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Variation of mechanical energy levels for normal and prosthetic gait

H. LANSHAMMAR

Department of Automatic Control and Systems Analysis, Institute of Technology, Uppsala University, Sweden and National Board of Occupational Safety and Health, Sweden

Abstract

Mechanical energy levels were investigated for normals and for below-knee amputees during level walking. The weight of the prostheses was varied by attaching 0·5 kg extra weight to the prostheses.

The measurements and analyses were made with the ENOCH system consisting of a minicomputer (HP 21 MX), an optoelectronic device for displacement data measurement (Selspot) and a force plate (Kistler) for measurement of ground reaction forces.

Results by Winter et al (1976) on the energy changes during normal walking obtained from displacement data on one leg only were verified using data from both legs and the trunk.

For the amputees it was concluded that the energy changes increased for the prosthetic shank when the weight increased. For the other body segments and for the body total no significant differences were found.

Introduction

The energy expenditure during walking is an important parameter for the evaluation of human gait. Ralston and Lukin (1969) found a fairly constant ratio between the metabolic expenditure and the positive work for subjects during normal walking and walking with extra trunk and foot loading. Therefore the mechanical energy changes can be used to get an indication of the energy requirements during walking. They defined the positive work from the changes in total mechanical (kinetic plus potential) energy of the body.

In recent years the mechanical energy levels during walking have been studied by Winter et al (1976) and Cappozzo et al (1976) among others. In both these investigations displacement data for one leg were used to describe the motion of the whole body.

For the construction of lower extremity prostheses the weight is an important design parameter. The optimum weight of the prostheses have been a subject of debate.

Inman (1967) claimed that a prosthesis should not be made too light because with a lighter prosthesis the amputee develops less kinetic energy at the end of swing phase to be fed back into the body to maintain his forward velocity.

Quigley et al (1977) reported on the oxygen consumption during walking with ultra-lightweight and standard BK prostheses. The trend was toward a higher oxygen consumption per metre and kilogram body weight with the heavier type of prosthesis.

The investigation presented in this paper was undertaken to test the effect of a small increase in the weight of BK prostheses on the mechanical energy levels during walking. Gait data from both legs and the HAT (head, arms, trunk) for normals were also analyzed to test the results of Winter et al (1976) and Cappozzo et al (1976).

Patients and methods

A minicomputer based system—called ENOCH—was used for the measurements and analyses. In this system, described by Gustafsson and Lanshammar (1977), an optoelectronic device, Selspot, is used for kinematic data collection and ground reaction data are obtained

All correspondence to be addressed to Hakan Lanshammar, D. Eng, Department of Automatic Control and Systems Analysis, Institute of Technology, Uppsala University, Box 256, S–751 21, Uppsala, Sweden.

from a Kistler force plate. Output of result diagrams are made on a graphic computer terminal with a hardcopy unit or in tabular form on a line printer.

Two Selspot cameras were used to obtain kinematic data for both legs. Landmarks (light emitting diodes) were placed on the shoulder, hip joint, knee joint, ankle joint, heel and toe base for both sides. The measurement area was approximately 3 × 3 metre allowing for the registration of 3 steps in each measurement. Data was collected at the rate of 158 Hz. The standard deviation of the measurement noise was 0·002 m and the systematic coordinate error was estimated to be less than 0·02 m after correction of errors related to floor reflections, lens distortion and detector characteristics (Gustafsson & Lanshammar, 1977).

The displacements of the centre of mass for the different body segments in the model, HAT, thighs and shanks, were calculated from the measured coordinate data. The required body segment parameters were obtained according to the method described in Gustafsson and Lanshammar (1977) which is based on data from Drillis and Contini (1966), Contini (1970), Contini (1972) and Chandler et al. (1972). The velocities of the different body segments were calculated by numerical differentiation of the displacement data with a method described in Gustafsson and Lanshammar (1977).

Angular velocities for the body segments were obtained from the linear displacement and velocities by straightforward application of trigonometric relations.

The energy levels for the body segments were calculated as

$$ET_i = 0{\cdot}5 \cdot m_i \cdot v_i^2 + 0{\cdot}5 \cdot j_i \cdot w_i^2 + m_i \cdot g \cdot h_i$$

where m_i is the segment mass, j_i is the rotational moment of inertia about the centre of mass, v_i is the linear velocity, w_i is the angular velocity and h_i is the vertical coordinate for the mass centre of the ith segment.

It should be noted that only planar motion was included in the analysis. Of course this restricts the validity of the results, especially for pathological gait. Further the shank and foot was treated as one rigid body. The total body energy was calculated as the sum of the energy levels of the five body segments.

Measurements were made on three male persons with below-knee prostheses and on two male normals. Data on the subjects are given in Table 1. All the amputees were using TPJ prostheses. The TPJ (Torsten Pettersson Jigg) system is a Swedish method for fabrication of light-weight BK prostheses.

For the amputees gait measurements were made with their normal prostheses but also when 0·5 kg extra weight had been applied to the prostheses. The extra weight was in the form of lead plates attached with a belt around the centre of mass of the prosthesis.

Table 1. Subject data

SUBJECT	SEX	AGE	WEIGHT (kg)	HEIGHT (M)	PROSTHESIS	NO. OF MEASUREMENTS	
						WITHOUT EXTRA WEIGHT	0.5 kg EXTRA WEIGHT ON THE PROSTHESIS
JK	M	46	55.5	1.68	TPJ (1.52 kg)	7	8
EL	M	41	74.5	1.77	TPJ (1.45 kg)	8	8
DR	M	26	87.5	1.77	TPJ (1.38 kg)	6	2
HL	M	29	72.5	1.91	NORMAL	1	
KÖ	M	34	73.1	1.84	NORMAL	1	

Before the first measurement the subjects were allowed approximately 15 minutes of gait training to get used to the new prosthetic weight.

The subjects were allowed to choose their own comfortable step rate. It was approximately 75 steps/min for JK, 105 steps/min for EL, 90 steps/min for DR, 90 steps/min for HL and 95 steps/min for KÖ. A metronome was used to help the subjects keep a constant step rate during all the measurements. The stride lengths were approximately 1·30 m for JK, 1·50 m for EL, 1·25 m for DR, 1·50 m for HL and 1·65 m for KÖ.

Errors

The most important error contribution to the calculated energy levels is the segment masses. According to Gustafsson and Lanshammar (1977), the standard deviation of these errors is approximately 10 per cent. Compared to this figure the errors in the kinematic variables are very small during most of the gait cycle. The standard deviation of the stochastic error in the linear velocities was calculated to be less than 0·01 m/s (Gustafsson & Lanshammar, 1977), which is below 1 per cent of the mean velocity. The systematic error in linear velocities due to the limited bandwidth of the differentiating filter is also very small during most of the gait cycle.

However, close to heel strike the coordinates have high frequency components which can result in substantial errors in calculated derivatives. This can be the reason for the large errors reported in Cappozzo et al (1976) concerning a comparison between the muscular work and the total body energy variations during the heel strike phase.

The rotational component of the energy is much smaller than the translational and therefore the error in this term is not so important. In the term for the potential energy finally, the error in the vertical coordinates of the centres of mass are at most a few centimetres, which again gives much smaller error contribution as compared to the 10 per cent expected error in the segment masses. Note especially that for the HAT, an error in the location of the mass centre will not effect the changes of its potential energy values very much. The only effect of such an error is that the level around which the changes take place will be increased or decreased.

To summarize, the expected error in the calculated energy levels is approximately 10 per cent except during the heel strike phase where the error can be larger.

When energy changes are calculated by subtraction of energy levels the situation is changed, especially for the potential energy of the trunk. Since the vertical displacement of the centre of mass for the HAT is only about 0·05 m, a coordinate error of 0·01 m due to skin movements for example, can result in a 20 per cent error in the potential energy change. For the kinetic energy of the HAT and for both kinetic and potential energy of the other body segments the relative energy changes are much larger and the results for energy levels are essentially applicable also for the energy changes.

Results

In Figures 1–4 statistics for the resulting energy changes are presented. The values for the shank and thigh represent the difference between the maximum energy level during swing phase and the minimal energy level found for the following stance phase.

For the HAT and the total energies, the values are the difference between the maximum energy value during the first measured swing phase and the minimum value that is obtained close to the next heel strike, plus the same difference for the following step.

All energy values have been normalized by dividing with the current stride length and the subject's body weight. The middle line of the three horizontal lines on top of each bar represents mean value and the two others give the standard deviation for all measurements on the subject. For comparison the corresponding results for two male normals are also given in the figures together with results taken from Winter et al (1976). In the latter case the subject was a 20 year old female (subject 1 in Winter et al, 1976). From the Figures 1–4 it is clear that there is a good correspondence in the energy changes for all segments between the normal data presented here and the results obtained by Winter et al (1976). The results presented in Cappozzo et al (1976) can be compared only for the HAT and total body energy changes, because shank and thigh energy values are not presented separately. For the HAT the energy changes for one normal subject (U.D. in Cappozzo et al, 1976) corresponded to 0·1 J/m/kg, while it was approximately 0·2 J/m/kg for the data presented in Figure 3. The total body energy changes in Cappozzo et al (1976) for the subject U.D. was 0·4 J/m/kg which concords well with the results in Figure 4.

The energy changes reported in Winter et al (1976) and in Cappozzo et al (1976) were obtained using some simplifying assumptions. Data was collected for one leg only while the trajectory of the other leg was assumed to be the same displaced in time by half a stride period. Further the trajectory of the centre of mass of the trunk was assumed to be the same as the average of the trajectories of the greater trochanter marker for the left and right leg. Rotation of the trunk was not included in the analysis.

Especially the first of these simplifications could be expected to give considerable errors in the total body energy variations. This is because the shank energy is rapidly decreasing prior to heel strike while it is rapidly increasing shortly thereafter for the other leg. Therefore the minimum value of the total body energy, which occurs shortly before heel strike, can be heavily affected if the assumed time displacement in the data for the two legs are erroneous.

In the present investigation none of the mentioned simplifications were made.

Shank

The energy changes found for the shank on the normal leg of the amputees were comparable to those obtained for the normals (Figure 1). For the prosthetic side however the energy changes are considerably lower for the amputees. The reason for this difference is that the weight of the prosthesis is much lower than the weight of the normal leg. Since the displacements of the prosthetic shank were approximately the same as those for the normal side this resulted in a smaller peak energy value during swing and hence a smaller energy change was obtained.

With 0·5 kg extra weight on the prosthesis the energy changes on the prosthetic shank were increasing significantly for all subjects. This increase was approximately proportional to the weight increase, which is what to expect when the gait pattern is not changing. For the normal leg no significant change was found.

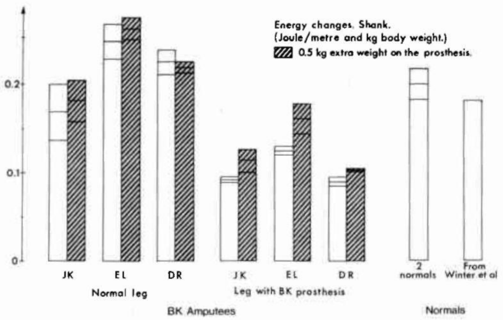

Fig. 1. Energy changes for the shank. The horizontal lines on each bar indicate mean value and standard deviation. The number of measurements involved is given in Table 1.

Thigh

The magnitude of the energy changes for the thighs were comparable between the amputees and the normals (Figure 2). This applies for both legs of the amputees.

When 0·5 kg extra weight was applied on the prostheses no statistically significant changes were found. For two of the subjects the values decreased slightly while they increased for one subject.

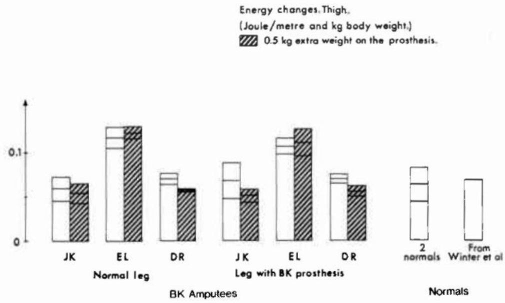

Fig. 2. Energy changes for the thigh.

HAT

Also for the HAT the energy changes for the amputees were close to those obtained for normals (Figure 3). The values did not change significantly when 0·5 kg extra weight was added to the prostheses.

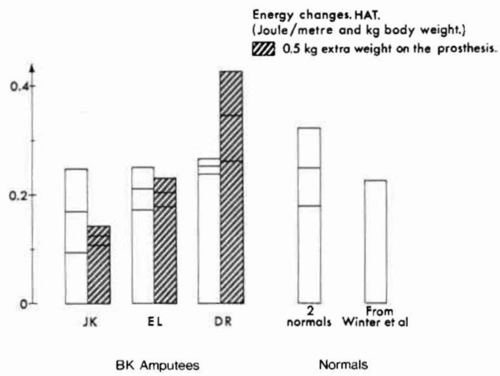

Fig. 3. Energy changes for the HAT (head, arms, trunk).

Total energy changes

The total energy changes are illustrated by Figure 4. Since the maximum values of the energy levels for the different body segments do not coincide in time the changes in the total energy level were always less than the sum of the segments energy changes.

In the total energy changes no significant difference was found when 0·5 kg extra weight was added. The values for the normals were also comparable to those for the amputees.

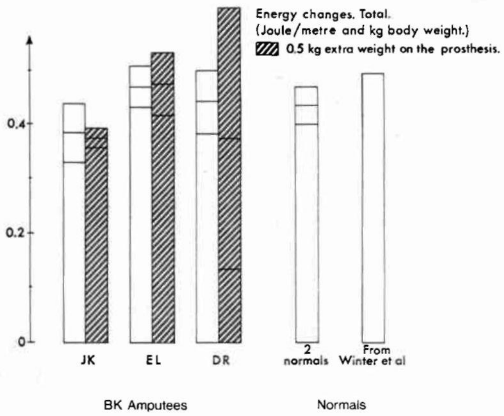

Fig. 4. Energy changes for the whole body.

Discussion

When 0·5 kg extra weight was added to the prostheses it was found that the energy changes for the prosthetic shank increased significantly for all subjects. For the other segments and for the total energy changes no significant effect of the extra weight was found.

Suppose that the energy changes of the prosthetic shank increase by the amount we have observed as a response to the extra weight, while the energy changes of the other segments are not influenced at all. Then there would be an increase in the total body energy changes, even if it would not be large enough to be statistically certified for the limited number of trials in this investigation.

However from our experiments with varying prosthetic weight there is not even a slight trend in the results on total body energy changes. It remains to verify if this result is purely coincidental, an effect of the stochastic nature of the human gait, or if there are any compensatory reactions to the increased prosthetic weight which make the total body energy changes rather insensitive to such weight changes.

Obviously the results presented here on the total body energy changes are not consistent with the results of Quigley et al (1977) who reported a higher oxygen consumption with a heavier type of prosthesis. Further research will be required to decide whether the above mentioned results of Ralston and Lukin (1969) do not apply to the present case, or if there is some other explanation of this discrepancy.

Conclusion

The results presented in Winter et al (1976) on energy changes during normal locomotion were verified using gait data for both legs and for the HAT (head, arm, trunk). In contrast to this Winter et al (1976) as well Cappozzo et al (1976) used displacement data for one leg only.

The results from Cappozzo et al (1976) were verified for the total body energy changes, while the values given for the HAT in Cappozzo et al (1976) were about half of those in the present study. For the shank and thigh no data was given in Cappozzo et al (1976).

Energy changes for BK amputees were also studied. It was concluded that for all body segments except the prosthetic shank, the results were quite similar to those obtained for normals. For the prosthetic shank the energy changes were much smaller than for a normal shank. This is explained by the light weight of the BK prosthesis as compared to a normal leg while the trajectories of the prosthesis were similar to those of the normal shank.

When the weight of the prostheses was increased by 0·5 kg the energy changes for the prosthetic shanks increased significantly. For the other body segments and for the total body no significant changes were noted.

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REFERENCES

CAPPOZZO, A., FIGURA, F., MARCHETTI, M. (1976). The interplay of muscular and external forces in human ambulation. *J. Biomechanics,* **9(1),** 35–43.

CHANDLER, R. F., CLAUSER, C. E., MCCONVILLE, J. T., REYNOLDS, H. M., YOUNG, J. W. (1975). Investigation of inertial properties of the human body. DOT HS-801 430.

CONTINI, R. (1970). Body segment parameters (Pathological). Technical Report No 1584.03. School of Engineering and Science, University of New York.

CONTINI, R. (1972). Body segment parameters, Part II. *Art. Limbs,* **16(1),** 1–19.

DRILLIS, R., CONTINI, R. (1966). Body segment parameters. Technical Report 1166.03. School of Engineering and Science, University of New York.

GUSTAFSSON, L., LANSHAMMAR, H. (1977). ENOCH—An integrated system for measurement and analysis of human gait. UPTEC 7723R. Institute of Technology. Uppsala University, Uppsala, Sweden.

INMAN, V. T. (1967). Conservation of energy in ambulation. *Arch. Phys. Med. Rehab.* **48(9),** 484–488.

QUIGLEY, M. J., IRONS, G. P., DONALDSON, N. R. (1977). The Rancho ultralight below-knee prosthesis. Rehabilitation Engineering Center at Rancho Los Amigos Hospital, Downey, California.

RALSTON, H. J., LUKIN, L. (1969). Energy levels of human body segments during level walking. *Ergonomics,* **12(1),** 39–46.

WINTER, D. A., QUANBURY, A. O., REIMER, G. D. (1976). Analysis of instantaneous energy of normal gait. *J. Biomechanics,* **9(4),** 253–257.