BIOENGINEERING METHODS OF WHEELCHAIR EVALUATION

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The Veterans Administration undertakes the responsibility for evaluation of a variety of orthopedic devices such as braces, artificial limbs, wheelchairs, and patient lifts. Its primary purpose is to provide an intelligent basis for selection, procurement, and application of those devices which can be of greatest benefit to disabled veterans. This program also serves to guide the manufacturers of orthopedic devices along lines of development that are of particular interest to the Veterans Administration. The evaluation may therefore range from analysis of the design and materials to a full-scale biomechanical analysis in the laboratory, as well as a field study in several hospitals and clinics around the country. Laboratory studies are generally delegated to the VAPC Bioengineering Laboratory, and field studies are usually conducted by the Prosthetic and Sensory Aids Service.

Orthopedic braces and artificial limbs for both the upper and lower extremities have been the subject of evaluation programs for many years, and relatively useful methods and techniques have been evolved. The parameters which provide the most useful assessment of these devices are fairly well established. In addition, there exists a body of basic data on normal human locomotion from which useful criteria for evaluating performance can be drawn.

A somewhat different situation prevails in an approach to the evaluation of such items as lift aids and wheelchairs since there is available little basic information about their performance factors. Such standards as do exist are descriptive in nature and relate primarily to dimensions and materials of the devices. A meaningful evaluation, however, depends on tests not only of hardware but also of the human factors that enter into efficient use.

To provide a basis for an adequate assessment of orthopedic lift aids and wheelchairs, there has evolved in the Bioengineering Laboratory an evaluation program that takes into account the man-machine combination and that is both descriptive and functional. This program consists of specifically designed test procedures to provide information on:

1. Analysis of mechanical design, adequacy of materials, and durability.
2. Convenience and ease of operation.
3. Patient acceptability in relation to appearance, utilization in the home, and the availability of other similar devices.
4. Stability and safety.
5. Force and energy requirements.

A careful analysis of the design of an orthopedic device indicates the extent to which the fundamental idea or purpose has been translated into appropriate mechanical features. A consideration of the mechanical design in relation to all aspects of the intended application frequently discloses serious limitations. In designing an orthopedic lift for extreme stability, for example, one developer failed to realize that the broad base of support was a grave handicap in confined areas where lifts are frequently used. Materials used in an apparatus strongly influence comfort, safety, and durability. These features are assessed in the light of good design principles and on the basis of some standards which, although not entirely adequate, can be usefully applied. Durability is frequently determined by means of cycling tests.

Convenience and ease of operation are not only important considerations for the therapist who may use the devices, but they are especially critical factors for the patients of limited strength and mobility who must operate them. Use tests are devised to assess these matters. In some cases, observation of several appropriately selected patients may suffice; in other cases, longer-term hospital use may provide the basis for judgment.

Patient acceptability is essential; even the most meticulously designed aid is useless if patients resist it; therefore, the reactions and opinions of patients are obtained to form the basis of judging acceptability. The survey method may consist of a few direct and specific questions in one case, a formal questionnaire answered by several patient users in another, or complete batteries of questionnaires designed for patients, families, and clinical personnel in a third.

Stability and safety are obviously of paramount importance in devices designed to aid the handicapped. In most cases, the patient's balance is impared, and he depends completely on the device for stability. Maintaining a condition of stability depends on the relationship between the vertical projection of the center of gravity (CG) and

the area of the base of support. This relationship is easily tested in a relatively static situation, such as a lift aid, where the CG of the total system does not move about very much. Observation of the patient as he is lifted provides a reasonably adequate empirical evaluation. In the dynamic situations encountered with such aids as wheelchairs, the CG projection of the total system must be determined for changing conditions of slope, speed, and body position. This provides a basis for determining the conditions under which the CG projection falls too close to the edge of the support base. From these data, safety limits can then be defined.

The amount of energy required of the patient to operate a device should be a major consideration in any evaluation of a man-machine combination. Although not extensively explored in the orthopedic field, the significance of energy costs and efficiency of operation is known to increase with the severity of the disability.

With lifting devices, where the patient's role is passive, measurements of the force necessary to operate cranks or switches are sufficient. Devices that provide locomotion (including artificial limbs), however, require energy inputs from the patients. The distribution of this energy between purposeful motion and friction is a key factor in the efficiency of the device. Simple force measures are readily obtained by means of tensiometers or dead-weight comparison techniques. Relating the physiological energy costs to the mechanical energy outputs, however, is not as easily accomplished. Difficulties are encountered with regard to the validity of basic assumptions, measurement methods, and modes of expression. At the present time, only classical methods from the field of work physiology are feasible for calculating metabolic energy costs; these methods are based on the determination of oxygen utilization rates. The most appropriate methods for measuring mechanical energy outputs, taken from the field of physics, are based on mass and velocity relationship, e.g., $E=\frac{1}{2}mv^2$. For evaluation purposes, both methods should perhaps be related in a more meaningful expression than either one alone, i.e., in terms of efficiency.

An early example of the kind of energy-cost evaluation program carried out by the VAPC Bioengineering Laboratory may possibly serve as a guide for setting up such studies elsewhere. The device in question, a lightweight wheelchair, was conventional in most outward respects except for the decreased weight resulting from the use of lighter materials. The basic purpose, therefore, was to determine whether this *prima facie* advantage actually accrued to the user and, if so, to describe the nature of any benefits to be expected.

The manufacturer of the light chair also produced a line of conventional chairs, so the evaluation program was designed to compare the experimental device with its conventional counterpart. The fol-

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lowing report appears as originally written except for a few minor editorial changes such as the deletion of commercial names.

The evaluation program to assess the features of the lightweight wheelchair was designed to include mechanical and biomechanical analyses as well as an objective consideration of subject reactions.

METHODS

1. Mechanical Analysis

The design and materials used in this chair depart from conventional specifications. Current specifications (MIL–C–15861 Bu Med dated 15 November 1950) do not appear to be wholly applicable, because innovation in the farbication of wheelchairs has introduced new materials and configurations which are at present unevaluated. Nevertheless, the lightweight chair was examined with respect to those criteria which seemed applicable, e.g., certain design features, materials, and dimensions.

2. Biomechanical Analysis

Although the weight differential was the primary distinction between the light aluminum and conventional chairs, other factors such as wheel diameter and handgrip were also significantly different features. The light chair and standard chairs were, therefore, evaluated with respect to (a) performance, (b) effort, (c) inertial forces, and (d) stability.

a. Performance. This factor was based upon a comparison of average wheelchair velocities, maintenance of uniform velocity in terms of time required to make one circuit of the track, and upon the average distance traveled by each chair per stroke.

b. Effort. The rating of this factor was based upon a determination of (1) the cardiac response to wheelchair propulsion as indicated by pulse and blood pressure changes above the resting rate, and (2) metabolic and energy costs as indicated by an analysis of the plumonary ventilation and calorie cost requirements of the light and conventional chairs respectively.

Data on performance and effort were collected in a series of tests in which each subject propelled himself around a level rectangular track (172 feet long) 10 times. The light chair and the conventional chair were used in alternate tests. Subject instructions were minimal; the subject was told only to proceed at his most comfortable speed. To avoid serial order effects, the sequence of chair use was reversed for each pair of runs.

Standard open-circuit techniques were used in monitoring pre-test, during-test and post-test ventilation volumes and in obtaining aliquot

samples of expired air for determination of oxygen, nitrogen, and carbon dioxide concentrations. Heart rate and blood pressure were recorded before and after each test.

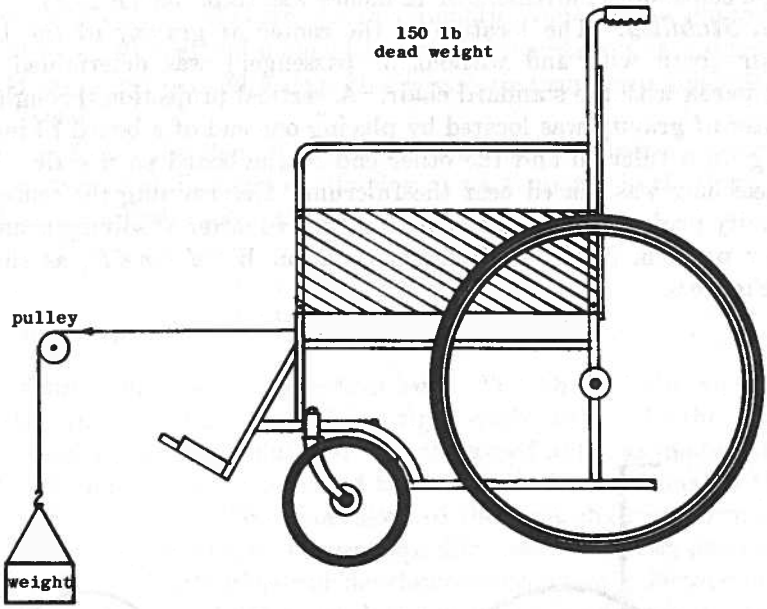


FIGURE 1. *Method of determining inertial force of wheel chair.*

A review of the literature revealed very little published material on the energy cost of propelling wheelchairs. Gordon (1) reported that wheelchair locomotion at a rate of 1.2 m.p.h. (approximately 106 ft. per minute) required an energy expenditure of from 134 to 138 percent over the basal level. In another publication by the same author (2), an energy cost of 2.4 Calories[1] per minute for the same task was reported. Because these findings were based on a study of two normal subjects and one pulmonary patient, the data were not representative of typical wheelchair users. Furthermore, Gordon's report (2) did not indicate whether the value of 2.4 Calories per minute was a gross or net figure, or whether it were expressed in terms of body weight or skin surface area. The available literature was thus of little use for comparison in our evaluation; however, data on the energy cost of several modes of locomotion appear in Appendix A.

c. Inertial Forces. The average starting force for each wheelchair was recorded in order to obtain a measure of "stop-go" performance, which is more typical of normal wheelchair activity than long uninterrupted periods of propulsion. Measurements of the forces

[1] The great calorie (1,000 calories) is designated as Calorie throughout this paper.

required to set the chair in motion under a standard load were made by pulling the chair with a light cable, which passed over a pulley and was connected to a weight pan. The weight (in ounces) producing a continuous movement of 12 inches was recorded (Fig. 1).

d. Stability. The location of the center of gravity of the light chair (both with and without its passenger) was determined and compared with the standard chair. A vertical projection through the center of gravity was located by placing one end of a board 76 inches long on a fulcrum and the other end of the board on a scale. The wheelchair was placed near the fulcrum. Determining the center of gravity projection for each chair was then a matter of solving a simple lever problem in the equilibrium equation $W \times d = w \times D$, as shown in Figure 2.

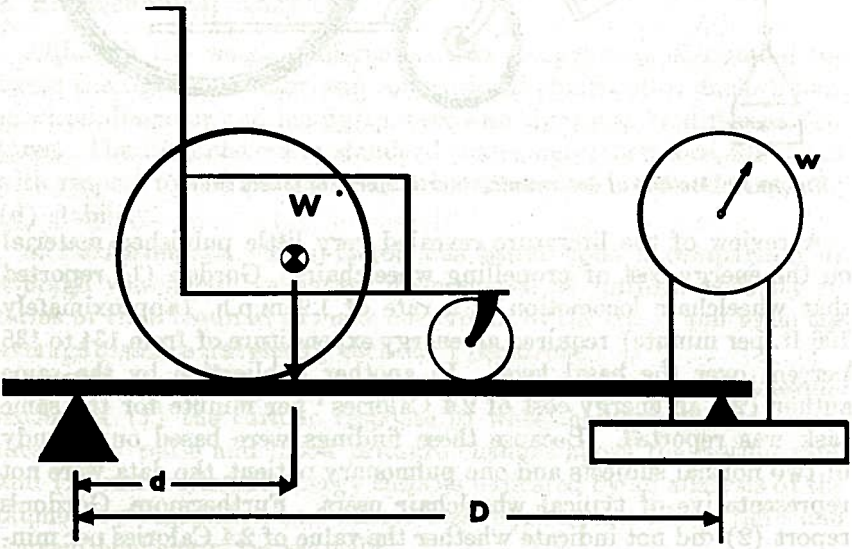


FIGURE 2. *Method of determining chair center of gravity projection. (W= weight of passenger and chair; d= distance from fulcrum to CG projection; D=effective length of platform; w=scale reading in lb.*

3. Subject Reactions

Three subjects were asked to express their opinions of the light chair after having performed several tests with it. In addition, two subjects were given the chair to use at home and at work for periods of four days each, and their experiences in daily use were recorded.

4. Subjects

Three volunteer paraplegic subjects, all of whom used wheelchairs routinely, participated in this study.

JG was a 32-year-old male with a traumatic cauda equina syndrome.

HB was a 50-year-old male with traumatic cauda equina syndrome.

RM was a 28-year-old male with a traumatic transverse myelopathy at T 4 level.

None of the subjects had any involvement of the upper extremities. A fourth veteran (partial quadriplegic) participated in the daily activities use test with subject JG.

FINDINGS

1. Mechanical

a. Compliance with Specifications. The obvious discrepancies between the specifications and the light chair were: (1) the use of aluminum as the frame material; (2) the use of nylon as the back and seat material; and (3) the use of two "drive" wheels of smaller than specified diameters. The dimensions of the light chair conformed to the specified dimensions; current specifications, however, should be reviewed in the light of recent developments since such factors as the use of aluminum and nylon, while not specifically proscribed, are not mentioned.

Construction was mainly of aluminum and chrome-plated lightweight steel tubing with seat and back rest fabricated of nylon.

Each hand rim and wheel, made of extruded aluminum, was a single, integral unit.

The closeness of hand rim to wheel tended to dirty the palmar surface of the hand in the thenar area.

Surface marring of the aluminum hand rim became evident after a short period of use during this evaluation.

The backrest was 1 in. higher than on most other models.

The "latch"-type brake on each wheel provided more positive locking than older designs.

The nylon fabric seat and backrest were light, washable, and strong.

With a load of 300 lb. placed in the chair for a period of 1 hour, no appreciable sag developed.

b. Dimensions. Overall width of lightweight chair when fully open, 26 in.; overall width of chair when folded, 11 in.; seat width at seat level, 18 in.; width between front uprights, 16 in.; arm height from floor (measured on forward section), 28 in.; arm height from seat, 9 in.; seat height from floor (measured on forward section), 19½

in.; overall height, 36 in.; overall length, 40 in.; seat upholstery depth, 16 in.; back upholstery height, 17½ in.

The lightweight wheelchair weighed 29 lb., which was 10 to 20 lb. lighter than most other adult-size wheelchairs.

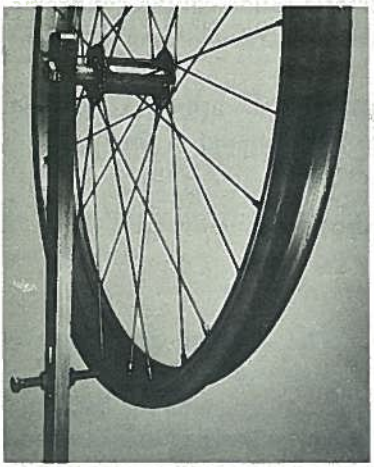


FIGURE 3. *Setup to measure changes in wheel characteristics during 100-cycle destructive testing program.*

c. Destructive Testing. The use of lighter materials usually implies a corresponding reduction in strength; however, a substantial weight reduction effected without material loss in strength is an obvious advantage. Experience indicated that in descending curbs, peak stress loads were applied to the major components (wheels, axles, and frames) of a wheelchair. A test was devised to simulate this condition, i.e., the chair was weighted with 200 lb. of shot bags and rolled off a 6-in. wooden platform (curb) repeatedly.

The test was designed to measure changes in wheel alignment, concentricity of axles, and radial clearance of bearings after 1, 5, 15, 25, 50, and 100 cycles during a 100-cycle program (Fig. 3).

Each cycle started with the weighted chair resting on the platform; the chair was manually propelled backward at a slow, uniform rate until both drive wheels rolled off the platform and rested on the floor; care was taken to avoid the application of extraneous external forces (Fig. 4).

The test was discontinued after 38 cycles, when both crosspieces of the wheelchair frame buckled. The failure in one member was so extensive that testing was discontinued (Fig. 5).

As shown in Table 1, no significant changes in concentricity of the axles or in the radial clearance of the bearings occurred during the 38 cycles. After 5 cycles, however, small deviations appeared in wheel alignment; a major alignment change occurred at 38 cycles when the crosspiece failed, and the test ended.

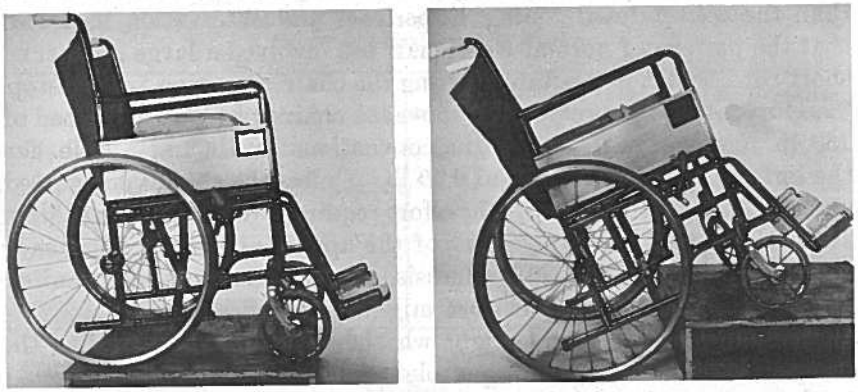


FIGURE 4. *Test devised to simulate curb descent with 200-lb. weight in wheelchair. (a) Chair on 6-inch platform at start of test; (b) Driving wheels on floor at end of test.*

2. Biomechanical

The biomechanical procedures identified several potential benefits to be derived from the use of the lightweight wheelchair (section *g*, items A, B, and C). Also identified were certain disadvantages (section *d*, *g*) as well as areas with no significant differences between the chairs (section *e*, *f*).

a. Cardiac Response. Use of the lightweight chair produced less cardiac strain than the conventional chair. Two measures of cardiovascular reaction to the stresses were employed: (1) excess pulse rate (the difference in pulse rate measured one minute before exercise and one minute after exercise), and (2) excess blood pressure similarly measured. Without exception, pulse rate after using the conventional chair was substantially higher than after using the lightweight chair, as indicated by the absolute differences (Table 2). Systolic pressure after using the conventional chair was also significantly higher in two subjects and remained the same in the third case. In relative terms, however, the differences in cardiovascular strain, expressed as multiples of the pre-exercise level, although still in favor of the light chair, were not quite so substantial (Table 3).

b. Pulmonary Ventilation. Use of the light chair produced lower pulmonary ventilation rates. Respiration rate and depth responded very quickly to changes in activity level. As an indication of relative stress, the total volume of expired air was expressed per unit of body weight (Table 4).

Ventilation volumes per minute were lower in all cases when the light chair was used, with the net difference ranging from 7 to 12 percent.

c. Starting Forces. The light chair required less force to start than the conventional chair. Experience and observation indicated that the pattern of normal wheelchair use involved a large number of short runs, which necessitated setting the chair in motion from a stop. The forces that were required to move the chairs with a vertical load of 150 lb. averaged 3.45 lb. for the conventional chair and 3.05 lb. for the lightweight, a difference of 0.40 lb. Unless the chair were started several hundred times a day, the effort required would seem negligible for patients with no involvement of the upper extremity. For quadriplegics and polio, multiple sclerosis, or stroke patients with limited strength or capacity, this feature might be significant.

d. Stability. The lightweight wheelchair was less stable than the conventional chair in resisting stroke reaction forces; when subjected to the same stroke and force pattern applied to conventional chairs, the front wheels of the lightweight chair lifted off the ground. The stability of a structure under static conditions depends primarily upon the relationship between the center of gravity and the base of support; consequently, as long as the vertical projection of the center of gravity falls within the base of support, the structure will not topple, i.e., it will remain stable (Fig. 6).

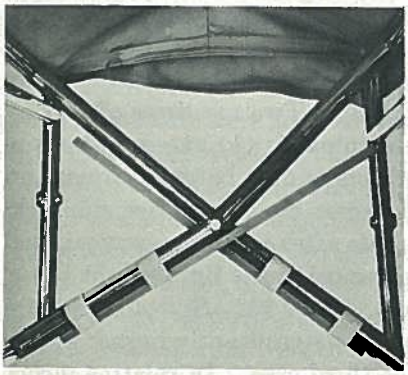


FIGURE 5. *Results of destructive test after 38 cycles: a broken crosspiece.*

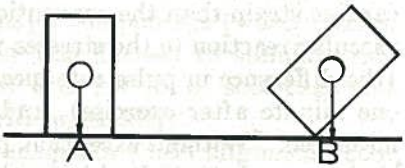


FIGURE 6. *Relationship between center of gravity of an object and base of support. (A) With vertical projection of CG falling within the base of support, the object remains stable; (B) With vertical projection outside base of support, the object topples.*

TABLE 1

Results of Simulated Curb Descent Test
Crosspiece Failure at 38 Cycles

No. of cycles	Maximum deviation of wheel (in.)	Concentricity of axles (in.)	Radial clearance of bearings (in.)
0	1/64	0. 001	0. 002
1	1/64	. 001	. 002
5	1/32	. 001	. 002
15	1/32	. 001	. 002
25	1/32	. 001	. 002
38	3/64	. 001	. 002

TABLE 2

Pulse Changes per Minute before and after Exercise with Conventional and Lightweight Wheelchairs

Subject	Conventional		Experimental	
	Net difference	Percent of increase	Net difference	Percent of increase
JG	16. 8	123	9. 7	113
HB	15. 0	123	8. 3	113
RM	59. 0	177	40. 0	153

TABLE 3

Systolic Blood Pressure Changes before and after Exercise with Test Wheel Chairs

Subject	Conventional		Experimental	
	Net difference	Percent of increase	Net difference	Percent of increase
JG	15	112	5	104
HB	13	112	13	112
RM	23	120	16	114

TABLE 4

Pulmonary Ventilation

Subject	Liters air/(kg.) (m.)		Liters air/(kg.) (min.)			
	Conventional	Experimental	Conventional	Percent of increase over resting	Experimental	Percent of increase over resting
JG	0. 0021	0. 0018	0. 166	143	0. 155	129
HB	. 0019	. 0017	. 150	182	. 126	147
RM	. 0025	. 0024	. 250	322	. 219	292

With each stroke to the wheel, the subject applied a reaction force against the backrest of the wheelchair, producing a clockwise movement around the rear axles. A resisting counterclockwise moment was then produced by the weight acting at the CG a certain distance in front of the rear axles. The shorter the resistance moment arm, i.e., the distance between the CG and the rear axle, the smaller the moment and, consequently, the smaller the clockwise moment it would resist (Fig. 7).

The key factor in this matter was the location of the center of gravity. As shown in Figure 8, the CG of the unloaded lightweight chair was more than ¾ in. farther forward than in the conventional chair. When the light chair was loaded, however, the CG was approximately ½ in. closer to the rear axles. Since the light chair was also approximately 20 lb. lighter than the conventional chair, the superimposed CG of the passenger raised the CG of the total system even higher. These two factors combined to make the lightweight chair less resistant to the clockwise reaction forces, and hence less stable. The conventional chair, therefore, could resist a higher toppling moment; this would indicate that it could probably climb a steeper hill with less toppling hazard than the lightweight chair.

e. Metabolic Cost. Use of the light chair did not offer any clear or substantial advantage in energy cost of propulsion. The energy cost of using the light chair was slightly higher than the conventional chair in one case, slightly lower in the second case, and substantially lower in the third. Use of the conventional chair caused significantly greater percentage of increase over resting rate in two cases and a negligible decrease in the third case (Table 5).

The similarity in caloric cost per meter moved indicated that each subject tended to adopt an optimal speed and operating pattern at a minimal energy requirement level for each chair.

The average energy expenditure of operating each of the chairs was calculated from measures of oxygen consumed. Net energy input rate was computed by:

$$\frac{(\text{Work } O_2 + \text{Recovery } O_2) - [(\text{Resting } O_2/\text{min.})(\text{work time} + \text{recovery time})]}{\text{Work time}}$$

This formula takes into account both aerobic and anerobic processes and expresses the oxygen utilization rate for the activity. The oxygen data were converted to units of energy (Calories) by using an RQ (respiratory quotient) for a mixed diet of 0.85. These data

were also expressed per unit of body weight to compensate for individual differences. As the subjects performed at "optimal" speeds, the data were also expressed in Calories per kilogram per meter.

Nevertheless, one might intuitively expect the 20-lb. difference in weight to have a more significant influence. Analysis of the pattern of wheelchair use, however, shows that the energy required to accelerate a heavier chair is greater than for a lighter chair, but that once accelerated to a nearly uniform velocity, only increments of energy input are required to supply the constant energy drain of friction. The frictional forces, neglecting air resistance, cause losses of energy in three areas, at the tire-floor interface, within the tire material, and at the axles. During a period of approximately uniform speed, therefore, the energy required to maintain the motion is related to the frictional forces rather than to the weight of the system. If the retarding forces of friction for both chairs were similar and consumed a high fraction of the input, energy costs for sustained propulsion would be approximately the same. The advantage of weight reduction is accrued only in starting and stopping, i.e., in gross velocity changes.

To check this theory, recordings were obtained of velocity changes throughout the entire system as the subject propelled the conventional and light chairs (Fig. 9). The underlying rationale was that of the total energy put into the system by the subject, part would be converted to kinetic energy and the rest would be lost in friction. During the glide, energy lost to friction continues to leave the system with the consequent loss of velocity and kinetic energy.

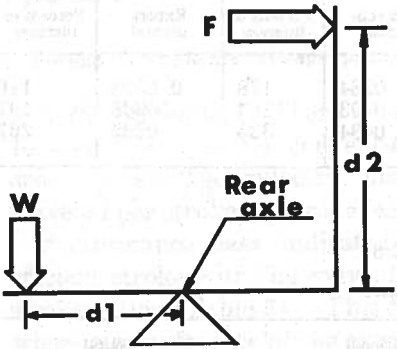


FIGURE 7. *Schematic of moments of forces on wheelchair as subject applies stroke to driving wheels. (Clockwise moment=F×d₂; counterclockwise moment = W×d₁; as d₁ diminishes, the counterclockwise moment diminishes.*

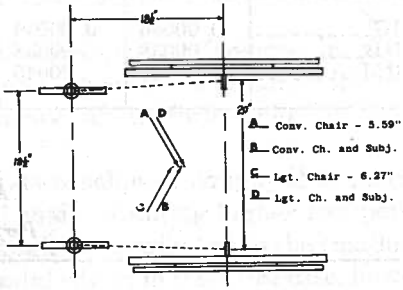


FIGURE 8. *Schematic showing change of CG forward with increase of weight in wheelchair. (Conv. = conventional; Lght. = light; Subj. = subject.)*

To consider the influence of the mass moment of inertia of the wheels, which were smaller and lighter in the light chair (21.75 vs. 23.5 inches in diameter and 3.5 lb. vs. 7.0 lb.), the chairs were compared with respect to total change in kinetic energy and rate of loss of kinetic energy over several typical cycles. Kinetic energy was calculated by:

$$KE = \tfrac{1}{2}mv^2 + I\varpi^2 \qquad (1)$$

where $\tfrac{1}{2}mv^2$ was the translation energy of the system and $I\varpi^2$ the additional rotational energy stored in the wheel. For simplicity, the radius of gyration was assumed to be the same as the rolling radius of the drive wheels, and the rotational effect of the small casters was neglected. The KE decay rate was taken as KE at maximum velocity minus the KE at minimum velocity, divided by the decay time during a cycle.

$$KE \text{ decay rate} = \frac{\tfrac{1}{2}(M+4m)(V_m^2 - V_n^2)}{t_g} \qquad (2)$$

where M is the mass of the man-chair system minus the wheels, m is the mass of the wheels, V_m is maximum velocity, V_n is minimum velocity, t_g is the time of the glide.

TABLE 5

Average Energy Cost

Subject	Cal./(kg.) (m.)		Cal./(kg.) (min.)			
	Conventional	Experimental	Conventional	Percent of increase	Experimental	Percent of increase
JG	0. 00036	0. 00039	0. 0284	178	0. 0303	140
HB	. 00038	. 00038	. 0293	183	. 0268	191
RM	. 00046	. 00045	. 0484	325	. 0392	267

TABLE 6

Performance

Factor	Conventional			Experimental		
	JG	HB	RM	JG	HB	RM
Velocity (ft. per min.)	254. 2	252. 5	281. 7	257. 2	234. 1	297. 4
Average lap time (sec.)	41. 5	40. 9	36. 7	40. 3	45. 0	34. 2
Ft. per stroke	6. 9	6. 3	5. 9	5. 4	4. 9	6. 5
Strokes per lap	24. 8	27. 3	29. 2	31. 6	34. 8	26. 4

The total energy losses (ΔKE) during the glide for the conventional chair in two typical cycles were 20.4 ft.-lb. and 15.3 ft.-lb. The light chair lost 17.7 and 17.9 ft.-lb. in two similar cycles. The *rate* of loss for the conventional chair was 48.4 ft.-lb. per sec. and 34.9 ft.-lb. per sec.; for the light chair 38.5 ft.-lb. per sec. and 44.7 ft.-lb. per second.

Although these data indicated more variability in the conventional chair (perhaps the result of greater use and wear of components), they did not reveal significant differences in frictional characteristics.

The reduction in weight was significant for wheelchair use in small confined areas and for patients with shoulder girdle and upper extremity involvement. A reduction in frictional requirements would be more helpful, however, for vigorous patients since their patterns of operation would probably include longer periods of sustained propulsion.

The available data indicated that wheelchair propulsion between 1.2 and 3.5 m.p.h. was less energy-consuming than normal walking, and substantially lower than above-knee amputee locomotion (Fig. 10).

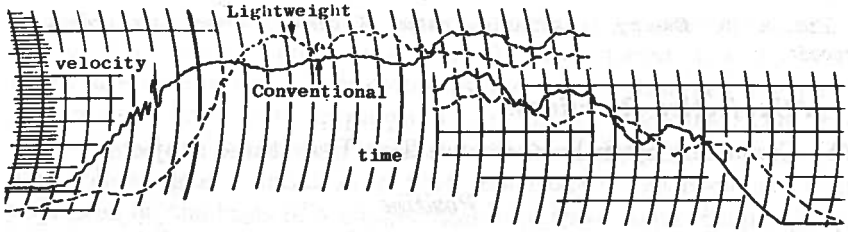
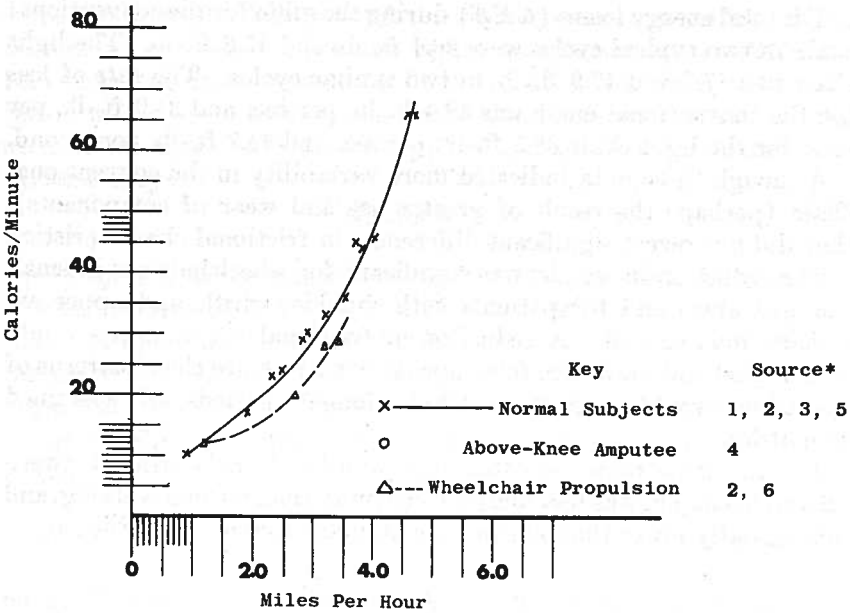


FIGURE 9. *Velocity change pattern of subject-propelled wheelchairs.*

f. Performance. There were no consistent differences in performance between the conventional and experimental chairs with respect to average velocity, maintenance of uniform velocity, feet traveled per stroke, or stroke length.

Performance data indicated that two subjects employed a more efficient stroke with the conventional chair, attaining higher feet-per-stroke ratios (Table 6). This fact might be attributed to the smaller wheel circumference of the experimental chair; in the third case, however, the opposite was true. Furthermore, stroke efficiency was not related to velocity since the subject with lowest average velocity had the highest ft.-per-stroke ratio. This result seemed to be influenced by the stroke and glide pattern adopted by each subject.

Any advantages to be accrued from the decreased weight of the experimental chair were not clearly reflected in these factors.



*See Reference Key

FIGURE 10. *Energy consumption rates at various wheelchair propulsion speeds.*

g. Subjective Reactions.

(1) Comments after Performance Test Use (three subjects):

Positive

A. Comfort

1. Easy to enter and reasonably comfortable to sit.
2. Solid drive wheel grip improves grasp and stroke.

B. Stability

1. Seems about as stable as any other chair at rest.

C. Effort

1. Seems easier to move.
2. Much easier to push, much easier to make turns, and to stop.
3. Easier to push, turns easily.
4. Easier at high speeds, long distances.

D. Weight

1. Feels much lighter.

Negative

A. Comfort

1. Seems awkward—too wide and too short.

2. Too narrow, not as comfortable.
3. Wheel rim too narrow to grasp easily.
4. Chair arms too high—hits medial side of arm during stroke.

B. Stability

1. Seems to be "back heavy" when driving; front end tips up.
2. Not as stable as conventional model when propelling.
3. Senses CG displacement rearward.
4. Fore and aft stability is reduced when pushing hard, requireing a fairly critical body position.

C. Effort

1. Top of wheel rim too low, requiring longer reach and reducing stroke length.
2. Grasps top of wheel with extended elbow, a poor mechanical position for application of force.
3. Light weight does not permit him to push as hard as he would like.

D. Miscellaneous

1. Aluminum driving rim stains hands.

(2) Use in Daily Activities:

Two veterans, one traumatic quadriplegic (incomplete C4 and C5 lesion) and one paraplegic (L5 lesion), participated in a daily activity use evaluation. Both subjects had homes designed for paraplegics, and both were employed in an office. Each used the light chair for 4 days at home, to travel to and from the jobs, and at work. The following is a collation of their reactions. The responses were requested on the basis of a comparison with their conventional chairs, but the questions were otherwise unstructured.

Advantages of light wheelchair:

1. Transferring chair in and out of car (both subjects agree).
2. Folding and opening chair (both subjects).
3. Turning at right angles (one subject positive; one neutral).
4. Up curbs (one subject positive; other subject negative).

Disadvantages of light wheelchair:

1. Pushing up incline (both subjects positive).
2. Descending curb (both subjects positive).
3. Pushing on level ground (one subject positive; other subject neutral).

Same with both chairs (both subjects positive):

1. Chair-to-car body transfer.
2. Bed-to-chair body transfer.
3. In and around bathroom.
4. Opening and closing door.

The two advantages about which both patients agreed were directly related to the lighter weight of the test chair. These findings were corroborated in a third subject who was observed as he folded and unfolded the chair and as he transferred in and out of his car.

The two disadvantages expressed by both subjects were related to the anterior-posterior location of the center of gravity, which fell closer to the rear axle in the light chair. In this connection, one subject reported the need to lean forward in ascending a slight incline in order to maintain balance, and he pointed out that it was harder and more awkward to push in this position. The other subject reported falling several times when descending a curb.

SUMMARY

1. Using the light chair produced after-exercise increases in pulse rate and substantial decreases in systolic blood pressure.
2. The light chair required lower (0.4 lb. less) starting forces.
3. Subjects found the light chair easier to put in motion; easier to fold and to place in a car.
4. Use of the light chair did not result in significant reductions in caloric cost.
5. The light chair offered no advantages in average velocity attained or stroke efficiency.
6. The light chair was less stable when accelerated under load due to its CG location.
7. Subjects criticized the small wheel diameter and instability in descending curbs and ascending inclines.
8. The light chair did not conform to specifications with respect to seat and back material, frame and wheel material, and wheel diameters.
9. The crosspieces of the frame buckled after 38 cycles of rolling off a 6-inch platform with a 200-lb. load.

RECOMMENDATIONS FOR IMPROVEMENT

1. Move CG forward by using heavier front wheels, weighting the chair, or improving the distribution of weight by redesign of the suspension or seating arrangement.
2. Consider larger-diameter driving wheels.
3. Coat aluminum handgrip to prevent staining of hands.
4. Increase strength of crosspiece of frame.

APPENDIX

The use of new materials, alignment concepts, and increasingly sophisticated control mechanisms has renewed interest in energy requirement data for evaluation and prescription purposes, particularly for severely handicapped or geriatric patients. Several laboratories have recently published valuable energy-cost data. Unfortunately, the information from different sources is not directly comparable, limiting its usefulness for other workers.

Much of the difficulty stems from the variety of units and terms used in reports. Data is often expressed in liters or cubic centimeters of oxygen utilized, in caloric equivalents, in rates per minute or per meter traveled, or per square meter of body surface. Often it is not clear whether the units are gross or net figures, or whether recovery oxygen has been included. Juggling between the metric and the English systems is frequently necessary if a comparison of data is desired. Since each worker selects the units and expressions most appropriate for his study, this situation will undoubtedly continue for some time. There is a need, therefore, to collate the most pertinent data on various methods of human locomotion and to express them in comparable terms. The available sources of energy-cost data include the classicists such as Benedict and Hill, and more recently, Erickson, Keys, Simonson, and Morehouse. The National Academy of Sciences has also published data on energy cost of progression in the *Handbook of Respiration* (3). Gordon, Ralston, and others have also contributed to this growing body of literature. In Table 7 energy-cost data have been expressed in terms of net Calories per minute. The original data were published in different forms, making a direct comparison without conversion impossible. While it may be more desirable to present these data in terms of body weight or other units to account for significant individual differences, the conversion information was not always available. The choice, therefore, was between delimiting the series or using an average figure for this purpose; net caloric cost per minute, the highest common denominator, was chosen.

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DISCUSSION OF EVALUATION TECHNIQUES

The approach to these evaluation problems as outlined seems to provide a generally adequate basis for intelligent appraisal. The selected parameters cover appropriate areas of inquiry and can be expected to yield information for guidelines of procurement and to serve as a valuable feedback to manufacturers for improvement and development. Although these *areas* of investigation are reasonably satisfactory, present *methods* require further refinement and the development of new techniques.

There is a need for more adequate standards for materials, dimensions, and durability as well as mechanical standards based upon the best materials available and the principles of good design. It is imperative to develop functional standards, including criteria based on the purpose of a device, the capabilities of the user (patient or other), and the environment in which it is to be used. Unfortunately, available standards are primarily descriptive and were patterned on preexisting devices.

There is a paucity of data on those aspects of human performance that are directly related to the use of wheelchairs, lifts, and similar devices. There is, of course, reference material pertaining to the normal subject on physical constants, ranges of motion, rates of work, and the like. Similar data for various categories of the handicapped are needed.

The currently used tachograph, for example, lacks versatility, is inconvenient to attach to a subject, and interferes with performance. The accelerometers available were not designed for these purposes and require extensive adaptation. More accurate devices are needed to regulate and measure velocities and accelerations of device components and patient body segments, under dynamic conditions.

Classical metabolic techniques for measuring metabolic energy expenditure need further refinement. Small, but sometimes significant, differences in human performance cannot be discriminated accurately because *average rates* of O_2 utilization rather than continuous measures are used.

In converting O_2 consumed to its caloric equivalent, respired air is analyzed for CO_2 as well as O_2 to compute the respiratory quotient. It is becoming common practice arbitrarily to assign a mixed diet RQ of 0.85. The error introduced may be slight under basal conditions but since most energy-cost experiments are conducted with subjects in a resting rather than basal condition, the error may be significant.

TABLE 7

Relative Net[1] Metabolic Cost of Locomotion

Activity	Speed (m.p.h.)	Energy expenditure (Cal./min.)	Reference
Lying, Basal	—	1.0	2
Lying, Basal	—	1.2	3
Sitting, resting	—	1.2	2
Sitting, resting	—	1.3	4
Sitting, resting	—	1.8	3
Standing relaxed	—	1.4	2
Standing relaxed	—	1.4	1
Standing relaxed	—	2.0	3
Walking.	0.9	1.0	1
	1.9	1.7	1
	2.3	2.3	5, 3
	2.5	2.4	2
	2.8	2.9	1
	2.9	3.0	5
	3.2	3.3	5, 3
	3.5	3.6	5, 3
	3.7	4.5	1
	3.8	4.4	2
	4.0	4.6	5
	4.6	6.6	5, 3
	4.7	6.6	1
Treadmill walking.	2.3	2.2	4
	2.5	3.0	3
	3.8	4.4	3
	5.0	8.3	3
Treadmill walking with long leg brace knee-locked.	2.3	2.9	4
A/K amputee walking with alignment of knee altered.	2.4 (Stable alignment)	4.5	4
	2.4 (Optimal alignment)	4.9	4
	2.4 (Trigger alignment)	5.1	4
A/K amputee walking with center of gravity (CG) and mass moment of inertia (I) of shank altered.	2.4 (I minimal, CG raised).	5.0	4
	2.4 (I normal, CG raised)	5.0	4
	2.4 (I normal, CG normal)	5.1	4
	2.4 (I maximum, CG normal).	5.5	4
	2.4 (CG lowest, I maximum).	5.9	4
Wheelchair propulsion.	1.2	1.2	2
	2.7	2.0	6
	2.9	1.7	6
	3.2	2.8	6
	3.4	2.9	6
Crutch and brace ambulation.	3.2	6.8	2

[1] Where original locomotion data were given in gross figures, 1.2 Cal. per min. (an average sitting metabolic rate) was subtracted.

Although much more convenient to use, resting metabolic rates introduce other sources of errors into energy-cost estimates. Findings at the VAPC Bioengineering Laboratory and other laboratories engaged in collecting energy-cost data indicate a high variability in the resting rates among subjects. An individual's resting rate may exhibit day-to-day changes, diurnal changes, or fluctuations that result from ingested food (specific dynamic action of food).

The whole concept and character of the steady state needs to be redefined. VAPC studies indicate a considerable fluctuation in the amounts of O_2 intake for each minute of exercise, casting doubt on the accuracy of data based on steady-state phenomena.

The utility of cardiac response measured as a pulse rate in excess of the resting rate is also questionable. A reliable method for monitoring the entire work activity, thus avoiding complete dependence on the response obtained during the first few seconds of the recovery period, would provide a much superior basis for evaluating stress.

In order to compare performances of upper extremity amputees, lower extremity amputees, and normal subjects, comparable work loads should be employed. At the present time there is no useful way of prescribing comparable work dosages in different activities.

Physiological measures alone may not always be the most useful way of expressing the energy exchanges during the use of a mechanical device. Mechanical parameters of human performance with a device similar to a wheelchair, for example, are being measured with comparatively simple instrumentation.

Figure 11 is an illustration of a tachograph recording of speed versus time. The man applies power to the vehicle, in the case illustrated, during four successive stroke periods (heavier line in curve). Short gliding periods occur between strokes, a natural pattern for reciprocating hand and arm movements where the effort is applied in one direction only. After four strokes, the vehicle is allowed to coast to a halt under the influence of friction (glide period). The mechanical energy input to the system by the man shows up as the *kinetic energy* of the system and is dissipated by friction during *all* the time motion occurs. To compute the total mechanical energy input, we can sum the contributions the man makes to the realized kinetic energy and the energy he supplies simultaneously to "feed" the concurrent frictional losses. These relations are calculated as follows:

$$E = (\text{kinetic energy}) + (\text{frictional energy})$$

$$E = \tfrac{1}{2}M(v_f^2 - v_i^2) + \tfrac{1}{2}M\left[\left(v_f^2 - v_i^2\right)\frac{t_1}{t_2}\right] \qquad (3)$$

where E = total mechanical energy input per cycle; M = a composite "inertial constant" to account for system mass and rotational inertia;

v_f=maximum velocity during a cycle (assumed to coincide with end of manual effort); v_i=initial velocity for the cycle; t_1=time of application of manual effort (see v_f above); t_2=time required for the system to lose velocity from v_f *to* v_i under the restraining influence of friction.

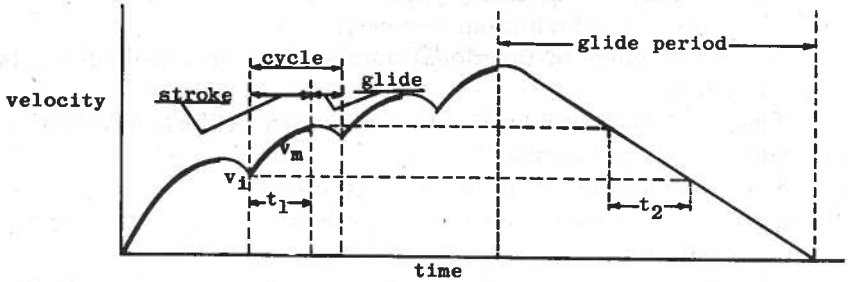

FIGURE 11. *Tachograph (speed vs. time) recording of subject propelling conventional and lightweight wheelchairs. (Heavy curve indicates 4 successive stroke periods; thin curve indicates short gliding periods between strokes.)*

A summation of the E for each cycle will give total energy input for the operation being considered. The ratio t_1/t_2 is applied to the frictional energy loss occurring during t_2, on a *pro rata* basis, to estimate the energy loss to friction during t_1 when the velocities are similar; however, the dissipative process may be occurring in more or less time. Inherent in this analysis, of course, is a number of assumptions and idealizations of the real world. Valid use was made in this first approximation of such concepts as rotational inertia, friction, and duration of effort.

It is also possible to simplify Equation (3) to:

$$E=\tfrac{1}{2}M(v_f^2-v_i^2)\left(1+\frac{t_1}{t_2}\right) \tag{4}$$

Then we can consider this the flow of energy from the man into the man-chair system, and from the man-chair system into the environment through the dissipative processes grouped together above as "friction." Inasmuch as the entire input is finally dissipated as friction, an alternative way to calculate the total E input would be to compute and account for all frictional energy during the operation. (Note that the system starts at rest and returns to this condition at the end of the period.)

A comparison between this mechanical energy input and the metabolic energy cost to achieve the input can be made by relating them in an index of efficiency. For example, the efficiency of the man can be expressed by

$$\text{Efficiency}=\frac{\text{output}}{\text{input}}$$

where the output is the mechanical energy produced and the input is the metabolic cost. Although several methods for describing efficiency have been used, there is little general agreement on the most appropriate units and terms of expression.

In the Bioengineering Laboratory several studies are being considered to improve and refine our methods:

1. The development of functional standards for orthopedic lift aids and wheelchairs.
2. The compilation of basic data on human performance related to the use of orthopedic aids.
3. The development of an instrument utilizing accelerometers to measure instantaneous velocities and accelerations of body segments and orthopedic devices.
4. The re-evaluation of classical energy expenditure data-collecting methods and the assumptions underlying them. In this connection, the VAPC Bioengineering Laboratory is preparing reports on (1) steady-state phenomena and (2) the effects of assuming a respiratory quotient figure when calculating energy-cost data. By the use of telemetric devices, relationships between cardiac response recorded during and following the exercise period and the energy expended for that activity may be established.
5. The reliability of the resting metabolic rate and the establishment of work dosage units.

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APPENDIX REFERENCES

(See Figure 10)

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