entire upper limb muscle effort MA_{arm} as a function of the resistance to motion moments for different values of the assistance factor W. Effort while driving a conventional wheelchair is marked as standard in the diagram

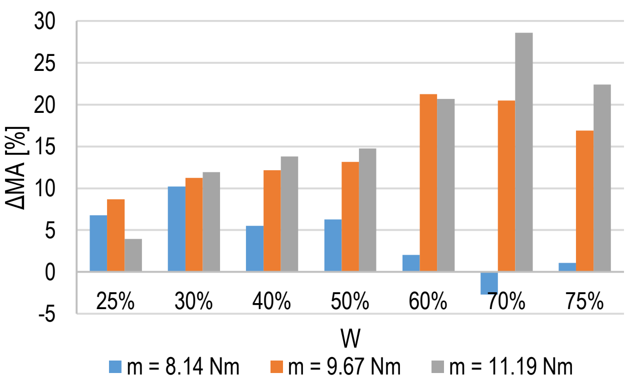

Fig. 6. Diagram of the percentage difference in muscle effort ΔMA depending on the value of the assistance factor W for the three values of the moment of resistance to motion m. MA, muscle activity

In analysing the effort of the upper limb, it was found that the value of the assistance factor *W* affects the muscular effort (Fig. 5). Based on the observed data, it was noticed that the greatest value of the assistance factor did not always translate into the greatest reduction in the effort of the upper limb MA_{arm}.

The above observation is confirmed by the analysis of the percentage difference in the entire hand muscle effort Δ*MA* depending on the assistance factor *W* setting (Fig. 6). Based on this analysis, it is clear that for each of the three tested moments of resistance *m*, there is a different value of the assistance factor *W* that is the most effective in reducing muscle effort. For the moment of resistance to motion *m* = 8.14 Nm, corresponding to driving on a paved smooth surface, it was found that the greatest reduction in muscle effort was obtained for the assistance factor *W* = 30%, whereas for the moment of resistance to motion *m* = 11.19 Nm, it was found that the greatest reduction in muscle effort was obtained for the assistance factor *W* = 70%.

4. DISCUSSION

Driving a wheelchair with a hybrid propulsion system requires the user to re-learn how to use the wheelchair and to feel its kinematics. According to Torkia et al. [34] in 2014, problems with the control of manually propelled wheelchairs are divided into four types and are related to moving in open space, manoeuvring in confined spaces, difficulties for novice users, and barriers and circumstances that are unpredictable and specific under certain conditions. The basic ones include manoeuvring in a confined space, driving through narrow doors, obstacles (thresholds, stairs, etc.), moving in crowds and weather difficulties (strong wind, rain, snow). The problems with using manual, electric and hybrid wheelchairs are similar. However, it is the wheelchairs with additional assisting propulsion systems that require the user to learn and feel the reaction of controls, also mentioned by other researchers [35].

In the research conducted, the limit value of the assisting torque is also noticed. It can be pointed out that too high a value of the assisting torque hinders the process of controlling the wheelchair, by increasing the muscular activity, which does not result mainly from driving, but from braking the wheels of the wheelchair in order to maintain the trajectory of movement. In the case of propulsion assistance solutions, the main task of the system is to provide additional drive torque to the wheels, as in bicycles equipped with such systems. However, in bicycles, it is easier to install and use such system [36] than in wheelchairs, since the propelling force in wheelchairs is transmitted to two wheels, and the direction of travel depends on their speed (or to be more precise, on the difference in their speed). Hence, the wheelchair user must have the ability to brake the individual wheels. Research has shown that too high a value of assistive propelling force may increase the muscle effort instead of reducing it. This phenomenon is also confirmed by the muscular activity of a different group of muscles than the group used during the pushing phase. An inaptly configured and operated hybrid manual–electric propulsion system will cause jerking of the upper limbs holding the push rims. This will have an adverse effect on the muscular system, as proved by the above analysis (Fig. 6). The phenomenon was precisely demonstrated when riding a wheelchair loaded with a small moment of resistance to motion (*m* = 8.14 Nm). In this particular case, after exceeding the value of the assistance factor *W*, a decrease in the reduction of muscle effort in the upper limb was observed. Moreover, for the highest values of the *W* factor, amounting to 70% and 75%, the high value hinders the wheelchair riding, increasing the effort of the entire upper limb.

The above results are confirmed by the analysis of the extensor carpi radialis longus muscle responsible for holding the hand on the push rims and stabilising the wrist while pushing. The analysis of the value of the muscle effort MA_4 (Fig. 7) has confirmed that the effective value of the *W* factor depends on the value of the moment of resistance that the user must overcome to move the wheelchair. It should be noted that as the moment of resistance decreases, the accuracy of the factor *W* calibration should be increased before using a wheelchair with a hybrid propulsion system.

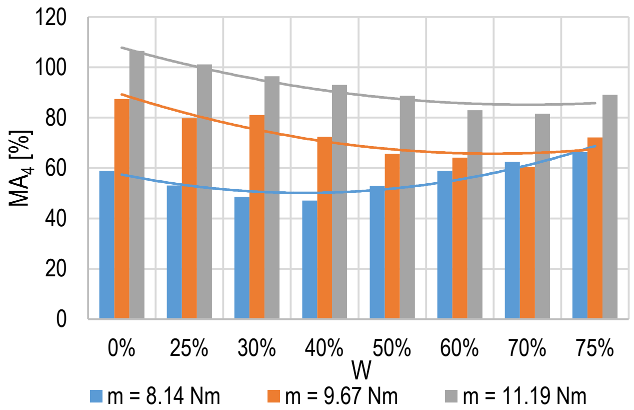

Fig. 7. Muscle Activity MA_4 diagram for the extensor carpi radialis longus musclea as a function of the assistance factor W for different values of the resistance torque m. MA, muscle activity

Since the performed analysis of a single extensor carpi radialis longus muscle coincides with the conclusions from the analysis of the muscle strength of the entire upper limb, it is possible to perform the electromyographic examination of this muscle as a non-invasive and individual calibration method of the propulsion system and the correct selection of the assistance factor *W* value depending on the area where the wheelchair is used. The values of the moments of resistance selected in the study have been estimated experimentally [35] and correspond to the resistance to motion occurring when riding a wheelchair on an even hard surface inside a building (*m* = 8.14 Nm), on pavements in an urban area contaminated with loose sand (*m* = 9.67 Nm) and approaching a paved hill with a slope of 4° (*m* = 11.1 Nm). The research fits into the modern trend of designing machines and devices using the analysis of EMG signals of the upper limb muscles, and the recorded values of this signal are similar to the results obtained by other researchers [37–39].

The conducted research has shown that the increase in the assistance factor *W* is not always reflected in the reduction of the muscle effort. In some cases, too high a value of the coefficient *W* will cause disturbances in the free movement of the arm driving the wheelchair. In the case of the tests carried out, this phenomenon was observed as the pulling movement of the grasping hand along with the hand rim in order to perform another driving phase. On the basis of the above observations, it was found that the value of the assistance coefficient *W* must be adjusted by the wheelchair user individually, depending on his physical abilities and the place of use of the wheelchair. Based on this observation, a design guideline was formulated. Namely, in the case of wheelchair propulsion systems with a manual–electric hybrid drive, it is imperative to enable the user ability to independently adjust the degree of assistance of the drive system by an electric motor.

5. CONCLUSION

Hybrid propulsion systems are currently one of the most important directions of wheelchair development. The selection of the drive torque value assisting the manual propelling of the wheelchair is a problem that includes multiple variables. As a part of the works performed, it was established that the most advantageous value of assistance factor for a man with a healthy musculoskeletal system above the waist up (refers to people with motor disabilities caused by an accident, lumbar section spinal cord injury [L1–L5] or sacral vertebrae section spinal cord injury [S1–S5] section) is within the range of 30 ± 10% when riding on a level and hard surface. When riding in conditions of increased load, for example, loose sand, the most advantageous value of assistance is 60 ± 10%, whereas when greater resistance occurs, for example, when climbing a ramp with a slope of 4°, the value is 70 ± 10%. The beneficial value of the assistance is considered to be the one that causes the least MA during movement. It has also been noticed that too high a value of the assisting propulsion torque causes problems with the control of the wheelchair movement and increases MA. This increased MA concerns a larger group of muscles in the upper limb than in the case of the conventional manual propulsion, which may indicate the need to put more force in the wheelchair riding control, which is carried out by pushing, but also braking the wheels. The presented results apply to people with disabilities who do not have problems with muscular dystrophy. In the future, the research can be extended to a larger group of users with different anthropometric dimensions (C5 and C95).

Considering the dependencies observed in the study, further research directions can be set out to automate the selection of the assistance factor W. In further works, the use of sensors determining the angle of the terrain inclination should be considered. Additionally, it would be advantageous to associate the continuous measurement of MA in the hand with the control of the value of the support factor.

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