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MANUAL WHEELCHAIRS PROPULSION – PART I: BIOMECHANIC ANALYSIS AND KINEMATIC MODELING

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Abstract: *Biomechanics are been applied in many areas of design. This paper focuses the application of biomechanics in the design of wheelchairs to prevent lesions and deformations in wheelchairs dependent individuals and to improve the efficiency of the propulsion characteristics of standard and race wheelchairs. Firstly, a review of the main factors that influence the wheelchair propulsion is showed. Then, a kinematic model of the subject is developed, as an "anthropomorphic robot", using the robotic methodology for manipulators modeling and motion analysis. The model obtained can be used in future works to analyse the kinematic motion of WDI's joints, force and moment application on hand-rims, to prevent lesions on musculoskeletal system due to long-term or incorrect use of wheelchair etc.*

Keywords: Biomechanics, Manual Wheelchair, Disabled People, Kinematic Model.

1. INTRODUCTION

The ability to manually propel a wheelchair is an important factor in independent functioning for a wheelchair user. Successful wheelchair propulsion relates to the user's ability, wheelchair design characteristics and wheelchair suitability for the user. The standard design of a wheelchair does not provide optimal propulsion efficiency, attractiveness and comfort for the user (Hughes *et al*., 1992). Due to these, for several years, biomechanics are been studied and applied to design wheelchairs, to prevent lesions and deformations in wheelchairs users and to improve the efficiency of the propulsion characteristics of standard and race wheelchairs. Also important is the application of the biomechanics on the study of user transferring tasks (Garg *et al.*, 1991 a-b and Owen *et al.*, 1991) to prevent and reduce back stress for nursing personnel by changing the physical job demands.

Today there are several researches investigating the relationship between man performance and wheelchair design factors. The behaviors of the cardiorespiratory and musculoskeletal systems (oxygen consumption, heart rate, joints ranges, kinetic movements of joints, work per cycle etc.) are used as control parameters by the biomechanics approaches. Unfortunately, standard wheelchairs have low mechanical efficiencies. Alternative methods of propulsion, as lever-drive systems, can provide greater mechanical efficiency than hand-rim propulsion, but this kind of mechanism has advantages like cost, weight, and complexity, besides the low attractiveness for the user (McLaurin & Brubaker, 1991).

There is a growing body of literature related to the biomechanics of wheelchair propulsion (Veeger *et al.*, 1992; Hughes *et al.*, 1992; Cooper, 1992; Bednarczyk & Sanderson, 1994; Dallmeijer *et al.*, 1994; Hofstad & Patterson, 1994; Rodgers *et al.*, 1994 Ruggles, *et al.*, 1994; Van der Linden *et al.*, 1996). Much of the wheelchair biomechanics literature is related to kinematic measurements of arm motions during wheelchair propulsion. Recently there are been more reports of kinetic measurements (Cooper *et al.*, 1996; Robertson *et al.*, 1996 and Cooper *et al.*, 1997). Unlike motion analysis systems, kinetic measurements require the use of custom push-rim force and moment measuring instruments, which are not currently commercially available (Cooper *et al*., 1997). The aims of this work are: firstly to make a review of the main factors those influence the wheelchair propulsion. Secondly, to model the wheelchair dependent individual (WDI), as an "anthropomorphic robot", as a sequence of one-dimensional rotations connected by rigid link segments, a method commonly used in robotics and motion analysis (Romilly *et al*., 1994 and Rosheim, 1997). These two parts will be attached in future works to develop the WDI dynamic model and then, it will be possible to estimate the forces and moments applied on hand-rims, study the kinematic motion of joints, avoid muscular lesions and improve the propulsion process and efficiency.

2. WHEELCHAIR PROPULSION BIOMECHANICS

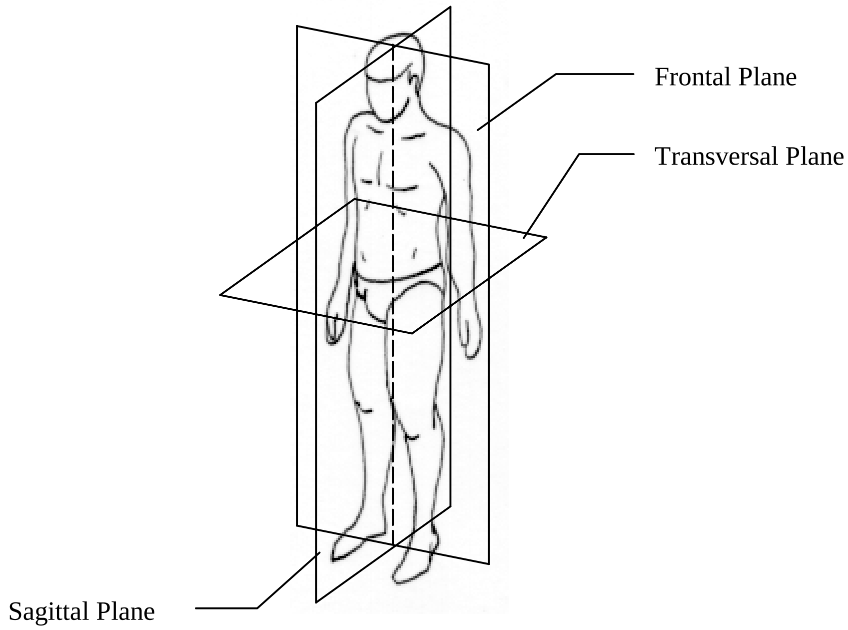


Figure 1 – Sagittal, Frontal and Transverse Planes on a human subject.

The wheelchair propulsion is a repetitive and cyclical movement that is the product of a user-machine interaction. Most user-machine systems, such as cycling or rowing, are often recreational and relatively short duration. The situation is entirely different for persons with disabilities who are dependent for their mobility on wheeled, human-powered machines. For children disabled in their lower extremities, the interaction with a machine is a lifelong experience (Bednarczyk & Sanderson, 1994).

There are several factors that influence the propulsion characteristics of manual wheelchairs, but the main factors are:

- The level of motor disability (Dallmeijer *et al.*, 1994);
- The user's position on seat (Hughes *et al.*, 1992 and McLaurin & Brubaker, 1991);
- The user's age (Bednarczyk & Sanderson., 1994);
- Fatigue (Rodgers *et al.*, 1994) and,
- Behavior and Design characteristics (McLaurin & Brubaker, 1991).

2.1 Level of motor disability

The influence of the level of motor disability, especially in WDI with spinal cord injury (SCI) can be observed in the medium power output and O_2 consumption. Of course, wheelchair athletes have better quantitative performances than sedentary wheelchair users, but for each group, subjects with a cervical lesion have a lower mean power output and O_2 consumption than subjects with thoracic or lumbar injury, and also, a complete different propulsion technique and kinematics characteristics. The relative low power output of cervical group indicates low physical performance capacity and higher risks for overload situations in daily life(Dallmeijer *et al.*, 1994).

Wheelchair propulsion is a complex form of arm work. The differences in the neuromuscular system is as a consequence of differences in lesion level and influence the O_2 consumption, aerobic and anaerobic power production and the characteristics of propulsion technique and kinematics. Different kinematic propulsion techniques are found inter- and intra-individually. The kinematic parameters that affect the wheelchair propulsion are attached to the hand rim grasp, contact and loose: strokes angle, begin angle, end angle, cycle time, push time and truck angle (Dallmeijer *et al.*, 1994). Despite the differences in output power between the group with cervical injury and the group with thoracic or lumbar injury, the kinematic parameters have no significant changes. The main change between these groups is in the kinematics of hand movements, due to the different limitations of motion on trunk, shoulder, elbow and wrist of each group (McLaurin & Brubaker, 1991).

2.2 Seat Position

The WDI position on seat is very important parameter for the biomechanic study of manual propulsion hand-rim wheelchair, due to the position on seat affects the motion of WDI's limbs and trunk during the propulsion and recovery phases. Besides the conventional position be the seat backrest in line with the hub axis, the ideal position for maximum efficiency is not necessary this because, this position depends on the dimension of the limbs and trunk. These values determine the joint points and range of muscle movements for the propulsion cycle (McLaurin & Brubaker, 1991).

Front seat positions, when compared to back seat positions, result in greater motion at the elbow and shoulder joints in the frontal and transverse planes. For front position, the propulsion phase occurs only on hand-rim frontal area and, for back position, on hand-rim top area. And, for back positions, the shoulder motion in sagittal plane and stroke arc are greater. Low seat positions result in significantly greater upper extremity motions and stroke arc than for high seat positions. Low positions require smaller forces and moments than high positions but, high positions produce high propulsion cycle frequency and less muscle energy consume on recovery phase (Hughes *et al.*, 1992 and McLaurin & Brubaker, 1991).

So, what is the ideal position? Besides the obvious answer be high-back position, as said before, the right position is a function of the WDI's dimension of limbs and trunk. So, only an anthropometric and physiological analysis can define the exact position.

2.3 Age

There are few published studies about the influence of WDI's age on kinematics of wheelchair propulsion. It is still not known whether children disabled from an early age develop movement patterns similar to those of individuals disabled in adulthood. It has been commonly assumed that children are miniature versions of adult wheelchair users in the area of seating. Bednarczyk & Sanderson (1994) on their previous work about kinematics of wheelchair propulsion, concluded that studies made in adult, athletic wheelchair users with paraplegia can be applied to pediatric group with the same disability, because they found propulsion style and kinematic characteristics very similar in both groups. Both groups responded in similar fashion in terms of wheeling velocities and spent comparable portions of the wheeling cycle in propulsion phase. Only the kinematics of upper limbs has little differences – the pediatric group showed more elbow extension than the adult group.

This suggest that, similar to other forms of human-powered movement such as bicycle riding, where the design features of two-wheeled cycle dictate pedaling style, there is only one way to propel a wheelchair. All WDI, independently of age, develop naturally to a similar, or common, wheeling style that is imposed by the nature of the wheelchair.

2.3 Fatigue

Injuries resulting from long-term or incorrect use of manual wheelchairs can impair the independence of WDI. Many musculoskeletal injuries (i.e., carpal tunnel syndrome, elbow/shoulder tendonitis etc.) appear to result from overuse and are related to the constant repetitive wrist, elbow and shoulder movements that occurs during wheelchair propulsion. Musculoskeletal problems can also arise from misalignment of the limbs that may occur with fatigue or inappropriate wheelchair use, design and/or prescription. Such injuries can be a detriment to manual WDI and can hinder rehabilitation efforts (Rodgers *et al.*, 1994).

Although the WDI temporal parameters are similar in nonfatigued and fatigued states, the patterns of muscle activity have some qualitative differences: the muscles are active for a slightly larger portion of the pushing cycle (propulsion phase).

The propulsion biomechanics with fatigue has significant changes in trunk forward lean, wrist radial/ulnar deviation, peak hand-rim force, and among upper extremity joints. This biomechanic behavior becomes the shoulder the upper extremity region most prone to overuse injury.

2.4 Behavior and design characteristics

The design and construction of the wheelchair and its component parts can have a marked effect on the performance, energy requirements and durability under various ambient conditions and use patterns. The components include wheels, tires, castors, bearings, materials and seats. Each component must be considered in relation to performance characteristics including rolling resistance, versatility, weight, comfort, stability, maneuverability, durability and maintenance (Becker & Dedini, 1997).

There are four factors, which govern the work, required to propel a vehicle: the surface over which it is rolling, the slope, wind and the rolling resistance of the vehicle. Only the latter is a function of the vehicle design, but the design can have an effect on performance with respect to the three environmental factors. For example, some tires may be suitable for hard pavement but not for grass. Tires are the single most important factor in determining rolling resistance on level terrain.

There are differences in the rolling resistance of different types of tires. For example, a high-pressure pneumatic tire required only one quarter of the pulling force of the solid gray rubber tires on smooth firm surface. The wheel alignment influences also in the pulling force. Camber angle up to 10° (tilting the top of the wheels inward) has no significant effect on rolling resistance. Toe-in or toe-out, however, resulted in a serious increase in the pulling force. Only one or two degrees misalignment could double the required force (McLaurin & Brubaker, 1991).

Studies regarding the rolling resistance of tires on grass or other off-pavement surfaces are difficult to perform since there is no practical way to characterize or simulate such surfaces. On soft ground or sand, it can be assumed that wide tires will roll more easily than narrow tires. The diameter of the tires also has a significant effect. As a general rule, the rolling resistance is inversely proportional to the diameter.

Although pneumatic tires are preferable to solid rubber from a standpoint of rolling resistance, comfort and weight, research has shown that this may change soon. Synthetic tires are superior in wear resistance and not subject to flats from slow leakage or punctures. They can be designed to be much more durable, cheaper, lighter and with a rolling resistance comparable to pneumatics, and synthetic tires do not provide as smooth a ride, springs may more than compensate for this deficit. Tires or springs, which absorb shock also, decrease the stress on the frame, axles and wheels.

Although the wheelchairs use castor because they allow motion in any direction, outdoor lever drive and racing wheelchairs have steerable wheels. This occurs because castors have some problems that can become the vehicle unstable during maneuvers (Becker & Dedini, 1997).

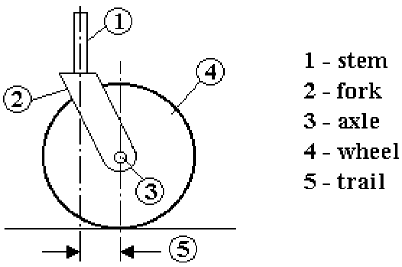


Figure 2 - Diagram of a Castor with a vertical stem.

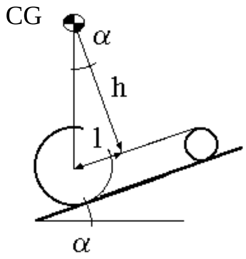
The castor stem is one of the most critical parts of the castor wheel vehicle. If the stem is not vertical, but is tipped to the left, for example, then the vehicle will turn to the left when coasting. This is the primary reason for poor tracking characteristics of this kind of vehicle. Also if the stem is tipped forward at the top, the effective trail is reduced. The trail is the distance from the ground contact of the tire to the spot where the axis of the stem would intersect the ground. With a vertical stem, the dimension is the distance of the axle behind the stem (Fig. 2 - no. 5). The trail is an important parameter. A long trail makes turning easier but causes the castor wheel to sweep through a greater arc. A long trail also means that castor flutter is less likely to occur. Castor flutter or shimmy is not only annoying and energy consuming, but can be very dangerous. The rolling resistance of a castor can multiply ten times or more when fluttering. Thus, when coasting down a gradient, the onset of flutter acts like a brake which can, and often does, cause the occupant to be thrown forward out of the vehicle (McLaurin and Brubaker, 1991).

Roll and lateral stability of wheelchair can be not important at low speeds in plane surfaces, but when the wheelchair is traversing a ramp or a side slope they will be very important to avoid a rollover. Ramps or inclines are commonly used to provide opportunities for WDI and to overcome differences between grade levels. But yet the limits of allowable grades have not been based upon stated scientific criteria. This reflected in the widely differing standards among various countries (maximum limit - France and Belgium: 5%, Poland: 12.5% and Brazil: 5 to 12.5%, see Tab. 1) (Cappozzo *et al*., 1991). The Brazilian Standard NBR 9050/1994 (ABNT, 1994) indicates that the ramps can have different inclination limits:

Table 1 – Design of Ramps (ABNT, 1994)

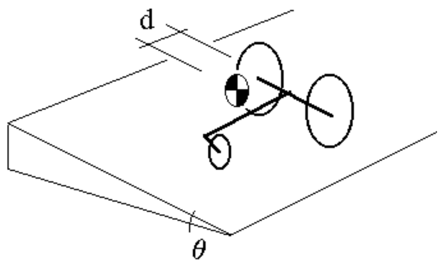
Ramp Grade (α) [%]	**Maximum Level Grade [m]**	**Level Maximum Number**	**Ramp Maximum Length [m]**
5.00	1.500	-	30.00
6.25	1.000 / 1.200	14 / 12	16.00 / 19.20
8.33	0.900	10	10.80
10.0	0.274 / 0.500 / 0.750	8 / 6 / 4	2.74 / 5.00 / 7.50
12.5	0.183	1	1.46

Due to this, the GC height must be as low as it is possible to avoid the rollover in a ramp.



Where:
α - ramp angle
h – CG heigh
l – CG width
and,

$$\alpha = \tan^{-1}\left(\frac{l}{h}\right)$$



(a) (b)

Figure 3 - Wheelchair diagrams: (a) in a ramp and (b) in a side slope.

So, as the GC height grows less, the ramp critical angle grows up and the wheelchair becomes more stable as for rollover stability problems. Besides, in a side slope, a front castored vehicle will tend to turn downhill and rear castored vehicle tends to turn uphill. This tendency to turn on a side slope depends upon the distance of the GC in front of or behind the axis of the main wheels (non-castors) and the angle θ of the slope (Fig. 3 - b).

3. KINEMATIC MODEL

Using the same method commonly applied in robotics and motion analysis, the WDI is modeled as a sequence of one-dimensional rotations connected by rigid link segments. The first documented using this concept of linking human kinesiology with anatomy is dated 500 years ago. Between 1495 and 1497, Leonardo da Vinci designed and possible built the first articulated anthropomorphic robot in the history of western civilization. This “anthrobot” is a beacon for the contemporary designer. His powerful concept is today applied in the mechanical design of machines that approximate human capabilities like: upper and lower limb orthosis, robots for the service industry, undersea and nuclear manipulation, space station maintenance etc. (Romilly *et al*., 1994 and Rosheim, 1997).

The human shoulder is a complex, sophisticated and interrelated system that can not be model as a simple spherical joint. The rib cage and scapula are basically ball-and-socket joints. Integral to scapula is a socket that receives the head of the humerus, thus creating a second ball-and-socket joint. The clavicle is integrated to scapula and the spine is modeled as a single rotational joint to allow the trunk forward lean.

This model does not consider the hand degrees of freedom. So, the spine has one degree of freedom, the shoulder five, the elbow two, and the wrist two. Figure 4 shows the kinematic model of the WDI.

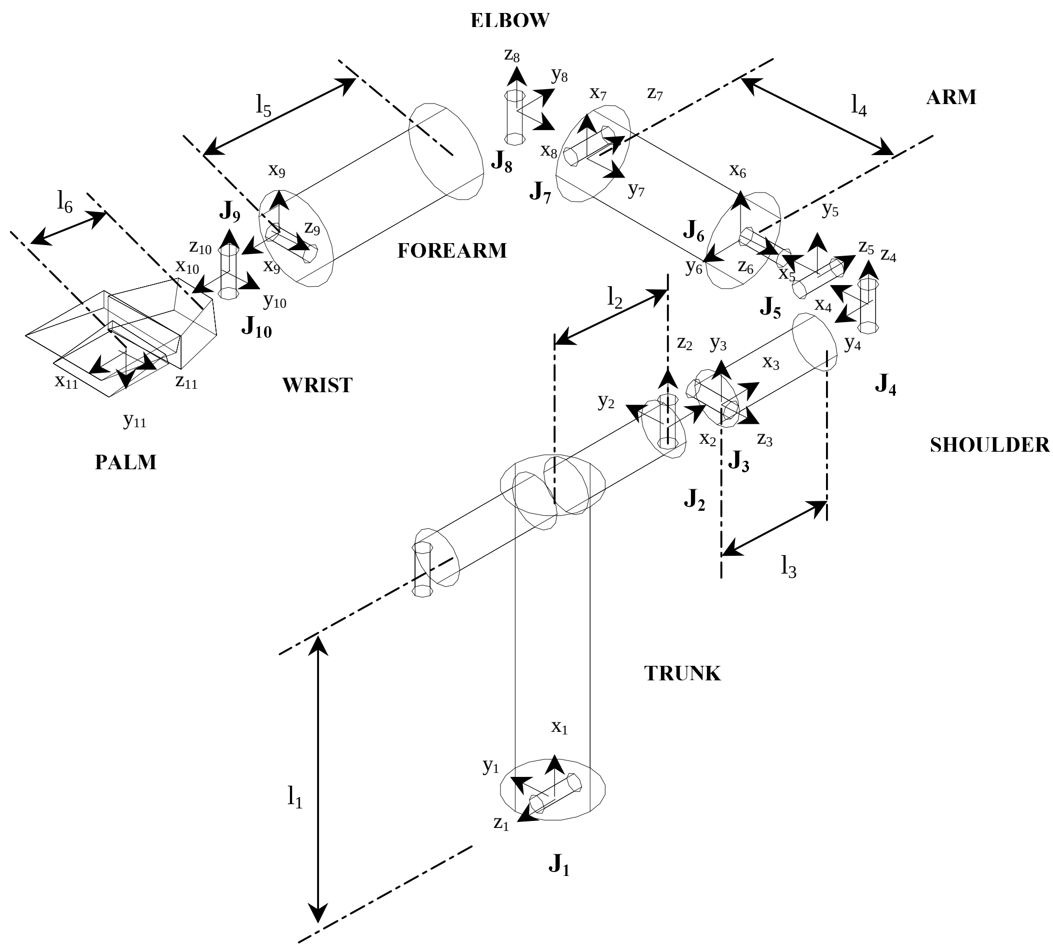


Figure 4 – Kinematic formulation axis definitions.

Where:

J_1: Trunk lean	J_6: Shoulder roll
J_2: Shoulder yaw	J_7: Forearm rotation
J_3: Shoulder pitch	J_8: Elbow flexion
J_4: Shoulder azimuth	J_9: Wrist flexion
J_5: Shoulder elevation	J_{10}: Wrist yaw

The geometrical model expresses the position and orientation of the WDI's hand with respect to a coordinate system jointly to the WDI's buttocks, in function of its generalized coordinates (angular coordinates in the case of rotational joints). The geometrical model is represented by the following expression (Sá & Rosário, 1998):

$$\underline{x} = f(\,\underline{\theta}\,) \tag{1}$$

Where:

$\underline{\theta} = (\theta_1,\ \theta_2,\ \ldots\ldots,\ \theta_n)$:angular position vectors for the joints, and

$\underline{x} = (X,\ Y,\ Z,\ \psi,\ \theta,\ \phi)$: position vector, where the three first terms denote the Cartesian position and the three last terms stand for the orientation of the WDI's hand.

This relation may be expressed mathematically by a matrix that relates the system of coordinates jointly to the WDI's buttocks with a system of coordinates associated to his hand. This matrix is called homogeneous passage matrix and is obtained from the product of the homogeneous transformations matrix, $A_{i,i-1}$, that relates the system of coordinates of an element *i* with the system of the previous element *i*-1, that is:

$$T_n = A_{0,1} * A_{1,2} * \ldots\ldots * A_{n-1,n} \tag{2}$$

$$T_n = [\ \underline{n}\ \underline{s}\ \underline{a}\ \underline{p}\ ] \tag{3}$$

Where:

$\underline{p} = [\ p_x\ ,\ p_y\ ,\ p_z\ ]$: position vector, and

$\underline{n} = [\ n_x\ n_y\ n_z\ ],\ \underline{s} = [\ s_x\ s_y\ s_z\ ]$ e $\underline{a} = [\ a_x\ a_y\ a_z\ ]$: Orthonormal vector that describes the orientation.

The development of a numerical algorithm to find the angular positions of WDI's upper-limb joints, contains the solution of the inverse kinematic problem through the usage of a recursive numerical method that uses the calculation of the kinematic model and of the Jacobian inverse matrix for the WDI. The need for finding references in angular coordinates referring to the trajectories defined in the Cartesian space is expressed mathematically by the inversion of the geometrical model, that is:

$$\underline{\theta} = f^{1}\ (\,\underline{x}\,) \tag{4}$$

Through the function *f* it is possible to calculate the movement of the WDI's hand resulting from the movement of the upper-limb joints. This function is non-linear and has no non-trivial analytical solution. The generation of trajectories through the usage of the kinematic inverse model presented excellent results and the computational simplicity of the method (Miss) allows the implementation of real time trajectory generation. This allows the use of the numerical algorithm on upper-limb orthosis control.

3. CONCLUSIONS AND FUTURE WORKS

A review of the main factors that influence the wheelchair propulsion was showed. The biomechanics of the wheelchair propulsion is very complex due to the differences on the WDI's motor disability level, position on seat, age, fatigue or non-fatigue and the design of the wheelchair. To simulate and study the kinematic motions and dynamic efforts of WDI's upper-limb joints, a geometrical model of WDI and wheelchair was developed. A modulate program in language Ada is been developed to generate the trajectories of WDI's hands during the propulsion cycle of the wheelchair. The next step is to add to the program the dynamic of the system and then, for a set of parameters (joint angles, forces and momentum limits) obtain the efforts on joints.

REFERENCES

ABNT, 1994, NBR9050/1994 - Accessibility of the handicapped to buildings and the urban environment, Rio de Janeiro, Brazil.

Becker, M. & Dedini, F. G., 1997, Design and simulation of AVDs (autonomous vehicles for disabled) – preliminary study, SAE Technical Paper 973085 E, p. 10.

Bednarczyk, J. H. & Sanderson, D. J., 1994, Kinematics of wheelchair propulsion in adults and children with spinal cord injury, Arch. Phys. Med. Rehabil., vol. 75, pp. 1327– 1334.

Cappozzo, A. *et al.*, 1991, Prediction of ramp traversability for wheelchair dependent individuals, Paraplegia, vol. 29, pp. 470 – 478.

Cooper, R. A., 1992, The contribution of selected anthropometric and physiological variables to 10K performance of wheelchair races: a preliminary study, Journal of Rehabilitation Research and Development, vol. 29, no. 3, pp. 29 – 34.

Cooper, R. A. *et al.*, 1996, Projection of the point of force application onto a palmar plane of the hand during wheelchair propulsion, IEEE Transactions on Rehabilitation Engineering, vol. 4, no. 3, pp. 133 – 142.

Cooper, R. A. *et al.*, 1997, Uncertainty analysis for wheelchair propulsion dynamics, IEEE Transactions on Rehabilitation Engineering, vol. 5, no. 2, pp. 130 – 139.

Dallmeijer, A. J. *et al.*, 1994, Anaerobic power output and propulsion technique in spinal cord injured subjects during wheelchair ergometry, Journal of Rehabilitation Research and Development, vol. 31, no. 2, pp. 120 – 128.

Garg, A. *et al.*, 1991 (a), A biomechanical and ergonomic evaluation of patient transferring tasks: bed to wheelchair and wheelchair to bed, Ergonomics, vol. 34, no. 3, pp. 289–312.

Garg, A. *et al.*, 1991 (b), A biomechanical and ergonomic evaluation of patient transferring tasks: wheelchair to shower chair and shower chair to wheelchair, Ergonomics, vol. 34, no. 4, pp. 407 – 419.

Hofstad, M. & Patterson, P. E., 1994, Modeling the propulsion characteristics of standard wheelchair, Journal of Rehabilitation Research and Development, vol. 31, no. 2, pp. 129 – 137.

Hughes, C. J. *et al.*, 1992, Biomechanics of wheelchair propulsion as a function of seat position and user to chair interface, Arch. Phys. Med. Rehabil., vol. 73, pp. 263– 269.

McLaurin, C. A. & Brubaker, C. E., 1991, Biomechanics and the wheelchair, Prosthetics and Orthotics International, vol. 15, pp. 24 – 37.

Owen, B. D. *et al.*, 1991, Reducing risk for back pain in nursing personnel, AAOHN Journal, vol. 39, no. 1, pp. 24 – 33.

Robertson, R. N. *et al.*, 1996, Push-rim forces and joint kinetics during wheelchair propulsion, Arch. Phys. Med. Rehabil., vol. 77, pp. 856 – 864.

Romilly, D. P. et al.,1994, A functional task analysis and motion simulation for the development of a powered upper-limb orthosis, IEEE Transaction on Rehabilitation Engineering, vol. 2, no. 3, pp. 119 – 129.

Rodgers, M. M. *et al.*, 1994, Biomechanics of wheelchair propulsion during fatigue, Arch. Phys. Med. Rehabil., vol. 75, pp. 85 – 93.

Rosheim, M. E., 1997, In the footsteps of Leonardo, IEEE Robotics & Automation Magazine, June 1997, pp. 12 – 14.

Ruggles, D. L. *et al.*, 1994, Biomechanics of wheelchair propulsion by able-bodied subjects, Arch. Phys. Med. Rehabil., vol. 75, pp. 540 – 544.

Sá, C. E. A. & Rosário, J. M., 1998, Algoritmo Numérico para a Solução do Problema Cinemático Inverso de Manipuladores (in Portuguese), Proceedings of V CEM – NNE, October 1998, Fortaleza – CE, Brazil, vol. 2, pp. 32 - 39.

Van der Linden, M. *et al.*, 1996, The effect of wheelchair hand-rim tube diameter on propulsion efficiency and force application (tube diameter and efficiency in wheelchairs), IEEE Transactions on Rehabilitation Engineering, vol. 4, no. 3, pp. 123 – 132.

Veeger, H. E. J. *et al.*, 1992, A computerized wheelchair ergometer, Scand. J. Rehab. Med., vol. 24, pp. 17 – 23.